# Programming Cochlear Implants

SECOND EDITION



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Jace Wolfe
Erin C. Schafer

A Volume in the Core Clinical Concepts in Audiology Series

Plural Publishing Inc.

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**Library of Congress Cataloging-in-Publication Data**

Wolfe, Jace, author.
Programming cochlear implants / Jace Wolfe, Erin C. Schafer.—Second edition.
p. ; cm. — (Core clinical concepts in audiology)
Includes bibliographical references and index.
ISBN 978-1-59756-552-3 (alk. paper) — ISBN 1-59756-552-0 (alk. paper)
I. Schafer, Erin C., author. II. Title. III. Series: Core clinical concepts in audiology.
[DNLM: 1. Cochlear Implants—Case Reports. 2. Software—Case Reports. WV 274]
RF305
617.8'82—dc23

2014023290

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# Foreword

*Programming Cochlear Implants* (2nd edition) by Jace Wolfe and Erin Schafer is the latest addition to the Cochlear Implant Component of the Core Clinical Concepts in Audiology series of books. Similar to the first edition, this book provides detailed and comprehensive information about programming cochlear implants that cannot be found in other books. They begin by providing an overview of basic cochlear implant components, including updated information regarding newly introduced electrode arrays and speech processor components. A separate chapter is devoted to each of the three currently available cochlear implant systems (Chapters 4, 5, 6), where they review programming equipment (i.e., programming interface tools), programming software, sound processing features, and more. They provide the reader with step-by-step overviews of various programming appointments, describe uses for various objective measures, and discuss rehabilitative activities that can be used with patients to enhance their performance.

In the final chapters, the authors provide an excellent update regarding hearing assistive technology and cochlear implants (Chapter 9). They describe various types of assistive technologies that can be utilized with the speech processor to enhance performance in difficult listening situations, including a comprehensive description of FM devices, speech processor telecoils, and recently introduced wireless accessories.

Both new and experienced clinicians will treasure this book and will refer to it regularly. We are both pleased and proud that Drs. Wolfe and Shafer took time out of their busy schedules to provide us with this valuable update to their original work.

—Teresa A. Zwolan, PhD
Series Editor
Cochlear Implant Series

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# Preface

Clinicians must effectively program a cochlear implant in order for the recipient to experience an optimal outcome. Effectively programming cochlear implant sound processors is a challenging task for a number of reasons. Recipients present with a variety of needs, significant differences exist in the hardware, signal processing, and programming strategies across the major cochlear implant manufacturers, and cochlear implant technology evolves rapidly. Unfortunately, there is not a single source to which clinicians may turn to obtain comprehensive and timely information pertaining to the programming of contemporary cochlear implants. That changes with this book!

The primary objective of this book, *Programming Cochlear Implants, Second Edition*, is to offer practical guidance for clinicians who program cochlear implants. It represents the only textbook in which a detailed step-by-step process of cochlear implant programming is provided with the goal of preparing the reader to successfully program the cochlear implants of recipients of all ages. This book is a thorough revision of its predecessor and includes additional content pertaining to recent advances, such as electroacoustic stimulation and advances in signal processing.

The basics of cochlear implant programming are introduced, but the authors also expand into advanced programming techniques. Additionally, the authors provide an in-depth manufacturer-specific description of cochlear implant hardware and programming strategies. Specific topics covered include the basics of cochlear implant terminology, hardware, and programming; clinical protocols for cochlear implant management; programming considerations for Advanced Bionics, Cochlear, and MED-EL devices; troubleshooting during the programming process; programming considerations for bilateral cochlear implant recipients; basic use of objective measures as a guide to program cochlear implants; use of wireless hearing assistance technology with cochlear implants; programming and management considerations for recipients using electroacoustic stimulation; management of the difficult-to-program recipient; case studies that illustrate the topics that are addressed throughout the book. This book is intended for practicing clinicians who are providing or who plan to provide services for cochlear implant users and also for graduate-level students.

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# Acknowledgments

The authors would like to express sincere gratitude to the Plural Publishing Inc. with particular thanks to Angie Singh, Terry Zwolan, Casey Stach, and Brad Stach. In addition, we offer sincere thanks to representatives from several hearing-technology manufacturers, including Sarah Downing, Dawn Koch, and Anna St. Romain of Advanced Bionics Corporation; Matthew Lipscombe, Michael Goorevich, Stefan Mauger, Ryan Carpenter, Christy Miller-Gardner, Tina Wall, Leigh Ann Monthey, Esti Nel, Aaron Parkinson, and Pete Arkis of Cochlear Ltd.; and Scott Hansson, Amanda Curtis, Pritesh Pandya, and Darla Franz of MED-EL; Christine Jones, Stefan Launer, and Hans Mulder of Phonak; and Maureen Doty Tomasula of Oticon, for providing helpful comments and suggestions to preliminary drafts of this text and photographs for this book. Finally, and most deservedly, we would like to thank all of the cochlear implant recipients we have served. It is their motivation, courage, and success that served as the impetus behind the development of this book.

I (Jace Wolfe) would also like to thank the numerous mentors who have shaped me professionally including Richard Talbott, Stephen Painton, Michael Dennis, Francis Kuk, and last but certainly not least, Michael Grim, who selflessly and aptly shared countless hours teaching a motivated but often dense student the nuances of Audiology. Additionally, I would like to thank Joanna Smith, Teresa Caraway, and Drs. Stan Baker and Mark Wood for introducing me to the world of cochlear implants and providing an excellent model of patient care for persons with cochlear implants. I would also like to thank Joanna and Dr. Wood for the mentorship they have provided for me. Also, I owe a debt of gratitude to the entire team at Hearts for Hearing for inspiring me on a daily basis to provide unlimited opportunities for persons with hearing loss. You guys are amazing, and I realize every day how blessed I am to work with you all. Furthermore, I would like to thank my dad, mom, and sister for instilling within me passion to live my life serving others. Finally, and most importantly, I owe a heartfelt gratitude to my loving wife Lynnette, and three children, Hayden, Harper, and Henley, for the love they have shown me and the pride they have in the work I do. It is only with their support that I am able to best utilize my resources to serve persons with cochlear implants.

I (Erin Schafer) would like to express love and thanks to my husband, Michael, daughters, Avery and Keira Schafer, and parents, Anne and Randy Smithson for their patience and assistance during the writing of this book. Without my family's support, completion of this book would not have been possible. I would also like to express my appreciation to my wonderful mentors, Linda Thibodeau, Carol and Jeff Cokely, Jackie Clark, Lee Wilson, Ross Roeser, and Emily Tobey, who helped me to become a confident audiologist and associate professor through their excellent examples of leadership and teaching. Finally, I would like to thank all of my wonderful colleagues at the University of North Texas who are supportive and encouraging every day.

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# 1

# Basic Components and Operation of a Cochlear Implant

*Jace Wolfe, Erin C. Schafer, and Sara Neumann*

In most cases of sensorineural hearing loss, the primary site of lesion resides within the cochlea. Cochlear hearing loss results in insufficient transduction of acoustico-mechanical energy into neural impulses at the auditory nerve. The term nerve deafness is frequently used to describe sensorineural hearing loss, but this term is frequently a misnomer. The auditory nerve is oftentimes relatively intact and functional but does not receive adequate stimulation from the cochlea. Consequently, the cochlear implant is surgically inserted into the cochlea to bypass the deficient part of the auditory system and to provide direct stimulation to the auditory nerve. As a result, most people with sensorineural hearing loss respond favorably to a cochlear implant. In fact, multiple channel cochlear implants are the most successful sensory prosthetic device in the history of medicine. Children with severe to profound hearing loss who receive an implant at an early age often develop age-appropriate spoken language (Geers, 2004; Geers, Brenner, & Davidson, 2004; Geers, Moog, Biedenstein, Brenner, & Hayes, 2009). Adults with postlingual deafness who receive cochlear implants frequently achieve excellent open-set sentence recognition (i.e., 100%; Gifford, Shallop, & Peterson, 2008; Helms et al., 2004) and can converse over the telephone (Anderson et al., 2006). Cochlear implants may also provide benefit to people with pathologic conditions of the auditory nerve, such as auditory neuropathy spectrum disorder and fairly significant degeneration of spiral ganglion fibers (Rance & Barker, 2008).

Success with a cochlear implant is strongly influenced by the quality of the user's program created by the audiologist. A cochlear implant program, also known as a MAP, is patient specific. The program determines how the cochlear implant will provide electrical stimulation to the auditory nerve to represent speech and environmental sounds detected by the sound processor microphone. At a fundamental level, cochlear implant programming involves the determination of stimulation levels required to restore audibility for soft sounds and achieve loudness normalization for a large range of inputs. In addition to stimulation

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levels, programming audiologists must also determine numerous other adjustable parameters to allow for optimal patient performance.

According to published research and in the author's clinical experience, recipients using inappropriate cochlear implant programs will experience poor performance and outcomes (Geers et al., 2004; Wolfe & Kasulis, 2008). However, optimization of these inadequate programs often allows recipients to realize remarkably good performance. Optimum initial programming is particularly important for young children who frequently cannot provide verbal feedback about the quality of the signal they receive. Children must have consistent access to speech and environmental sounds during the first few years of life to prevent long-term delays in speech, language, and auditory development.

Although creating a cochlear implant program may seem complicated, and even intimidating, clinicians can rely on basic principles of hearing science, audiology, and aural rehabilitation/habilitation to create suitable programs for recipients. The provision of a quality program is one of the most rewarding facets of audiology, as it serves to restore the recipient's access to social, academic, and professional opportunities. The primary objective of this book is to provide direction to graduate students and professionals on creating the best possible programs/MAPs for cochlear implant recipients of all ages.

The chapters in this book address the basic components of cochlear implants, basic terminology of programming, basic principles of programming, manufacturer-specific programming considerations, practical clinical protocols for cochlear implant programming, troubleshooting of patient problems and complications, and use of assistive listening devices with cochlear implants. Additionally, this book addresses electroacoustic stimulation (EAS), which is sometimes referred to as hybrid cochlear implantation. Furthermore, the final chapter in this book provides several case studies that serve to reinforce the content from the preceding chapters. The information in this book is applicable to almost all cochlear implant systems used in contemporary clinical settings. The manufacturer-specific information provided in Chapters 4, 5, and 6 focuses on the three manufacturers that provide cochlear implant systems in North America: Advanced Bionics Corporation, Cochlear Corporation, and MED-EL Corporation.

## BASIC OPERATION OF COCHLEAR IMPLANTS

Although differences exist in hardware across manufacturers, there are several common components among all cochlear implant devices. As shown in Figure 1–1, these components include an external sound processor with a transmitting cable and electromagnetic radio frequency (RF) transmitting coil, an internal receiving coil, an internal stimulator with an electrode array, and an interface device to connect the recipient sound processor to the clinician's programming computer.

The basic operation of a cochlear implant is also similar regardless of manufacturer. The microphone of the external sound processor captures acoustic signals in the user's environment and transduces the input into an electrical signal. This electrical signal is typically sent to a preamplifier to improve the signal-to-noise ratio during transmission to the processor. The preamplifier occasionally provides a greater boost for high-frequency components of the input signal because high-frequency phonemes, such as /s/, are less intense and more susceptible to masking. Next, the signal is analyzed by a sophisticated digital signal processor (DSP) in the external sound processor to classify the input according to intensity, frequency, and time domains and to convert the signal into an electrical code that will represent these features at the auditory nerve. The coded signal is then converted from a digital signal back into an electrical signal and sent to the RF coil via a transmitting cable. At the RF coil, the electrical signal is converted to an electromagnetic signal (i.e., magnetic lines of flux are created as the electrical signal travels through the transmitting coil) and transmitted via electromagnetic induction to an internal receiving coil (antenna) that is directly wired to the internal stimulator. Magnets are located in the center of both the external RF coil and internal receiving coil, which provides adhesion of the external RF coil to the head and

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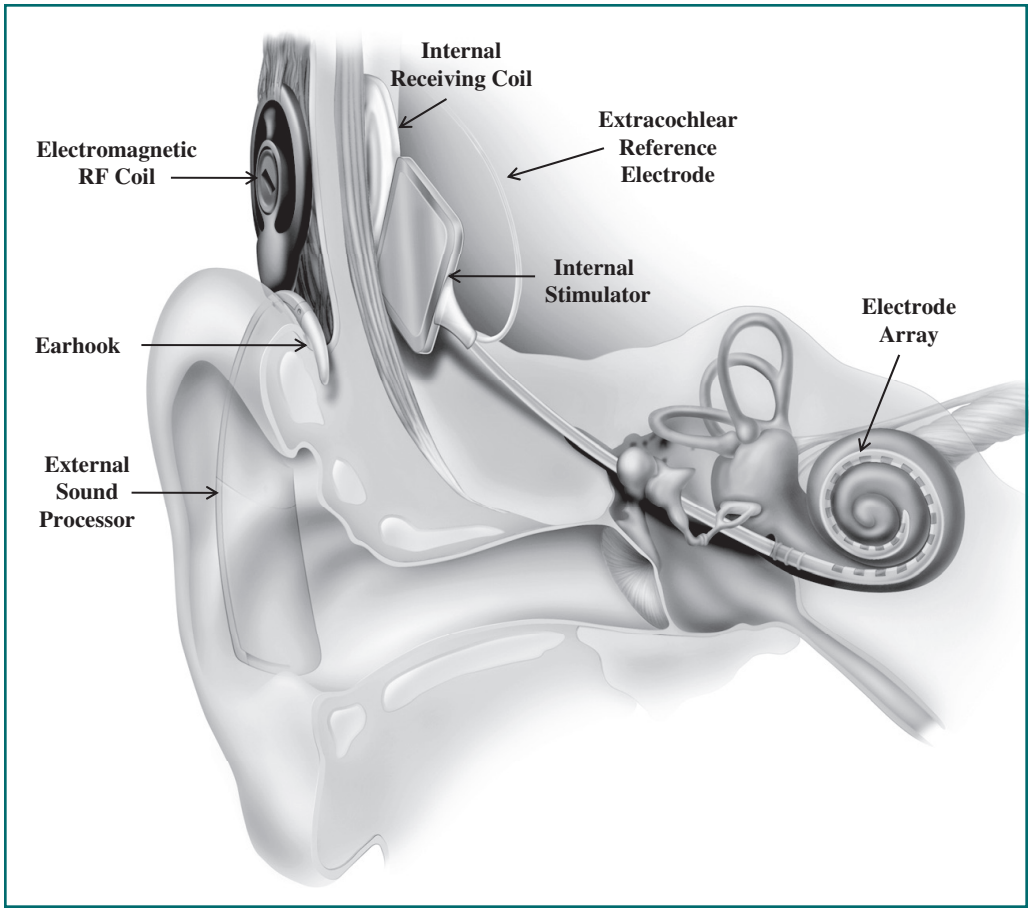



**FIGURE 1–1.** Basic operation of a cochlear implant. *Note.* Provided courtesy of Cochlear™ Ltd., © 2009 Cochlear Ltd.

alignment directly over the internal receiving coil. The RF signal, which is device specific, also serves as the power supply for the internal stimulator. When the magnetic lines of flux (RF) pass over the internal receiving coil, an electrical signal is induced in the internal coil and passed onto the internal stimulator. The internal stimulator, which also contains a DSP, converts the electrical signal into a digital code. The processor determines the stimulation needs for the user, and it converts the digital code to electrical pulses based on the characteristics of the input signal and a set of rules defined by the coding strategy. The electrical pulses are then sent along the electrode lead to the stimulating intracochlear electrode contacts (see Figure 1–1), where the pulses stimulate auditory nerve fibers innervating the cochlea. The electrical stimulation delivered to the intracochlear electrode contacts returns to an extracochlear electrode, which serves as the return or ground electrode (also known as the reference electrode). Extracochlear return electrodes can be located on the internal device (i.e., on the case as shown in Figure 1–2) or at a location remote from the primary electrode lead (see Figure 1–1).

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As previously mentioned, the electric signal from the external transmitting coil provides power to the internal device. The strength of this electric signal is adjusted on an individualized basis to minimize the necessary transmission strength and power consumption (i.e., optimize battery life). At the same time, this signal will need to provide enough power to the internal device to stimulate and effectively represent without interruption high-level inputs, which require stronger RF strength and higher levels of stimulation.

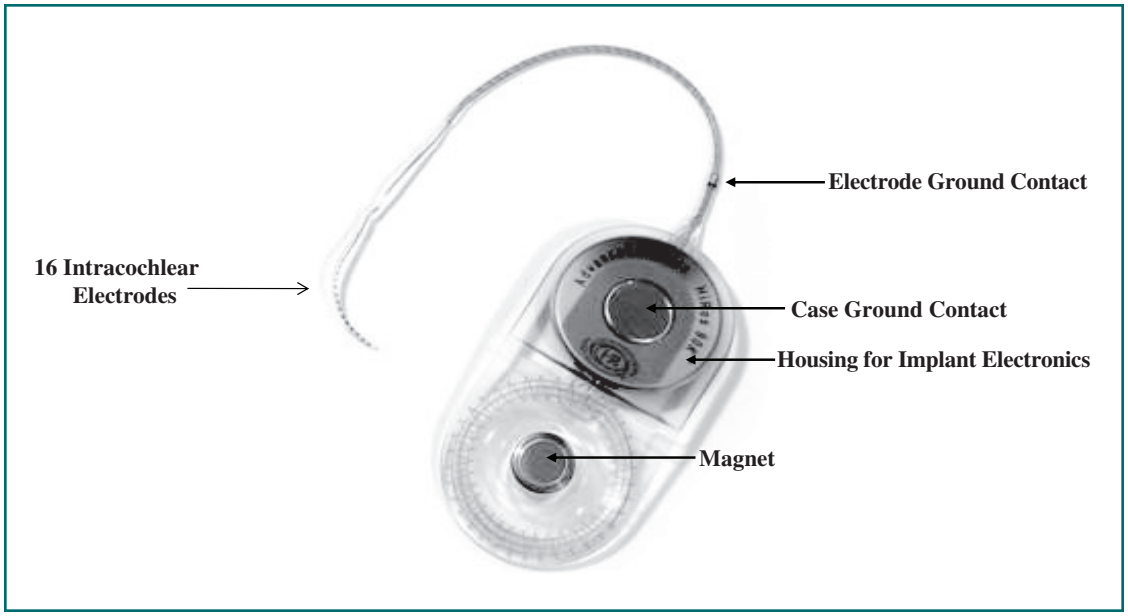



**FIGURE 1–2.** Advanced Bionics HiRes 90K internal stimulator and HiFocus 1j electrode. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

In the most current cochlear implant systems, the external sound processor uses digital bandpass filtering, Fast Fourier transformation, or Hilbert transformation to divide the complex input signal into individual frequency segments referred to as channels (Figure 1–3). The output from these bandpass filters is then sent to a rectifier that captures and produces the spectral envelope (i.e., boundary across frequencies) of the input. Next, the output of the rectifier is used to modulate a train of biphasic electrical pulses (see Figure 1–3), which are delivered to the electrode contact that corresponds to a given channel. Multiple channel cochlear implants take advantage of the natural tonotopic organization of the cochlea by delivering high-frequency signals to electrodes located toward the basal end of the cochlea and low-frequency signals to more apical locations.

The power used to operate the external sound processor is delivered from a battery that is directly coupled to the sound processor. The most recent external sound processors may be operated with manufacturer- and device-specific lithium ion rechargeable batteries. This battery has several advantages over other types of rechargeable batteries including a long shelf and service life, a flat voltage discharge curve, a relatively robust voltage capacity, and no memory effect. As a result, the battery does not have to be completely discharged prior to recharging it. Many sound processors also may be operated by zinc-air (similar to those used with hearing aids) or conventional alkaline batteries.

## BASIC COMPONENTS OF CURRENT COCHLEAR IMPLANT SYSTEMS

There are significant differences in the appearance and features of most current commercial sound processors for each manufacturer; however, many of the components are similar across the three manufacturers. For instance, each sound processor contains an earhook to facilitate retention of the sound processor on the external ear, a microphone to capture acoustic signals in the user's environment, a DSP to analyze input signals and determine how these signals should be relayed to the user, a single power source for the sound processor and the internal device, and a transmitting cable and RF coil to transmit the signal to the internal device. The following sections

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provide details specific to the hardware of the most recently introduced cochlear implant system of each manufacturer (in alphabetical order). In addition, general information on previous generations of cochlear implant systems for each manufacturer is provided in Tables 1–1, 1–2, and 1–3.

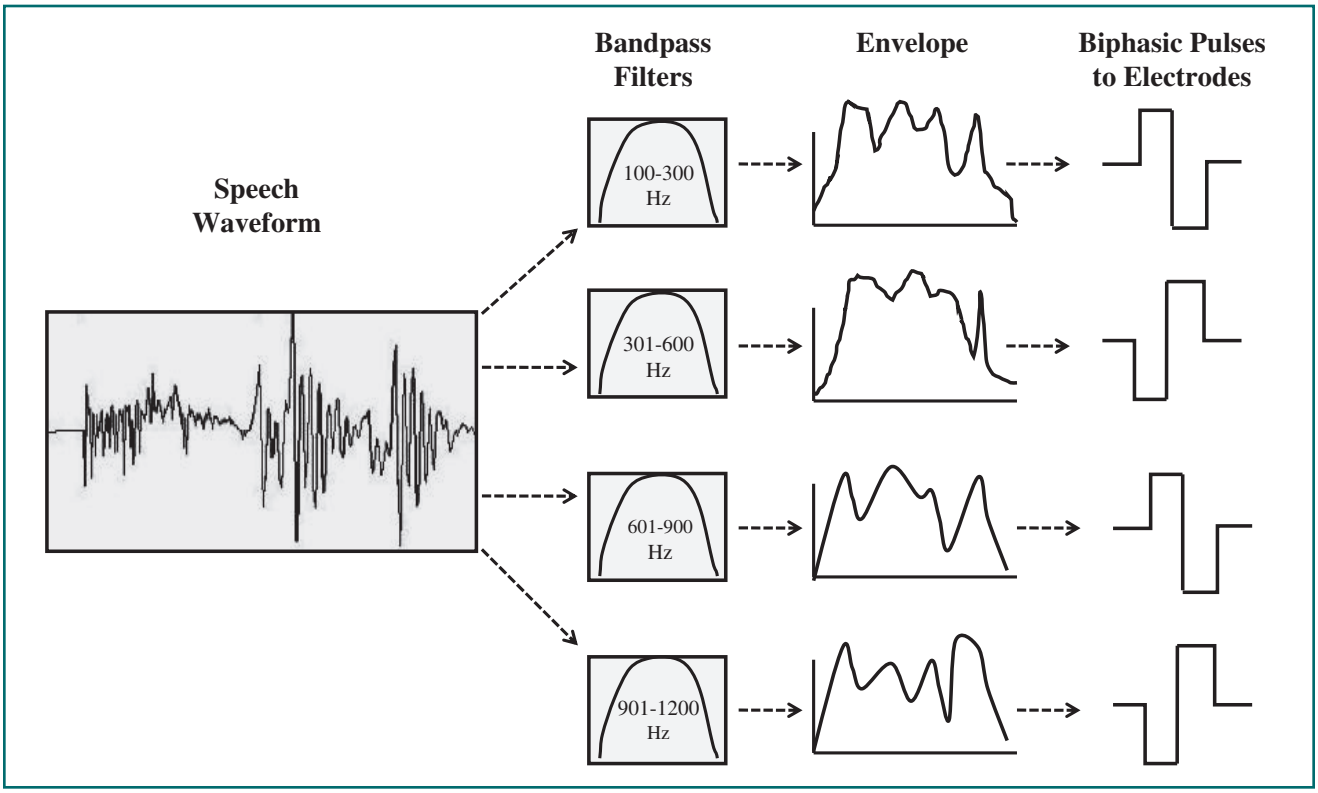



**FIGURE 1–3.** Diagram of how a complex signal is sent through a bandpass filter and a rectifier to determine the temporal envelope of each channel before biphasic pulses are sent to the electrode contacts.

**Table 1–1.** Previous Internal Devices from Advanced Bionics Corporation

| *Device* | *Release Date* | *Description* | *Advantages* |
|---|---|---|---|
| Clarion 1.0 & 1.2 | • Clarion 1.0: 1991 for adults<br>• Clarion 1.2: 1995, smaller version approved for pediatric trials | • Ceramic case & DSP<br>• 16 electrodes stimulate ≤8 electrodes at once<br>• Max stimulation rate: 1.0 is 833 pps per channel; 1.2 is 1,666 pps per channel<br>• 3 arrays: Radial Bipolar Standard, Enhanced Bipolar, HiFocus I | • 3 coding strategies: CIS, MPS, and SAS<br>• Case biocompatible<br>• HiFocus array: good performance with low current levels<br>• Telemetry system: measures electrode impedance |
| CII | 2002 | • Case similar to 1.2, electronics similar to HiRes 90K<br>• Max stim rate 5000 pps per channel with PSS | • Electronics more sophisticated than 1.2<br>• HiRes and HiRes 120 Signal Coding |

*Note*. *CIS* = continuous interleaved sampling; *DSP* = digital signal processor; *ECAP* = electrically evoked compound action potential; *MPS* = multiple pulsatile sampler; *pps* = pulses per second; *PSS* = partially simultaneous stimulation; *SAS* = simultaneous analog strategy; *stim* = stimulation.

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**Table 1–2.** Previous Internal Devices from Cochlear Corporation

| *Device* | *Release Date* | *Description* | *Advantages* |
|---|---|---|---|
| Nucleus 24 | 1998 | • Titanium case durable<br>• DSP; max stim rate: 2,400 pps per channel; 14,400 pps across channels<br>• 22 intracochlear electrode contacts<br>• 3 arrays: Nucleus 24K Straight, Nucleus 24 Contour Advance (CA), Nucleus 24 Double Array<br>• Telemetry: impedance & ECAP | • Durable & reliable<br>• Supported good open-set SR<br>• Removable magnet for MRI<br>• Case & ball extracochlear ground electrodes; multiple electrode coupling modes<br>• ACE, CIS, & SPEAK coding |
| CI512 | 2009<br>Removed from market in 2011 | • DSP; Maximum stimulation rate: 32,000 pps<br>• Thickness: 3.9 mm; Perimodiolar array<br>• 22 platinum intracochlear electrodes, non-linear spacing; 2 extracochlear reference electrodes (1 on separate lead) | • Telemetry system<br>• Coding strategies: ACE, ACE(RE), CIS, CIS(RE) & SPEAK<br>• Removable magnet for MRI up to 1.5 tesla |

*Note*. *ACE* = advanced combination encoder; *CIS* = continuous interleaved sampling; *ECAP* = electrically-evoked compound action potential; *PPS* = pulses per second; *SPEAK* = spectral peak; *SR* = speech recognition; *stim* = stimulation.

**Table 1–3.** Previous Internal Devices from MED-EL Corporation

| *Device* | *Release Date* | *Description* | *Advantages* |
|---|---|---|---|
| COMBI 40+ | Combi 40+: 2001 for adults | • Ceramic case & DSP<br>• Max stim rate: 1,515 per channel; 18,180 pps across all channels<br>• 12 electrodes over 30 mm, provides LF<br>• Reference electrode: case -CIS+ coding | • Good open-set speech recognition for most<br>• Monopolar stimulation/ electrode coupling |
| PULSAR CI$^{100}$ | 2005 | • Max stim rate: 4,225 pps per channel; 50,704 pps across channels<br>• Most current signal coding -24 electrode contacts in pairs; 12 electrode sites<br>• Individual output circuits for each site | • High rates of stim<br>• Telemetry: electrode impedance & ECAP<br>• Case extracochlear electrode; monopolar stim<br>• MRI safe at up to 1.5 tesla |
| SONATA TI$^{100}$ | 2007 | • Max stim rate: 50,704 pulses/second<br>• Thickness: 5.9 mm<br>• Long array: 31.5mm<br>• 24 intracochlear concave electrodes in pairs for 12 intracochlear stimulation sites; primary and EAP reference electrodes | • Most current coding strategies<br>• Telemetry system<br>• Coding strategies: FSP, HD-CIS, & CIS+<br>• MRI safe up to 1.5 tesla |

*Note*. *DSP* = digital signal processing; *CIS* = continuous interleaved sampling; *ECAP* = electrically-evoked compound action potential; *LF* = low frequencies; *pps* = pulses per second; *stim* = stimulation; *FSP* = fine structure processing; *HD* = high definition.

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## ADVANCED BIONICS CORPORATION

The sound processors and internal devices from Advanced Bionics are discussed in the following sections.

### Naída CI Q70 Sound Processor

At the time of this writing, the newest Advanced Bionics sound processor is the Naída CI Q70 processor. In 2009, Sonova Holding AG acquired Advanced Bionics. Sonova is also a manufacturer of hearing instruments with brands including Phonak and Unitron. Sonova's acquisition of Advanced Bionics expanded its portfolio to offer comprehensive solutions to treat for all major forms of hearing loss. Moreover, Sonova's collaboration with Advanced Bionics allowed many of the advanced signal processing strategies of its hearing aids to be incorporated into a cochlear implant sound processor. The Naída CI Q70 sound processor, which was commercially approved by the United States Food and Drug Administration (US FDA) in September 2013, represents the first example in which Phonak hearing aid technology has been integrated into an Advanced Bionics sound processor. As a result, the Naída CI Q70 possesses a number of features that are unique to the cochlear implant industry and that provide a significant improvement in signal processing, aesthetics, usability, and access to hearing assistance technologies relative to the Advanced Bionics Harmony sound processor. The Naída CI Q70 must be programmed in the Advanced Bionics SoundWave 2.2.25 (or later version) software platform, which will be discussed in detail in Chapter 4.

The Naída CI Q70 processor is shown in Figure 1–4 and is significantly smaller (40%) than its ear-level predecessor, the Harmony (Figure 1–5). In fact, the width of a Naída CI Q70 sound processor is similar to a Phonak Naída power behind-the-ear (BTE) hearing aid, which is commonly used by persons with severe to profound hearing loss prior to cochlear implantation. When worn in the standard wear configuration, the Naída CI Q70 possesses four components: (1) the sound



**FIGURE 1–4.** Advanced Bionics Naída CI Q70 sound processor. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com




**FIGURE 1–5.** Size comparisons of Advanced Bionics previous sound processor (SP), Naída CI, and a typically sized hearing aid. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

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processor, (2) the power source (battery), (3) the T-Mic™ 2, and (4) the headpiece, which comprises a cable and transmitting coil.

The Advanced Bionics Naída CI Q70 may be used with two different remote accessories: (1) the AB myPilot remote control pictured in Figure 1–6, and (2) the Phonak ComPilot shown in Figure 1–7, which is a wireless streaming device that allows connectivity to consumer electronics including Bluetooth-enabled mobile telephones, laptop computers, remote microphones, televisions, and so forth. Information regarding these remote controls and their functions are provided in Table 1–4. Additionally, with the release of the




**FIGURE 1–6.** Advanced Bionics myPilot remote control. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com




**FIGURE 1–7.** Phonak ComPilot wireless streaming accessory and remote control. *Note.* Image provided courtesy of Phonak. http://www.phonak.com

**Table 1–4.** Overview of Remote Controls and Basic Functions

| *Manufacturer* | *Remote Model* | *Pairing & Basic Functions* |
|---|---|---|
| Advanced Bionics | myPilot | • **To pair:** must be done within 5 minutes of Naída start-up, select *Pairing* from the remote main menu, select scan for devices<br>• Color LCD to monitor settings; change programs<br>• Volume and sensitivity adjustments, separately if bilateral<br>• Monitor the status of the processor battery<br>• ZoomControl direct access; choose direction of focus, bilateral only<br>• Monitor status of remote battery<br>• Store and Recall settings<br>• Programmable/manual automatic keypad lock |

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**Table 1–4.** *continued*

| *Manufacturer* | *Remote Model* | *Pairing & Basic Functions* |
|---|---|---|
| Phonak | ComPilot | • **To pair:** turn on ComPilot with switch on side, put neckloop around neck, hold the Connect and Main buttons until Audio indicator turns violet<br>• Adjust volume with +/– buttons<br>• Listen to music and TV with cable or with Bluetooth functionality<br>• Connect via Bluetooth phone or Bluetooth-compatible device<br>• Access FM wirelessly by plugging in FM receiver to Europort |
| Cochlear | Nucleus 6 Remote Assistant (CR230) | • **To pair:** hold coil on the coil guides on back of the remote assistant<br>• LCD screen to monitor settings; change programs<br>• Volume and sensitivity adjustments, separately if bilateral<br>• Monitor the status of the processor, battery coil, and cable<br>• Activate and adjust the telecoil settings for phone use<br>• Bass and treble adjustments<br>• Audio accessory status and adjust accessory mixing ratio<br>• Lock/unlock processor buttons; control processor beeps & lights |
| | Nucleus 6 Remote Control (CR210) | • **To pair:** hold coil on the back of the remote control<br>• LCD screen to monitor settings<br>• Volume and sensitivity adjustments<br>• Monitor status of remote battery<br>• Activate telecoil; turn on/off autocoil<br>• Turn audio accessories on/off<br>• Program up/down button |
| | Nucleus 5 Remote Assistant (CR110) | • **To pair:** hold coil on the coil guides on back of the remote assistant<br>• Display screen to monitor settings<br>• Volume and sensitivity adjustments, separately if bilateral<br>• Activate telecoil; turn on/off autocoil<br>• Monitor status of remote assistant and processor<br>• Turn audio accessories on/off<br>• Monitor accessory status and adjust microphone mixing ratio<br>• Lock/unlock processor buttons; control processor beeps & lights |
| MED-EL | FineTuner | • **To synchronize/configure:** place the coil of the processor over the MT key on the FineTuner<br>• 15 keys/buttons<br>• LED indicator lights for alarms & functions<br>• Volume and sensitivity adjustments, separately if bilateral<br>• Default key to use configured settings by audiologist<br>• Access up to four programs<br>• Signal source selection: microphone, telecoil, or combined input<br>• Lock/unlock remote keys |

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Naída CI Q70 sound processor, Advanced Bionics also introduced a new programming interface called the CPI-3 (Figure 1–8). The aforementioned hardware are discussed in detail in the following pages.

### *Sound Processor Microphones*

The Naída CI Q70 sound processor possesses several unique features. In the standard wearing configuration, there are four microphones: the T-Mic™ 2, two omnidirectional microphones that may be used to provide adaptive directional beamforming, and an integrated microphone within the Universal Headpiece (UHP). Thus, the recipient has access to a variety of microphone configurations to meet diverse listening needs in the real world. The T-Mic 2 (Figure 1–9) is an updated version of the T-Mic, but the T-Mic 2 provides better aesthetics and comfort than its predecessor. Additionally, the T-Mic 2 comes in three sizes to accommodate different ear sizes and shapes. The T-Mic 2 also includes a design-integrated microphone cover to protect against dirt or debris. This cover may be replaced by the user or clinician, if necessary. Furthermore, the T-Mic 2 may be removed and replaced by a conventional behind-the-ear processor earhook that only serves the purpose of maintaining retention of the processor on the recipient's ear. The conventional earhook is available in a small and standard size (Figure 1–10). The T-Mic 2 and conventional earhooks are both tamper-proof and must be removed by using a small tool to remove a retention pin.

The original objectives of the T-Mic, first introduced in 2003 for use with the CII BTE sound processor, were to facilitate telephone use by placing the microphone at the opening of the external auditory meatus (ear canal) and to provide for a more natural location of sound direction compared to the typical microphone placement at the top of the auricle with a standard behind-the-ear sound processor. Research has also shown that the T-Mic provides an improvement in speech recognition in noise when compared to use of an omnidirectional microphone located at the top of the auricle (Gifford & Revit, 2010). Anecdotally, recipients also report that the T-Mic simplifies the use of circumaural (over the ear) earphones.



**FIGURE 1–8.** Advanced Bionics CPI-3 Programming Interface. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

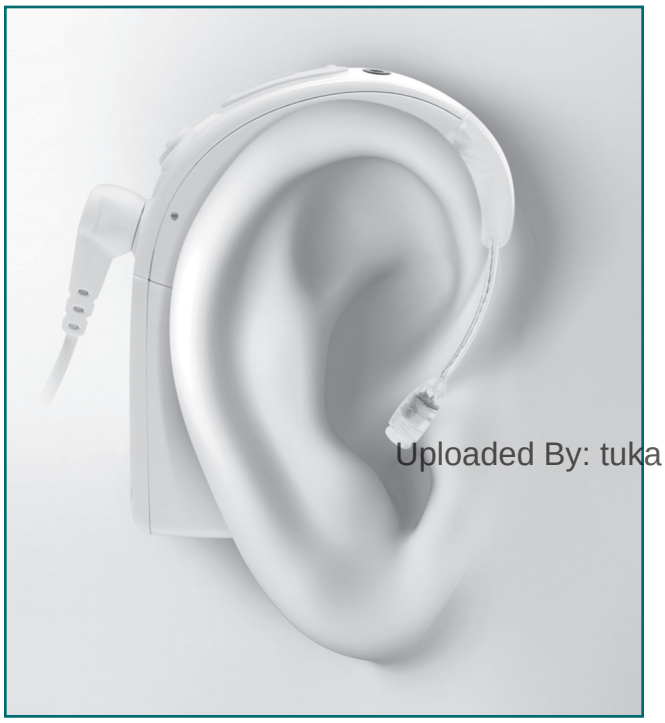


**FIGURE 1–9.** Advanced Bionics T-Mic™ 2. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

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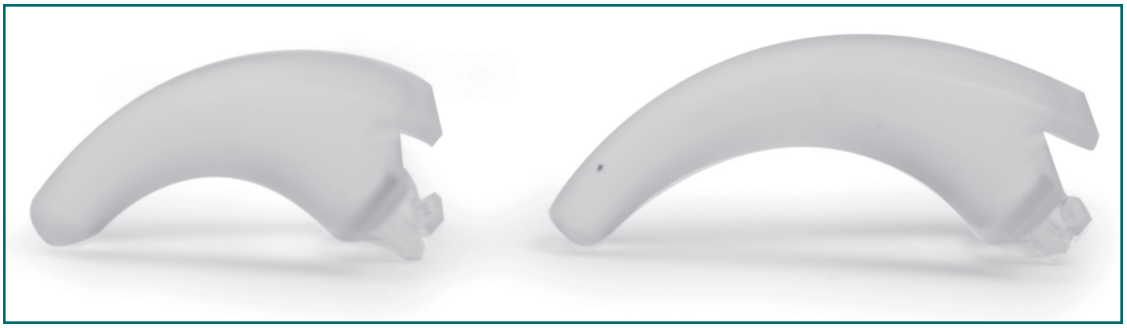


**FIGURE 1–10.** Advanced Bionics small and standard sized behind-the-ear processor earhooks. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

### *Digital Signal Processing (DSP)*

The Naída CI Q70 possesses a sophisticated DSP, which has a 16-bit analog-to-digital converter allowing for a front-end DSP dynamic range of 96 dB. The sampling rate of the DSP is 17.4 KHz, which allows for accurate representation of the input spectrum beyond 8000 Hz. The input dynamic range (IDR) of the processor (which is determined by a host of factors including but not limited to the DSP dynamic range, the noise floor of the microphone, and the maximum limit of the microphone) ranges from 20 to 80 dB, with the maximum range of input levels captured without overload or compression limiting spanning from approximately 25- to 105-dB SPL. The DSP chip of the Naída CI Q70 features the same technologies available in the Phonak Quest hearing aid chip. This chip enables access to many of the advanced signal processing features available in the Phonak Quest hearing aid line, including adaptive directionality (UltraZoom) and adaptive signal processing that attempts to automatically adjust processor settings to meet recipient needs across a wide variety of acoustic settings. This technology, known as Binaural VoiceStream Technology™, allows for binaural processing and audio streaming for bilateral Advanced Bionics users (with CII/HiRes 90K or later cochlear implants) to benefit from binaural directionality (ZoomControl), binaural telephone use (DuoPhone), and bilateral processor adjustments (QuickSync).

Furthermore, the Naída CI Q70 offers a wide variety of signal coding strategies ranging from Continuous Interleaved Sampling (CIS) to the recently introduced (2013) HiRes Optima strategy. Detailed information pertaining to the management, programming, and signal processing capabilities of the Naída CI Q70 may be found in Chapter 4.

### *Processor Buttons*

The Naída CI Q70 sound processor possesses a dedicated push-button volume control (see Figure 1–4) that is designed to promote enhanced tactile sensation and simple adjustment. Pressing the top half of the volume control button increases stimulation, whereas pressing the bottom half of the volume control results in a decrease in stimulation. The range of the volume control may be changed or disabled in the programming software by the programming clinician. The effect of programming adjustments on the operation of the volume control is discussed in detail in Chapter 4.

The Naída CI Q70 also has a program button, which may be used to access up to five programs per ear (i.e., separate programs for typical listening situations such as noisy environments, music, telephone, hearing assistance technology). It should be noted that the Naída CI Q70 may be safely programmed for both the right and left ears for bilateral cochlear implant recipients. As a result, up to five programs for each ear may be loaded onto one sound processor for a total of ten programs within the processor. Bi-telemetry communication between the Naída processor and the cochlear implant internal device allows for automatic determination of which ear the processor is

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stimulating. The Naída CI Q70 is the industry's only sound processor that possesses this feature. Additionally, the program button can be disabled by the clinician within SoundWave.

The program button may also be used to place the processor in standby mode, a low power state in which no stimulation is provided from the processor. This may be desirable if the recipient wants to refrain from receiving an audio signal from the device but does not want to remove the processor from the ear or the power source from the processor. Examples include a recipient who is concentrating on a task and does not want to be distracted by outside noises or a recipient who is subjected to unwanted, annoying noise. To actively enable standby mode, the user must press on the program button for 4 seconds and then release. Additionally, the processor passively enters standby mode if the processor does not achieve lock with the internal device for more than 5 minutes. The recipient must simply press the program button to return to the settings used prior to the initiation of standby mode. Standby mode is a programmable feature within SoundWave 2.2.

### Processor Water Resistance and Lights

The Naída CI Q70 sound processor possesses water-resistant covers over both microphones in order to provide resistance to water damage. Additionally, the Naída sound processor's buttons and points of connection to the headpiece cable and battery are hermetically sealed to enhance water resistance, and as a result, the Naída CI Q70 possesses an IP (ingress protection) rating of 57, indicating that it may be left in a meter tall bucket of water for at least 30 minutes and still function to the manufacturer's specification. The Ingress Protection (IP) rating system classifies devices in relation to their degree of resistance to the intrusion of foreign objects. The IP rating is a two-digit number, with the first number referring to the resistance to the intrusion from solid objects and the second number referring to the intrusion of liquid objects. Tables 1–5 and 1–6 provide a complete description of the IP rating scale.

It should be noted that the IP57 rating of the Naída CI Q70 sound processor is only valid when the T-Mic 2 is removed and substituted with the conventional earhook (i.e., the T-Mic 2 is not

**Table 1–5.** Ingress Protection (IP) Rating System for Solid Objects (first number in rating)

| Rating | Object Size | Description |
|---|---|---|
| 0 | Not Applicable | No protection against ingress of solid objects |
| 1 | >50 mm | Protection against ingress of any large surface of the body (e.g., back of the hand) but not against deliberate contact with a body part |
| 2 | >12.5 mm | Protection against ingress of smaller body parts (e.g., finger) or similar objects |
| 3 | >2.5 mm | Protection against ingress of small tools (e.g., screwdriver) or thick wires |
| 4 | >1 mm | Protection against ingress of thin wires, screws, etc. |
| 5 | Dust Protected | Ingress of dust is not prevented in entirety, but dust ingress will not affect operation of the device |
| 6 | Dust Tight | Complete resistance to the ingress of dust; complete protection against ingress of solid objects |

IEC 60529—Degrees of Protection Provided by Enclosures (IP Code). Located at http://www.maximintegrated.com/app-notes/index.mvp/id/4126

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**Table 1–6.** Ingress Protection (IP) Rating System for Liquid Objects (second number in rating)

| *Rating* | *Liquid/Application* | *Details* |
|---|---|---|
| 0 | Not Applicable | Not protected against liquid ingress |
| 1 | Dripping water (vertically falling drops) shall have no effect | 10-minute exposure duration; Liquid equivalent to 1 mm of rainfall/ minute |
| 2 | Vertically dripping water causes no harmful effect when device is oriented at an angle up to 15° from its typical use position. | 10-minute exposure duration; Liquid equivalent to 3 mm of rainfall/ minute |
| 3 | Protection against vertical spraying water falling from an angle of 60° above the device | 5-minute exposure duration; Water volume: 0.7 liters/min; Pressure: 80–100 kPa |
| 4 | Water splashing against the device from any angle does not harm the device | 5-minute exposure duration; Water volume: 10 liters/min; Pressure: 80–100 kPa |
| 5 | Water projected from a nozzle (6.5 mm) will not harm the device | 3-minute exposure duration; Water volume: 12.5 liters/min; Pressure: 30 kPa at 3 m distance |
| 6 | Water projected in powerful jets (12.5 mm nozzle) will not harm the device | 3-minute exposure duration; Water volume: 100 liters/min; Pressure: 100 kPa at 3 m distance |
| 7 | Ingress of water is not harmful when device is completely immersed in water | 30-minute exposure duration; Bottom of device immersed in ≥1 m of water with the top of the device immersed in ≥15 mm of water |
| 8 | Continuous immersion in water at a depth of more than 1 meter in under typical use conditions without harmful effect | Continuous immersion; Depth specified by manufacturer; Waterproof for typical use |

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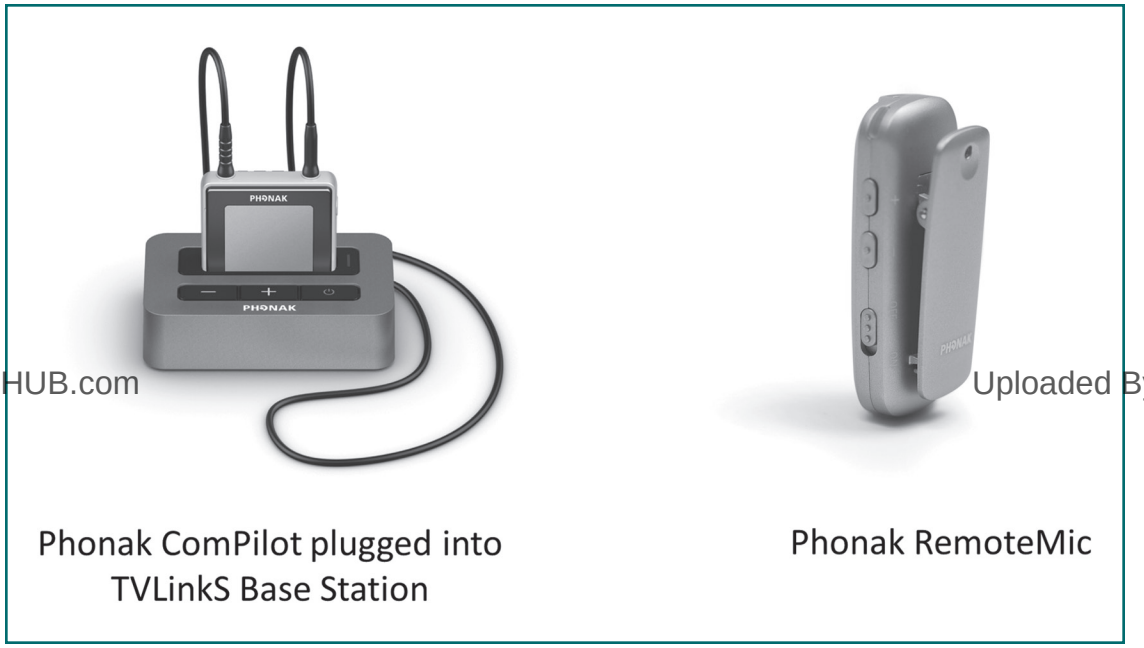
water resistant). It should also be noted that the on-board microphones of the Naída CI Q70 are protected by microphone covers that the recipient or clinician should periodically replace (typically every 3–6 months). The Naída CI Q70 sound processor also features a light emitting diode (LED) that is designed to provide information pertaining to a variety of different functions and operations of the processor. The LED can signify the remaining life of a battery, the program the recipient is using, the status of the communication between the external and internal devices (i.e., lock), the processor is receiving an audio input of moderate level or greater. The LED illuminates in three different colors including green, red, and orange. The exact function of the LED is summarized in Chapter 4.

### *Connecting Processor to External Sources*

The Naída CI Q70 sound processor is equipped with a number of options that allow the recipient to connect to external electronic devices. First, it possesses a programmable internal telecoil that

the recipient may access via a separate telecoil program. The management and programming of the telecoil are discussed in Chapter 4. The Naída CI Q70 also allows the user to wirelessly couple to the Phonak ComPilot (see Figure 1–7), a neck-worn wireless interface that allows the user to wirelessly connect to Bluetooth-enabled mobile and landline telephones, computers, and personal audio devices (e.g., Apple iPod). The ComPilot may also be used to receive signals from the Phonak RemoteMic and TVLink S accessories (Figure 1–11), which both transmit the audio signal to the ComPilot via Bluetooth transmission. The TVLink S also serves as the charger for the ComPilot. At the time of this writing, there has been no research conducted to formally investigate the benefits of the ComPilot, the RemoteMic, and the TVLink S for cochlear implant users. However, it is well established that hearing assistance technologies typically provide considerable improvement in speech understanding and recipient satisfaction, so it is likely that these accessories will be well received by Advanced Bionics recipients. Finally, the ComPilot may also be used as a remote control that allows the recipient to adjust volume and programs. Furthermore, the recipient may also choose to use the AB myPilot, a dedicated remote control that allows the user to make a number of adjustments to his or her Naída CI Q70 sound processor. Additional information pertaining to the AB myPilot, ComPilot, and associated accessories is provided in Chapters 4 and 7.

Although it is not technically classified as an accessory, a Naída CI Listening Check was also developed for use with the Naída CI Q70 processor. Briefly, the Naída CI Listening Check device (Figure 1–12) allows a clinician, caregiver, or significant other with normal hearing to listen to the input to the microphones or accessory input of the Naída processor using commercial earphones with a mini-audio plug. The AB myPilot remote control may also be used to allow the tester to evaluate the input to each microphone separately as well as to listen to the internal alarms available to alert the recipient of important information (e.g., low battery, program change). The Naída CI Listening Check is a portable troubleshooting/assessment tool that is powered by the PowerCel




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**FIGURE 1–11.** Phonak ComPilot Plugged into TVLink S Base Station (*left*) and Phonak RemoteMic (*right*). *Note.* Image provided courtesy of Phonak. http://www.phonak.com

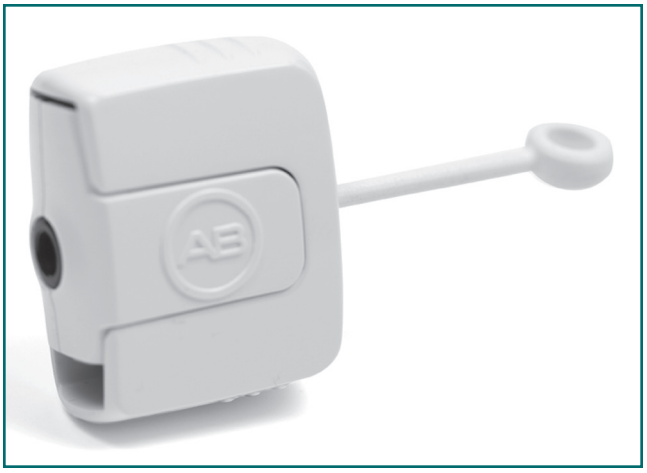


**FIGURE 1–12.** Advanced Bionics Naída CI Q70 Listening Check device. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com



**FIGURE 1–13.** Advanced Bionics headpiece tool used to remove cover. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

of the Naída processor. Programming clinicians should have this tool and use it routinely with every recipient seen in the clinic in order to evaluate the function of the microphones, accessory inputs, FM/Roger input, and so forth. Also, parents of young children should receive this device and be provided with counsel as to its appropriate use and the importance of daily listening checks.

### *Advanced Bionics Universal Headpiece (UHP)*

The UHP (round headpiece shown in Figure 1–4) was introduced with the Neptune sound processor and is compatible with all Advanced Bionics sound processors. The headpiece comprises the transmitting coil and magnet, and an integrated electret condenser omnidirectional microphone, which may be activated when an ear-level microphone is not being used (e.g., for infants with small ears that do not accommodate a BTE processor or for adults who use the Neptune for sports). However, the authors strongly recommend ear-level microphones when possible in order to take advantage of interaural timing cues that are more likely to be available if binaural hearing technology is worn on or near the ears rather than on the body.

A special removal tool (Figure 1–13) is provided to separate the headpiece cover from the headpiece. Up to four magnets may be placed in a small "well" that resides in the transmitting coil underneath the headpiece cover. If less than four magnets are required to facilitate satisfactory headpiece retention, then a small foam "spacer" may be placed in the well along with the magnets to prevent the latter from moving in the well and producing a rattling sound. The clinician must determine the appropriate number and configuration (i.e., magnets placed in the upper well, the shallow underneath well, or both) for each individual in order to achieve consistent headpiece retention without unnecessary pressure and/or irritation of the skin underneath the headpiece site. The UHP also has a bottom cover patch that also may be removed to add one UHP magnet to the external well for additional magnet configurations. The reader is referred to Chapter 3 for guidance on determining appropriate transmitting coil/headpiece strength.

The UHP may be used with a variety of cable lengths depending upon the processor type and wearing configuration/preference. The headpiece cables are available in four different color options (beige, dark brown, black, and white). Finally, the Naída headpiece cable rotates 360 degrees in the

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headpiece connector to allow for a more comfortable fit and to facilitate better retention to a wide variety of head sizes and shapes.

### Battery Module

As shown in Figure 1–14, the The Naída CI Q70 battery options include a lithium-ion rechargeable battery (A), a zinc-air battery cartridge that uses two 675P (power) zinc-air batteries (B), and an AAA PowerPak off-ear battery option (C). The rechargeable PowerCel battery is available in three different sizes (110, 170, and 230). The number (e.g., 110, 170, 230) refers to the capacity of each battery in milliamperes per hour (i.e., 230 refers to a capacity of 230 milliamperes per hour). Each of the different rechargeable batteries provide the same voltage to power the processor, with the only functional difference between the three being the fact that the smaller batteries have a shorter use time between charges. In addition, the 170 battery allows for use of the integrated Phonak Roger 17 receiver. Advanced Bionics reports that for the typical user, the 110 PowerCel will last 12 hours after a full charge (up to 17 hours for users with lower power needs), while the 230 typically lasts 25 hours (up to 36 hours).

Additionally, the 675P zinc-air cartridge typically allows 31 hours of battery life (up to 56 hours). The zinc-air cartridge contains two vent holes on each side of the case to allow air to power the batteries. The presence of these vent holes makes the zinc-air battery module less water-resistant than the rechargeable PowerCel batteries. The zinc-air cartridge is available in a tamperproof option to prevent a young recipient from removing and ingesting the disposable batteries. Finally, the Naída CI Q 70 may also be powered by a module referred to as the AAA PowerPak, which uses three AAA rechargeable or disposable alkaline batteries. The AAA PowerPak is designed to be clipped onto the user's clothes or worn in a pocket. It is coupled to the Naída CI sound processor by a special cable and "CI Power Adapter." Advanced Bionics reports that the AAA PowerPak typically lasts 129 hours when powered by three alkaline AAA batteries.

The PowerCel batteries are charged in a special charger referred to as the PowerCel Charger (Figure 1–15). The PowerCel charger's USB power cord may be plugged into any USB port (i.e., USB port on a computer) and may also be connected to a power supply adapter provided by the manufacturer. Power supply adapters include plugs for outlets of all countries and a special car charger adapter. The PowerCel charger possesses a LED that glows green when the PowerCel charger is receiving sufficient power to charge the batteries.

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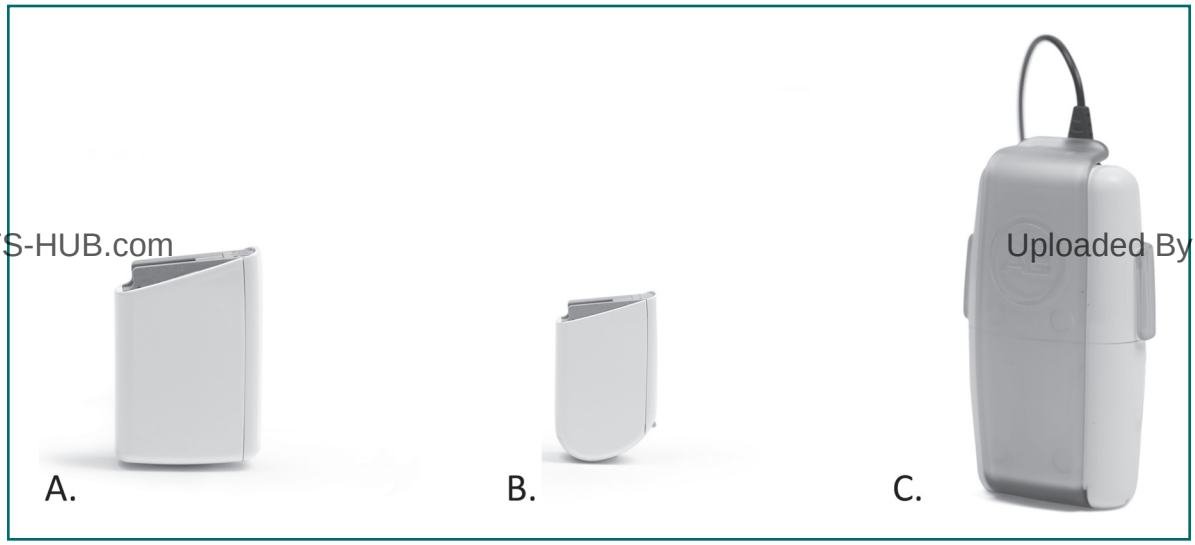



**FIGURE 1–14.** Advanced Bionics Naída CI Q70 AAA PowerPak off-ear battery option. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

The PowerCel charger possesses four charging stations so that four PowerCel batteries may be charged simultaneously. Each charging station has an LED to indicate the charging status of the battery. A blue light denotes that the PowerCel is charging, whereas a green light indicates that the PowerCel is fully charged. A red light (or no light) indicates that the battery or the charging station is faulty. A fully depleted PowerCel battery requires 2 to 3 hours of charging to acquire a full charge. Finally, no damage will occur to the charger or to the battery cartridge if a zinc-air battery cartridge is inadvertently placed on the PowerCel charger.

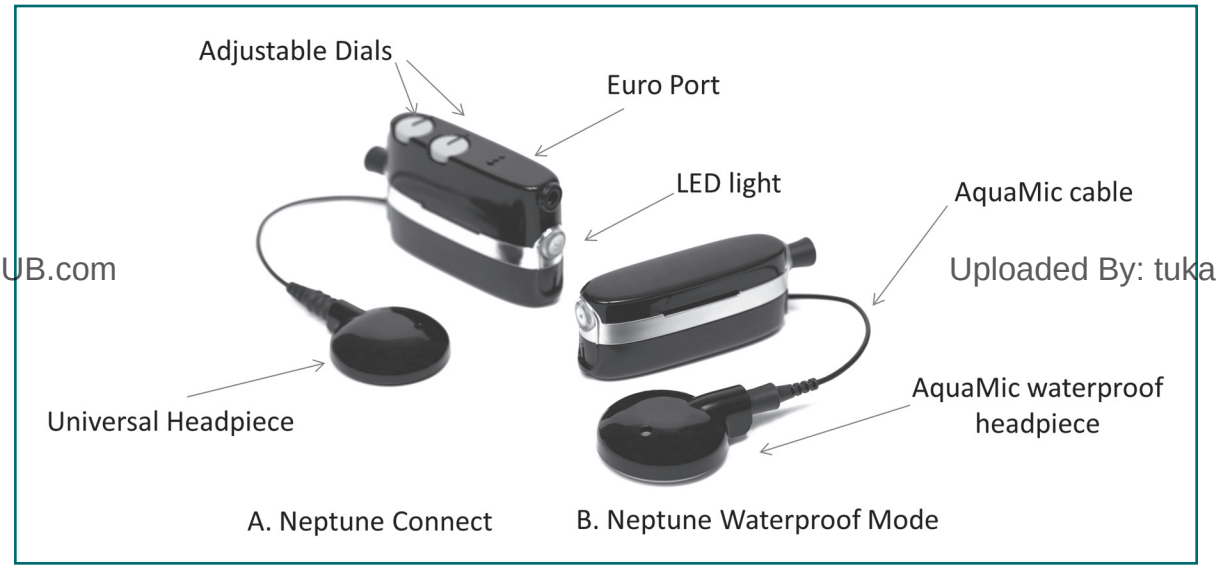



**FIGURE 1–15.** Advanced Bionics PowerCel Batteries and Charger. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

## Neptune Sound Processor

In January of 2012, Advanced Bionics received U.S. FDA approval for the Neptune sound processor (Figure 1–16). The Neptune sound processor must be programmed in the Advanced Bionics SoundWave 2.1 (or later version) software platform, which is discussed in detail in Chapter 4. The Neptune processor was revolutionary in several ways. First, for almost 15 years, the cochlear implant industry had evolved from the use of body-worn processors toward the use of progressively smaller and more sophisticated ear-level processors. The Neptune sound processor was counter to this trend because it was designed to be worn off the ear as a body-worn or head-worn (clipped to a swim cap or to the hair of women) processor.

Another unique design feature of the Neptune sound processor is the fact that it is the industry's




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**FIGURE 1–16.** Advanced Bionics Neptune sound processor configurations. *Note.* Images provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

first fully waterproof, swimmable sound processor. With an IP (ingress protection) rating of 68, the Neptune can resist moisture damage even when the recipient wears it while swimming vigorously or diving into pool, lake, or ocean water. This feature is important to many recipients and their families, because it allows for inclusion in water recreation activities and in situations in which the user perspires heavily. Historically, many implant recipients have reported that they have had to remove their processor to prevent damage in these types of activities, and consequently, they have been unable to enjoy listening during important social and/or occupation (in the case of a job that requires vigorous manual labor) settings.

Yet another unique design objective of the Neptune processor was the fact that it was developed to look similar to many personal consumer electronic devices (see Figure 1–16). Specifically, it resembles MP3 players, which are commonly used by persons with normal hearing. Furthermore, the Neptune sound processor offers all of the same sound processing strategies and input processing technologies as the Harmony sound processor (discussed later in this chapter) including HiRes Fidelity 120 sound processing and the ClearVoice strategy, proven to improve performance in noise (Koch et al., 2014). Programs created for the Harmony processor may simply be transferred to the Neptune processor. Of course, differences in microphone placement for the Harmony and Neptune sound processors may result in differences in sound quality and/or performance between the two sound processors, even when the same program is used. It should be noted that a special T-Mic may be ordered for use with the Neptune processor.

One other novel aspect of the Neptune processor, shown in Figure 1–16, is the fact that it may be used in two configurations: (1) the Neptune Connect mode, and (2) the more streamlined, water-proof mode, in which the controls are fully removable to allow for the smallest wearing configuration. The Neptune Connect configuration is the mode of operation intended for daily use and is often used with the UHP described in the Naída CI Q70 section of this chapter.

The Neptune Connect also offers user-adjustable dials that allow the recipient to change the volume and sensitivity of his or her processor (if the clinician has programmed these controls to be enabled for use by the recipient). The Neptune Connect also provides a slider-program switch (Figure 1–17), which provides the user with three different programs as well as a special setting (depicted as a triangle) referred to as the "monitor" mode. This mode allows a caregiver or clinician with normal auditory function to listen to the input from the microphone or Euro Port Connector with a set of conventional earphones plugged into the 3.5-mm auxiliary connector of the Neptune Connect. Furthermore, the Neptune Connect has a Euro Port (see Figure 1–16) that allows for the connection of a typical Euro Port FM/Roger receiver. Finally, the Neptune Connect has an audio input port to allow the user to connect personal audio devices (e.g., MP3 players, tablets, notebook computers) with an auxiliary cable with a 3.5-mm phono plug. It should be noted, once again, that the Neptune processor is not waterproof when the Neptune Connect configuration is used.

To use the Neptune as a waterproof processor, the recipient must remove the Neptune Connect (see Figure 1–16B), which makes the Neptune controls inaccessible. Management of Neptune controls for both the Connect and swimmable options is discussed in Chapter 4. The recipient must also switch from the UHP to the slightly

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**FIGURE 1–17.** Advanced Bionics Neptune program switches. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

thicker AquaMic waterproof headpiece (see Figure 1–16B) when using the Neptune for water-related activities. Accessing the magnet well for the AquaMic is similar to the process described for the UHP. The AquaMic also has an external magnet; however, it may also require use of a special "thin magnet" packaged separately. In order to achieve the IP68 water resistance, the AquaMic headpiece must also be used with the AquaMic cable. Of practical note, the underneath side of the UHP is black, whereas the underneath side of the AquaMic is gray. A helpful mnemonic to help the clinician or recipient distinguish between the two headpieces is the phrase "gray for water play" to indicate the gray headpiece as the AquaMic. Many recipients prefer to use the Neptune in the swimmable configuration, particularly in active situations, because it is significantly smaller than the Connect configuration.

### *Pediatric-Friendly Features*

The first pediatric-friendly feature on the Neptune is an LED within the power button (see Figure 1–16) to alert the user or caregiver of the status of operation of the devices as well as to program changes. Figure 1–18 provides a handout from Advanced Bionics that explains the information provided by the LED light. Second, the Neptune also features an audible alarm that can be programmed to alert the caregiver when the headpiece has fallen from the child's head (i.e., lock is lost between the external sound processor and the cochlear implant internal device) or when the battery is nearly depleted. Third, the covers of the Neptune case are difficult to remove without a specific hand movement or removal tools (see Figure 1–13). Finally, the Neptune sound processor may be used in with several different wearing accessories (Figure 1–19). The authors' anecdotal experience indicates that many Advanced Bionics recipients report use of their Neptune during recreational activities, such as tennis, basketball, and so forth.

### *Neptune Battery Options*

Another attractive feature of the Neptune processor is the fact that it is powered by one AAA battery. Most recipients use rechargeable NiMH AAA batteries, which are provided by Advanced Bionics along with a AAA/AA battery charger. The user may also use a AAA alkaline battery in instances in which a power source is unavailable to charge rechargeable batteries. Advanced Bionics reports that the typical user will receive about 13 hours of use with a AAA NiMH rechargeable battery, 20 hours of use with a lithium battery, and 12 hours of use with a AAA alkaline battery. There are at least two benefits of using AAA batteries to power the Neptune. First, AAA batteries are readily available at most every department store, pharmacy, and convenience store around the world, so the recipient does not need to order specialized batteries from the implant manufacturer. Second, because of the widespread commercial application and availability of AAA batteries, they are much more affordable than the batteries that are specifically developed for use with implant sound processors.

## Harmony Sound Processor

The previous external sound processor for the Advanced Bionics Corporation, the Harmony sound processor, is shown in Figure 1–20. The Harmony sound processor possesses a digital signal 16 bit (96 dB) front end processor, sampling rate of 17.4 kHz, maximum input dynamic range of 96 dB (i.e., maximum range of sound inputs handled without distortion), and a built-in, manually accessible telecoil. The Harmony sound processor has two user controls: (1) a rotary volume control on the anterosuperior side of the processor and (2) a three-position slider switch just below the volume for three distinct user programs. A LED (light emitting diode) light on the upper spine of the processor allows for troubleshooting and for indication of device status. A post at the top of the sound processor is used for coupling to specialized earhooks. The primary microphone (an omnidirectional electret microphone) of the sound processor is located at the base of this post, which also serves as an input for auxiliary signal sources.

The conventional earhook for the Harmony may be replaced with the T-Mic, iConnect, and Direct Connect earhooks. The conventional earhook does not deliver an auxiliary signal to the

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Recipient's Name:

❑ Right Ear
❑ Left Ear

## Understanding Neptune Processor Indicator Lights

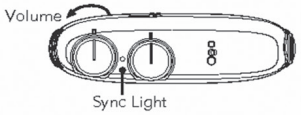
| Color | Behavior | Programmable | Indication |
|---|---|---|---|
| Orange | Blinks at start-up | No | • 4 quick blinks indicate that battery is fully charged<br>• 2-3 quick blinks indicate that the battery is sufficiently charged to power the system<br>• 1 quick blink indicates that the battery is nearly depleted<br>• No blinking indicates depleted battery. Replace with charged or new battery |
| | Solid | Yes | Battery is almost depleted |
| | Blinks twice every three seconds | Yes | Battery is depleted and cannot support stimulation |
| | Fades out | No | Sound processor is powering down |
| Red | Blinks once per second | Yes | Loss of lock with the implant |
| | Blinks rapidly *(more than once per second)* | No | Wrong implant connected |
| | Solid | No | Sound processor error condition. Fully remove and re-insert battery to reset processor |
| Green | Flickers in response to loud inputs | Yes | The sound processor and microphone are responding to sound |
| | Solid green | No | Indicates an empty program slot or monitor (△) position |
| | Blinks at start-up after battery status and upon program change | No | • 1 blink indicates program one<br>• 2 blinks indicate program two<br>• 3 blinks indicate program three |

## Understanding Processor Alarms

| Behavior | Indication |
|---|---|
| Beeps once per second | Loss of lock with the implant |
| Beeps rapidly (more than once per second) | Wrong implant connected |
| Beeps slowly (once every five seconds) | Low battery |
| Two beeps every three seconds | The battery is depleted and cannot support stimulation |
| Continuous beep | Sound processor error condition |

## Syncing the Neptune Connect and Processor

1. Turn the volume all the way down by moving the dial in a counter-clockwise direction.
2. Sync Light will flash green once.
3. Return volume dial to typical use setting.




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## Understanding Sync Light Indicators

| Color | Behavior | Indication |
|---|---|---|
| Green | Blinks once | Neptune Connect and processor settings are synced |
| Red | Blinks once per second | Neptune processor and Neptune Connect settings are **not** synced |
| | Solid | Error Condition. If this occurs, remove and re-attach the Neptune Connect |

NEPTUNE™ Sound Processor

**FIGURE 1–18.** Advanced Bionics handout about Neptune processor indicator lights. *Note.* Provided courtesy of Advanced Bionics.




**FIGURE 1–19.** Advanced Bionics Neptune Wearing Accessories. *Note.* Images provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

sound processor and serves only to retain the device over the ear. The iConnect earhook is used to wirelessly couple a miniaturized, personal frequency modulation (FM) receiver with a three-prong Europlug connection. The Direct Connect earhook along with an accessory audio cable allows the user to connect to a MP3 player, a computer, or any other recreational audio device.

The Harmony features a socket for the headpiece. Advanced Bionics uses the term "headpiece" to describe the external transmitting cable, the transmitting coil/antenna, and the external magnet. The UHP is the primary headpiece available for use with the Harmony sound processor, but a different cable (e.g., the Harmony UHP cable) is required than what is used for the Naída and Neptune processors. Previously, the Harmony processor was used with the same headpiece that was designed for its predecessor, the Auria sound processor. That headpiece, which is no longer distributed, was available in two shapes, a flat shape (designed for use with the CII internal device) and a concave shape (designed for use with the HiRes 90K internal device) that conforms to the curvature of the head and improves retention. A magnet housed inside the headpiece allows for coupling to the receiving coil and magnet of the internal device. If the magnet strength does not allow for sufficient retention to the head, a separate magnet can be inserted into a shallow well located on

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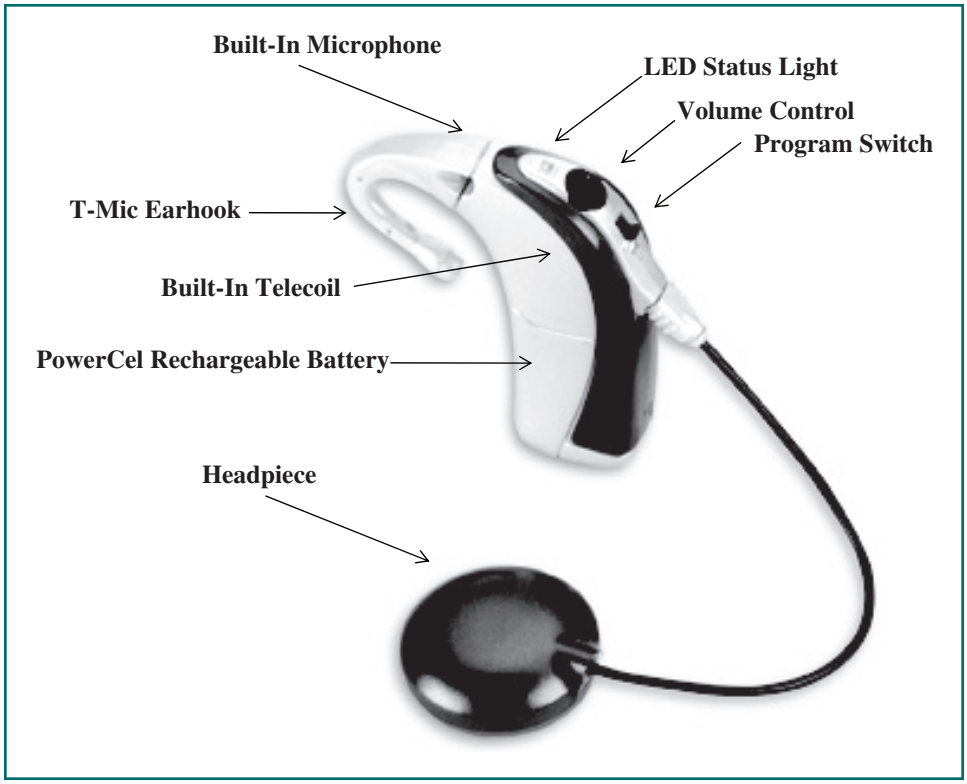



**FIGURE 1–20.** Advanced Bionics Harmony sound processor. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

the underside of the headpiece. It should be noted that the cable of the older model Auria/Harmony headpiece was hard-wired into the headpiece, so when the cable failed, the entire product had to be replaced. This was one of the primary reasons the headpiece was replaced with a separate Harmony cable and the UHP headpiece coil.

The Harmony sound processor is powered by a lithium ion rechargeable battery (i.e., PowerCel). It is available in two different sizes, a compact version (PowerCel Slim) and a larger version (PowerCel Plus). The former version should be suitable for all-day use for many users; however, the latter is used when more power is required to last through an entire day (as might be the case for recipients with exceptionally high stimulation levels, thick skin flaps, or unusually high electrode impedances). These battery packs are easily coupled to the device by sliding onto tracks located on the processor.

## HiRes 90K Advantage Implant

In September 2012, Advanced Bionics received FDA approval for the HiRes 90K Advantage implant (Figure 1–21). The HiRes 90K Advantage contains the same electronics as the HiRes 90K implant, which is discussed later in this chapter. The primary difference between the two implants is an improvement in the mechanical design of the antenna coil. The case of the HiRes 90K Advantage was designed to withstand high impact forces (i.e., blow to the head, automobile accident, fall) safely enduring an impact of the six joules, giving it the highest rated impact resistance in the industry. For the Advantage implant, the antenna coil was reinforced with a high-density polymer fiber component to address a small number of previous failures in which the coil had failed. The fiber, positioned adjacent to the braided gold coil wire in the antenna, offers resis-

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**FIGURE 1–21.** Advanced Bionics HiRes 90K Advantage implant. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

tance to any movement in the Silastic carrier that would otherwise induce a tensional force in the gold wire.

### *Mid-Scala Electrode Array*

In May 2013, Advanced Bionics commercially released the HiFocus Mid-Scala electrode array in the United States. At the time of the development of the HiFocus Mid-Scala array (Figure 1–22), cochlear implant manufacturers and researchers were exploring and continue to explore the theoretical advantages and limitations of perimodiolar or "modiolar-hugging" (i.e., near the modiolus, which is on the medial wall of the cochlea and is in close proximity to the auditory nerve fibers) versus lateral wall electrode arrays (which lie away from the medial wall, under the basilar membrane). The overall prevailing themes are fairly straightforward. Specifically, a perimodiolar array should require less electrical current to facilitate a response from the auditory nerve and thereby require less power to achieve a given loudness percept. In addition, perimodiolar placement may result in reduced channel interaction (electrical crossover from the electrode intended for stimulation to surrounding electrodes). On the other hand, lateral wall electrodes may be less traumatic for the delicate cochlear structures in the scala media, and consequently, maybe ideal for preserving residual hearing and for preserving the cochlea for future medical interventions (i.e., hair cell regeneration). Indeed, research has demonstrated that recipients of perimodiolar electrode arrays require lower levels of electrical stimulation to achieve and given level of loudness and performance (Cohen, Saunders, & Clark, 2001; Tykocinski et al., 2001), and channel interaction may be lower in these recipients (Hughes & Abbas, 2006).

Although studies have shown that some recipients of lateral wall electrodes do achieve almost full preservation of low-frequency residual hearing (Gantz & Turner, 2004; Gantz, Turner, & Gfeller, 2004; Gstoettner et al., 2009; von Ilberg, Baumann, Kiefer, Tillein, & Adunka, 2011), there are also certainly instances in which hearing preservation has been achieved with perimodiolar electrodes. To the author's knowledge, there is no published research available at this time to indicate that lateral wall electrode arrays enhance the likelihood that a recipient will be able to pursue inner ear pharmacological treatments that may arise in the future.

In the midst of these uncertainties, the HiFocus Mid-Scala was developed in an attempt to capitalize on the theoretical advantages of both a perimodiolar and a lateral wall electrode array. As the name implies, the Hi-Focus Mid-Scala array is designed to lie in the center of the scala tympani

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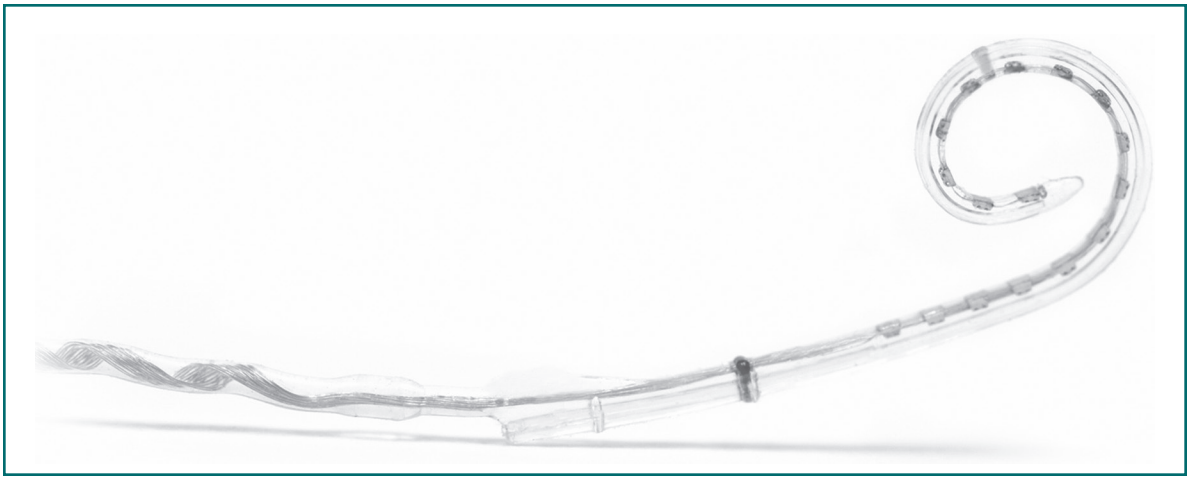


**FIGURE 1–22.** Advanced Bionics HiFocus Mid-Scala electrode array. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

in order to remain reasonably proximal to the modiolus (and cochlear neural elements) while also not making physical contact with the delicate cochlear structures necessary to maintain residual hearing. At the time of this writing, the HiFocus Mid-Scala array has not been available long enough to report surgical and performance outcomes in peer-reviewed publications.

It is also important to note that the HiFocus Mid-Scala array is much thinner than other Advanced Bionics electrode arrays. In fact, it is the thinnest precurved electrode array in the industry (lateral wall electrode arrays are not precurved, because there is no intention to maintain proximity to the modiolus). A thinner diameter is important, because the array is less likely to come in contact with delicate cochlear structures, and the array is more likely to be successfully inserted through the round window. Some surgeons and researchers believe that round window insertion provides a higher likelihood of hearing preservation and cochlear atraumaticity than insertion of the array into a cochleostomy. However, conclusive data do not exist to clearly indicate one entry point (round window versus cochleostomy) to be superior over another. At any rate, with a thinner electrode array, the surgeon is able to make a smaller cochleostomy, which may be beneficial.

Like the conventional HiFocus electrode arrays, the HiFocus Mid-Scala electrode possesses planar (flat, plate-like) electrode contacts, which are positioned for direct stimulation toward the modiolus and hence the neural elements. The HiFocus Mid-Scala electrode array is intended to be inserted to an angular depth of approximately 420°. Additionally, the HiFocus Mid-Scala electrode array is designed so that it may be inserted with stylet tool or by "free hand." It is also possible to reload the array onto the stylet if the surgeon needs to do so.

### *HiRes 90K Implantable Cochlear Stimulator*

A picture of the previous generation Advanced Bionics HiRes 90K internal stimulator (i.e., HiRes 90K Implantable Cochlear Stimulator or ICS) and HiFocus electrode is provided in Figure 1–2. The characteristics of devices released prior to the HiRes 90K are provided in Table 1–1. The entire internal device package is hermetically encased in a biocompatible silicone covering to prevent rejection by the human body and reduce the likelihood of bodily fluids entering the internal device. The electronics package of each internal device includes a DSP that performs complex analyses on the signal received from the external sound processor and determines the magnitude of electrical stimulation for each electrode contact. The HiRes 90K device features a DSP with a sampling

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rate of 90,000 updates/second and a stimulation rate of 83,000 pulses/second. It has 16 independent output circuits (each with its own capacitor), which are each linked to one of the 16 individual intracochlear electrode contacts. The independent circuits allow for a high level of control over the stimulation provided to the user. The internal device also includes a bidirectional telemetry system, which enables continuous transmission of information from the external sound processor to the internal device and vice versa. The signal from the external processor is transmitted at a carrier frequency of 49 MHz, whereas the signal from the internal device is transmitted at 10.7 MHz.

The internal receiving/telemetry coil is shown in Figure 1–2 with the internal magnet positioned within a Silastic sleeve in the center of the coil. The Silastic sleeve permits temporary removal of the internal magnet so that the patient may undergo magnetic resonance imaging (MRI) with a strength of 1.5 tesla. Also, a 0.3 tesla MRI may be completed with the magnet in place. In either case, the recipient's surgeon should ideally be consulted prior to the procedure. The internal stimulator, which is encased in a durable, biocompatible titanium material, is positioned just below the telemetry coil. The primary reference electrode (i.e., return or ground contact) is located in the center of the casing of the internal stimulator.

Advanced Bionics uses the term electrode array to collectively describe the electrode lead and the electrode array. The electrode array houses the electrode contacts that ultimately provide electrical stimulation to the cochlea. The electrode lead exits the internal stimulator through a component called a feedthrough tube. There is a fantail shape to the lead as it exits the internal stimulator, which provides flexibility and durability to the lead. The electrode lead contains 16 small wires that carry the signal from each of the 16 independent output circuits in the internal stimulator to the 16 individual electrode contacts. Medial to the fantail is a secondary return (ground) electrode referred to as the ring electrode. In rare cases, the clinician may switch the return electrode used for stimulation from the case ground electrode to the ring electrode. The ring electrode is also used as the reference electrode for neural responses from the auditory nerve using the Neural Response Imaging (NRI) assessment module. The NRI module is the platform in the Advanced Bionics system used for measurement of the electrically evoked compound action potential.

The Advanced Bionics HiRes90K internal device features two additional electrode array choices, the HiFocus Helix electrode array and the HiFocus 1j electrode array (aside from the Mid-Scala array). Both electrode arrays possess nonstimulating markers, which are used by the surgeon as a guide to determine how deeply to insert the electrode array. Medial to these markers are 16 platinum-iridium alloy electrode contacts numbered 1 through 16 from apex (lowest frequency) to base (highest frequency). The HiFocus electrode array has planar or "plate" electrode contacts that are positioned toward the modiolus to collectively provide a more focused delivery of stimulation to the auditory nerve.

The Helix electrode array has a subtly coiled shape, which is intended to facilitate close proximity of the electrode contacts to the modiolus and peripheral auditory neural elements. The Helix electrode is tapered so that the apical end is much thinner than the basal end to allow the surgeon better visualization of the cochleostomy during electrode array insertion. The Helix electrode is 0.7 mm in diameter at the apical end and 1.2 mm at the basal end. This type of electrode design is often referred to as a perimodiolar ("peri" refers to near or around the modiolus) electrode array. The theory behind a perimodiolar design is to allow for a reduction in stimulation current levels required to elicit an appropriate loudness percept for the user. However, research comparing the performance with perimodiolar Helix and the HiFocus 1j electrode arrays fails to show significant differences.

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The HiFocus 1j electrode has a gently curved electrode array and is smaller in overall diameter from the basal end to the apical tip than the Helix electrode array. Specifically, the diameter of the HiFocus 1j electrode is 0.4 mm at the apical end and 0.8 mm at the basal end. Therefore, the 1j electrode permits a smaller cochleostomy for insertion. This array is often selected for a revision surgery, which refers to the removal of a faulty device and subsequent replacement of a new cochlear implant. After insertion of the initial

electrode array, fibrous tissue may develop in and around the electrode array. As a result, when the array is removed, the channel where it resided may be quite small in diameter. As a result, the narrower diameter of the 1j facilitates an easier insertion of the replacement array. The selection of the most appropriate electrode for a recipient often depends on the personal preference of the surgeon.

## COCHLEAR CORPORATION

In August of 2013, Cochlear Americas received Food and Drug Administration (FDA) approval to commercially distribute the Nucleus 6 cochlear implant system , which includes the CP910 and CP920 external sound processors (Figure 1–23), the CR210 and 230 wireless remote assistants (Figure 1–24), and the Nucleus Freedom and Nucleus CI422 internal devices (Figure 1–25). The system is programmed in the CustomSound 4.0 software platform. In 2014, Cochlear Americas also received FDA approval to distribute the Nucleus Hybrid cochlear implant, which may also be used with the Nucleus 6 sound processor. The Nucleus Hybrid cochlear implant is discussed in later in this chapter and also in detail in Chapter 9 of this book.




**FIGURE 1–24.** Cochlear Nucleus 6 CR230 (**A**) and CR210 (**B**) wireless remote assistants. *Note.* Provided courtesy of Cochlear™ Ltd.

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**FIGURE 1–23.** Cochlear Nucleus 6 CP910 (**A**) and CP920 (**B**) external sound processors. *Note.* Provided courtesy of Cochlear™ Ltd.

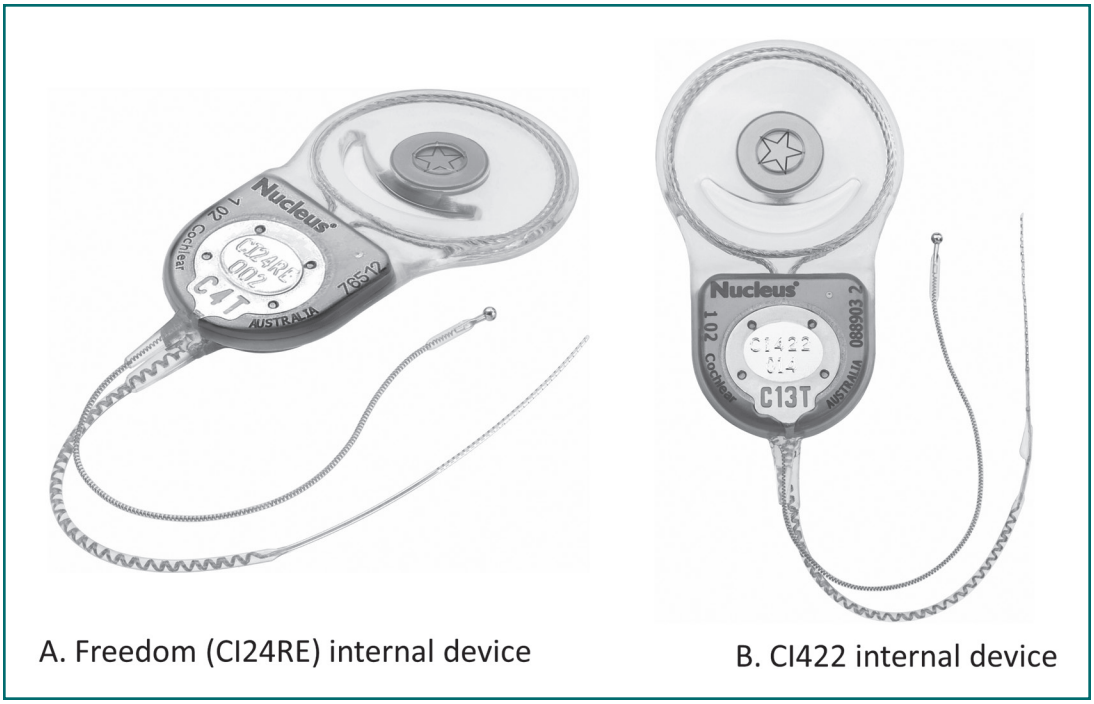



**FIGURE 1–25.** Cochlear Nucleus Freedom CI24RE Full-Band Straight Implant (**A**) and CI422 Slim Straight Implant (**B**). *Note.* Provided courtesy of Cochlear™ Ltd.

## Nucleus 6 Sound Processor

The Nucleus 6 sound processor is available in two different models, the Nucleus CP910 and Nucleus CP920 (see Figure 1–23). The primary difference between the two models is that the CP920 possesses a direct auditory input (DAI) port, whereas the CP910 does not. As a result, the recipient may couple a personal FM (RF) receiver or a personal audio device, such as an MP3 player or mobile telephone (by way of the audio accessory cable), to the CP920 sound processor but not to the CP910. In contrast, the CP910 sound processor is 6% smaller than the CP920 and Nucleus 5 sound processors (see Figure 1–23), and at the time of this writing, it is the smallest commercially available sound processor in the commercial cochlear implant market. It should be noted that both the CP910 and CP920 sound processors are compatible with Cochlear's proprietary wireless hearing assistance technologies (this portfolio of wireless accessories, which were designed by the GN ReSound hearing aid company, is described in detail later in this chapter).

The remaining electronic characteristics of the CP910 and CP920 are identical, and as a result, the processors are referred to collectively as the Nucleus 6 sound processor for the remainder of this text unless a distinction must be made regarding the DAI functionality or size difference between the two devices. The Nucleus 6 sound processor features full DSP and three major components: (1) a cable/coil with a removable magnet, (2) a battery module, and (3) a sound processor with user controls.

The Nucleus 6 sound processor also includes two omnidirectional microphones (Figure 1–26) to collect acoustic inputs, a digital circuit to analyze the signal from the microphones and determine the magnitude of stimulation to be provided to the electrode contacts, and a telecoil to allow for improved hearing on the telephone and with induction loop hearing assistance technology (HAT). Additionally, the telecoil of both processors is an automatic telecoil, which is designed to automatically detect the presence of signals from the telephone and induction loop systems when used by the recipient. Furthermore, the Nucleus 6

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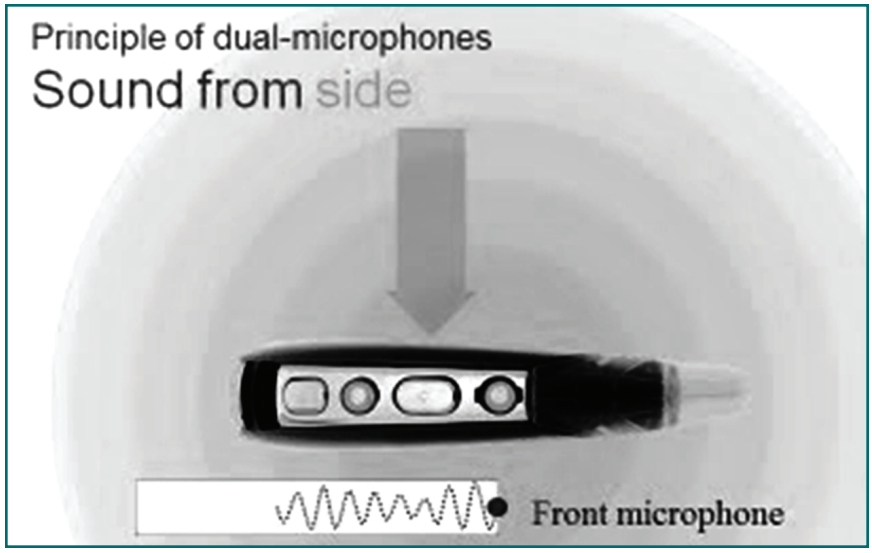



**FIGURE 1–26.** Cochlear Nucleus 6 sound processor with two omnidirectional microphones. *Note.* Provided courtesy of Cochlear™ Ltd.

sound processor possesses an acoustic driver and the capability to replace the conventional earhook with a receiver-in-the-canal acoustic component (Figure 1–27). This option, which is discussed in greater detail later in this chapter and in Chapters 5 and 10, allows for the delivery of low-frequency acoustic information for recipients who have low-frequency residual hearing after implant surgery. Finally, the Nucleus 6 sound processor features a 2.4-GHz radio frequency receiver to allow for audio streaming from a remote microphone or a mobile telephone or television via an accessory transmitter designed specifically for the Nucleus 6 via a partnership between Cochlear and the GN ReSound hearing aid company. Wireless streaming with the Nucleus 6 sound processor is discussed in further detail later in this chapter and in Chapters 5 and 9.



**FIGURE 1–27.** Cochlear Nucleus 6 sound processor with the receiver-in-the-canal acoustic component. *Note.* Provided courtesy of Cochlear™ Ltd.

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### *Digital Signal Processing (DSP)*

Additionally, Cochlear reports that the DSP chip of the Nucleus 6 processor is five times more powerful than the DSP of its predecessor, the Nucleus 5 sound processor. The Nucleus 6 has several upgraded signal processing features compared to the Nucleus 5, and as a result, it was

imperative to design the DSP chip that could support these features while also not using more electrical power. The DSP chip of the Nucleus 6 features six DSP cores that are controlled by a 16-bit microcontroller. In comparison, the Nucleus 5 sound processor (with the CHAMP-L ASIC chip) possesses four separate digital cores, which are controlled by an 8-bit microcontroller. The six different cores present within the processor allow for multiple operations to be executed simultaneously (e.g., directional processing, scene analysis, compression), whereas the microcontroller serves to integrate the functions of the six digital cores to ensure efficient and effective signal processing and stimulus delivery. Furthermore, the DSP of the Nucleus 6 sound processor features a processing speed that is twice as fast as the DSP of the Nucleus 5 sound processor. The Nucleus 6 DSP is a 16-bit processor with a 16,000-Hz input sampling rate. The overall input dynamic range for broadband sounds delivered to the Nucleus 6 sound processor is just shy of 80 dB SPL (i.e., a low end of approximately 23 dBA for broadband inputs with an upper range of just over 100 dBA). Additionally, the Nucleus 6 sound processor has a hardware accelerator, which allows for audio streaming. This was an essential inclusion in order to allow the N6 processor to receive the 2.4-GHz digital RF signal that is streamed from the wireless accessories developed by the GN ReSound hearing aid company (explained in detail later in this chapter and in Chapter 5). Finally, the DSP of the Nucleus 6 possesses a substantially larger memory relative to the Nucleus 5 sound processor in order to allow for the comprehensive datalogging feature available within the Nucleus 6 processor.

Likewise, the Nucleus 5 (CP810) sound processor possesses a DSP that is substantially more sophisticated than the DSP available in its predecessor, the Nucleus Freedom processor. The Nucleus 5 sound processor features the CHAMP-LP ASIC digital processing network, which contains four separate but interrelated DSPs controlled by a separate digital microcontroller. The DSP network possesses an asynchronous architecture that allows several different functions to occur simultaneously, but in an integrated fashion, to allow for optimal efficiency and effectiveness in signal processing. For instance, the level detector and noise management (i.e., autosensitivity) are both processing the signal at the same time to enhance efficiency, but the microcontroller evaluates the information from each processor to determine the signal output necessary to optimize signal audibility and comfort for the user. Overall, this DSP network features 16-bit converters, with a sampling rate of 16000 Hz and an input dynamic range of approximately 75 dB, which in this case refers to the range between the softest sound the CP810 can detect to the most intense signal it can handle without distortion.

### Wearing Options

The CP910 and CP920 sound processors are designed to be worn as an ear-level processor with a standard battery module (Figure 1–28), but there are several wearing options designed to facilitate use on the ear for a variety of different use applications. Multiple earhook sizes are available. The "small" earhook is often used by children and adults with small ears. The small earhook may also be configured with a small pin, which must be inserted through a hole at the base of the ear



**FIGURE 1–28.** Cochlear Nucleus 6 sound processor worn with the standard battery module.

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hook where it attaches to the processor. This pin is intended to allow a young child's caregiver to "lock" the earhook to the processor to prevent it from becoming accidentally dislodged. Earhooks in "standard" and "large" sizes are also available for adults and older children for whom the "small" earhook is too small.

Nucleus 6 users may also use the "Snugfit" device, which features a pliable, "C-shaped" earhook that may be looped around the auricle to enhance retention (Figure 1–29). The Snugfit device is also available to various sizes to accommodate different ear sizes, and a tamper-proof Snugfit option is available. The recipient may also use the "Mic Lock," which is composed of a small loop that couples to the processor and is attached to a piece of tubing that may be connected to the earhook of the processor (Figure 1–30).

Finally, "off-the-ear" wearing options also exist. The battery module may be attached to the sound processor via a long cable and clipped to the user 's clothing via a specialized holder (Figure 1–31). This "body-worn" option, which is referred to as "Litewear," may be selected for very young children (Figure 1–32) and recipients involved in sports because it minimizes the size of the unit worn on the ear. Of course, the Nucleus 6 sound processor may also be attached to a standard ear mold to facilitate retention.



**FIGURE 1–29.** Cochlear Nucleus 6 sound processor Snugfit device. *Note.* Provided courtesy of Cochlear™ Ltd.

For recipients who use the acoustic component, the Nucleus 6 sound processor will be attached to the acoustic receiver, which is coupled to either a generic dome or to a custom-fitted acrylic ear mold. More specifically, the earhook of the Nucleus 6 sound processor may be removed and replaced with a receiver-in-the-canal (RIC) acoustic component (see Figure 1–27). The Nucleus 6 sound processor possesses an acoustic driver to operate the acoustic component, which may be used to amplify low-frequency sounds for recipients who have preservation of low-frequency hearing following implant surgery. The Nucleus 6 acoustic component may be worn with either an open or closed dome or with a custom-made shell. The open dome is likely to be more appropriate for recipients who have normal hearing up to a mild hearing loss at 500 Hz and below, while the closed dome and custom shell options are likely to be more appropriate for recipients who have more significant amounts of low-frequency hearing loss. In particular, the custom shell will probably be imperative for recipients who have moderate to severe low-frequency hearing loss. At the time of this writing, there have been no research studies examining the relationship of vent size to sound quality or speech recognition for implant recipients using electroacoustic stimulation, so specific, evidence-based recommendations for electroacoustic users are unavailable.

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Additionally, at the time of this writing, there is a paucity of published studies examining the efficacy of low-frequency amplification for implant recipients as a function of the degree of their low-frequency hearing loss. It should be noted that Cochlear suggests clinicians consider amplification when residual hearing thresholds are better than 90 dB HL. Additional information on clinical management of electroacoustic stimulation is provided in Chapter 10 of this textbook.



**FIGURE 1–30.** Cochlear Nucleus 6 sound processor Mic Lock.



**FIGURE 1–31.** Cochlear Nucleus 6 sound processor off-the-ear battery wearing option, known as Litewear. *Note.* Provided courtesy of Cochlear™ Ltd.



**FIGURE 1–32.** Cochlear Nucleus 6 sound processor Litewear battery option shown on a young child. *Note.* Provided courtesy of Cochlear™ Ltd.

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Cochlear has developed two options to allow the recipient to swim while using the CP900 sound processor. The Aqua Accessory (Figure 1–33) is approved by the FDA and designed to allow for a waterproof solution for recipients who desire to swim with their processor. The Aqua Accessory provides an ingress protection (IP) rating of 68, which suggests that the product is fully protected from dust and is also protected when immersed under water for long periods of time. The Aqua Accessory should be used only with rechargeable batteries, and the microphone covers should be in place while swimming with the Nucleus 6 sound processor and the Aqua Accessory.

The Aqua+ (Figure 1–34) is another accessory that may be used to allow the processor to be waterproof while swimming. The Aqua+ is a harder plastic case, which can house the processor and also may be used multiple times without being replaced. The Aqua+ is considerably smaller than the Aqua Accessory. At the time of this writing, the Aqua+ was recently approved by the FDA for commercial use in the United States. Use of the Aqua+ accessory also provides an ingress protection rating of 68, which allows the Nucleus 6 sound processor to function as a waterproof/swimmable processor. It should be noted that the Aqua+ and Aqua Accessory products may also be used with the Nucleus 5 sound processor.

In summary, the manufacturer designed the N6 processor to have cosmetic appeal, comfort, and good retention when worn on the ear. The CP910 model is currently the smallest external sound processor on the market. The authors' preliminary experience, as part of the third phase of an FDA-approved validation study, is that adult and pediatric recipients are able to comfortably wear the Nucleus 6 sound processor with good retention during daily use and during strenuous activity. Also, it is worth noting that most recipients do not require use of a personal ear mold to facilitate retention.

### Battery Options

As shown in Figure 1–35, the battery is available in three configurations: (1) a smaller, compact rechargeable battery module (a), (2) a standard rechargeable battery module using a built-in lithium-ion battery (b), and (3) a standard, zinc-air battery module with two #675 zinc-air batteries (c and d). Tamper-



**FIGURE 1–33.** Cochlear Nucleus 6 Aqua Accessory. *Note.* Provided courtesy of Cochlear™ Ltd.



**FIGURE 1–34.** Cochlear Nucleus 6 Aqua+ case. *Note.* Provided courtesy of Cochlear™ Ltd.

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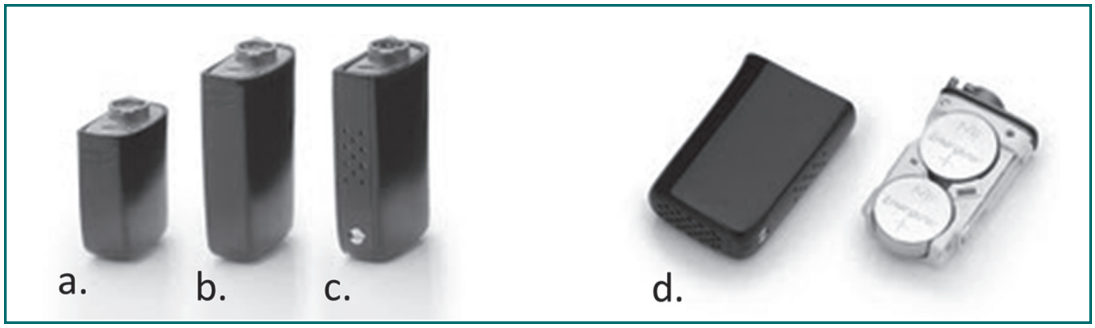



**FIGURE 1–35.** Cochlear Nucleus 6 battery options: **a.** compact rechargeable battery module; **b.** standard rechargeable battery module; **c** and **d.** standard, zinc-air battery module with two #675 zinc-air batteries. *Note.* Provided courtesy of Cochlear™ Ltd.

proof locks are available to prevent young children from removing each of the different types of battery modules from the processor as well as the cover case from the disposable battery module. When using the standard, zinc-air battery module, the Nucleus 6 sound processor has a water resistance rating of IP-44 (see Tables 1–4 and 1–5). Cochlear reports that the standard, zinc-air battery module will provide up to 60 hours of battery life. It may be splashed with water and still maintain normal function. When used with the rechargeable battery modules, the Nucleus 6 has an IP-57 rating, which means it will retain normal function after being submerged in 3 feet of water for up to 30 minutes. Cochlear developed the rechargeable batteries with the objective that they will provide the typical recipient with one full day of use per charge. Recipients are able to obtain up to 30 hours of use with the standard rechargeable battery of the Nucleus 6 up to 18 hours of use with the Nucleus 6 compact battery (80% of Nucleus 6 users will obtain at least 8 hours of use from the compact rechargeable battery). It should be noted that the Nucleus 6 rechargeable batteries have a lifespan of at least 365 charge cycles.

The Nucleus 6 features titanium connectors to couple the battery module to the sound processor. The predecessor of the Nucleus 6 processor, the Nucleus 5, was the first sound processor to feature titanium connectors, a design that allows for a sturdy sound processor that is resistant to damage when accidentally dropped to the floor. The Nucleus Freedom sound processor, which preceded the Nucleus 5 processor, had plastic connectors that were prone to breaking during routine "wear and tear," especially when used by active, young children. In short, the titanium connectors on the Nucleus 5 and Nucleus 6 sound processors substantially improve the durability and reliability of these devices.

Additionally, the Nucleus 6 sound processor contains a nano-coating, which is intended to enhance the reliability of the processor. Nano-coating technology is commonly used in commercial electronics to improve water resistance, particularly in humid environments or other situations (e.g., sports, strenuous occupational settings) in which a gas may convert into a liquid (i.e., condensation). This nano-coating technology likely enhances reliability of the Nucleus 6 sound processors relative to its predecessors.

### *Processor Microphones*

The Nucleus 6 sound processor contains two omnidirectional microphones (two ports shown on the spine of the processors in Figure 1–23), which allow for higher-order directionality and adaptive beam-forming. The two microphones are calibrated on each processor at Cochlear to account for inevitable variances that occur from microphone to microphone and subsequently allow for the desired directional response (and tighter tolerances/control of the sound quality—not just directionality, but also the amplitude-frequency response). Furthermore, the implementation of the directional system can be altered via the programming software through firmware changes, a

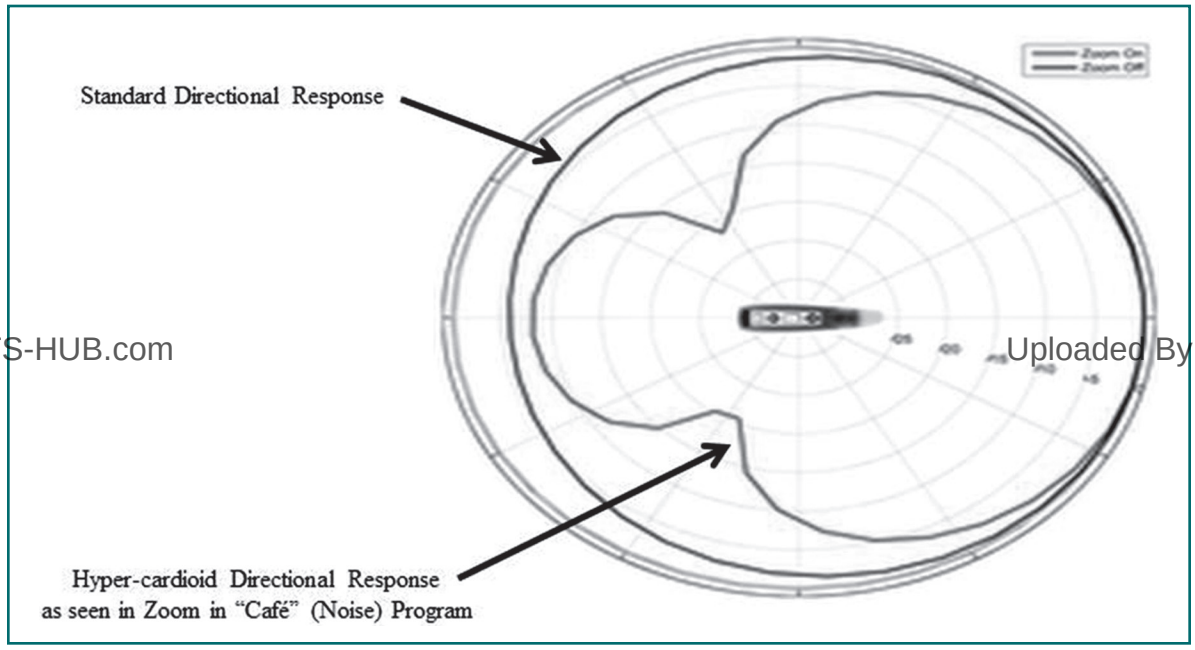
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feature that will allow access to a fully automatic, adaptive directional system (i.e., it automatically switches from omnidirectional to directional mode in noisy environment) once it is approved by the United States FDA.

In the default mode, a relatively minimal amount of directivity is provided (Figure 1–36) in order to mimic the natural and slightly positive directional index of the unaided ear (attributed to the attenuation the auricle provides primarily to high-frequency sounds arriving from behind a person). The user can manually select between programs that provide an adaptive directional response (i.e., the polar plot pattern adaptively changes to attenuate the signals from the location in the rear hemisphere where the most intense noise is arriving) or a fixed-mode response (i.e., hyper-cardioid) that offers substantial attenuation to sounds arriving from the sides and rear of the wearer. As mentioned earlier, although it is not clinically available in the United States at the time of this writing, the directional system of the Nucleus 6 is also capable of functioning in a fully automatic, adaptive manner.

The new "Scan" preprocessing algorithm performs an auditory scene analysis, which determines if speech is present in moderate- to high-level competing noise (greater than 50 dB SPL). If this occurs, the sound processor automatically switches to a more directional response and attempts to provide the most attenuation in the rear hemisphere where the most intense noise is arriving. In an environment with diffuse noise (i.e., the level of the noise is similar from all locations at which it is arriving at the head of the recipient) and no speech signal present, the sound processor will automatically switch to a fixed hyper-cardioid response (i.e., Zoom, which is described in detail in Chapter 4). When speech is present and the ambient noise level exceeds 50 dB SPL, then the processor will automatically switch to adaptive beam-forming (Beam, which is described in detail in Chapter 4). The primary objective of the adaptive beam-forming algorithm is to optimize speech recognition in noise by providing a polar plot that maximizes attenuation of the competing noise signal.

A removable polyester waterproof filter/protective cover (Figure 1–37) resides over each of the two microphones to offer protection against moisture and debris. This filter/cover should be replaced at least every 3 months or sooner if loudness is attenuated.




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**FIGURE 1–36.** Cochlear Nucleus 6 directional microphone polar plot responses. *Note.* Provided courtesy of Cochlear™ Ltd.

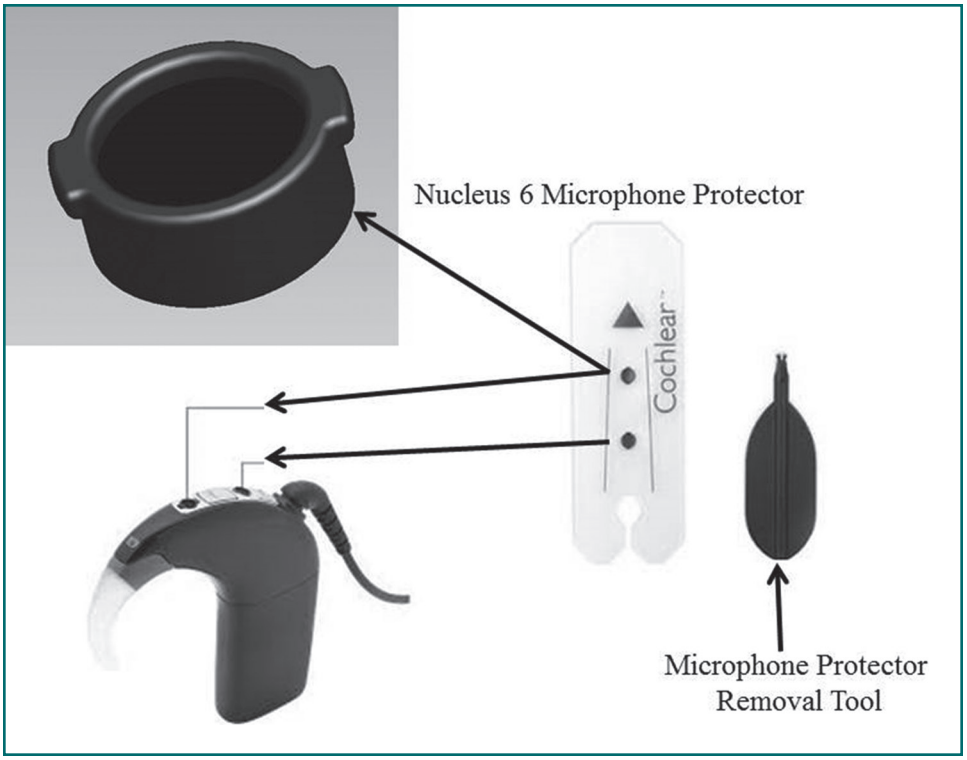



**FIGURE 1–37.** Cochlear Nucleus 6 polyester waterproof filter/protective cover for the processor microphones. *Note.* Provided courtesy of Cochlear™ Ltd.

### *Processor Buttons and Lights*

The Nucleus 6 sound processor has two push buttons for adjustments of various parameters by the recipient. In the default "Basic" mode, the bottom button allows the recipient to change to one of four programs with a short press. The top button may be used to manually activate or deactivate the built-in telecoil. The clinician may also program the processor in the "Advanced" mode, which enables the recipient to make adjustments to the volume or the microphone sensitivity. In the "Advanced" mode, a short press of the bottom or top buttons decreases and increases the volume or sensitivity (depending upon whether the clinician has programmed the processor buttons to function as a volume or sensitivity control), respectively. A long press of the bottom button changes the program, and a long press of the top button activates and deactivates the telecoil. It should be noted that the "Advanced Mode" is rarely used by adults or children. Finally, the buttons may be locked for young children to disallow changes to the controls.

Additionally, the Nucleus 6 sound processor has a bicolor LED light (Figure 1–38) at the apex of the processor adjacent to the removable earhook. The LED light indicates the status of the processor as well as the status of the signal delivered to the internal device. The light may be deactivated, programmed to indicate the function of each button push, or programmed to alert caregivers to the status of the processor during routine use (e.g., battery life, sound detection, transmission to internal device). Figure 1–39 provides detailed information pertaining to the function of the LED light.

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### *Processor DAI Port*

A DAI accessory port is located on the spine of the CP910 sound processor just below the cable (Figure 1–40). This port allows for coupling to a personal FM (RF) receiver and external accessories, such as an external microphone, MP3 player,



**FIGURE 1–38.** Cochlear Nucleus 6 LED light. *Note.* Provided courtesy of Cochlear™ Ltd.

## Lights

### Everyday use

| LIGHT | WHAT IT MEANS |
|---|---|
| ●●●●● ... | Processor flashes while receiving sound from telecoil/audio accessory (if set up by your clinician). |
| ●●●●● ... | Processor flashes while receiving sound from microphones (if set up by your clinician). |
| ●<br>●●<br>●●●<br>●●●● | Turning on and changing programs. Number of flashes indicates the number of the current program. |
| ▬ | Turning off processor. |

### Locking buttons

| LIGHT | WHAT IT MEANS |
|---|---|
| ▬ ▬ | Locking processor buttons. |
| ▬ ▬ | Unlocking processor buttons. |
| ● | Processor buttons are locked. |

### Alerts

| LIGHT | WHAT IT MEANS |
|---|---|
| ● ● ● ... | Processor flashes while coil is off (or connected to the wrong implant). |
| ●●●●● | Processor battery is empty. Charge battery. |
| ▬ | Fault. Contact your clinician. Stays on until the issue is resolved. |

### Telecoil/audio accessories

| LIGHT | WHAT IT MEANS |
|---|---|
| ●●●●● ... | Processor flashes while receiving sound from telecoil/audio accessory (if set up by your clinician). |
| ▬ | Changing from microphone to telecoil/audio accessory. |
| ▬ | Changing from telecoil/audio accessory to microphone. |

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**FIGURE 1–39.** Cochlear Nucleus 6 indicator light chart for recipients. *Note.* Provided courtesy of Cochlear™ Ltd.

or smartphone via the personal audio cable (Figure 1–41A), mains isolation cable, and monitoring earphones. The sound processor is designed to automatically recognize the connection of a DAI (and mix the signal appropriately). Two of the external audio accessories are shown in Figure 1–41 and discussed further in Chapter 9. It should be noted here that monitoring earphones (Figure 1–41B), which may be plugged into the accessory port, may be used by a person with normal hearing sensitivity to listen to inputs captured by the processor microphones or telecoil. It is important to remember that the monitoring earphones do not allow one to listen to the signal the implant user ultimately receives, but instead, the normal-hearing listener is able to evaluate the signal after it has been subjected to input processing but prior to delivery to the recipient. Finally, a special set of monitoring earphones (Figure 1–42) may also be used to evaluate the signal captured by a personal FM (RF) receiver.



**FIGURE 1–40.** Cochlear Nucleus 6 direct audio input accessory port with Personal Audio Cable. *Note.* Provided courtesy of Cochlear™ Ltd.

### *Transmitting Coil*

The transmitting cable and coil of the Nucleus 6 sound processor possess design improvements relative to the cable and coil used with the Nucleus 5 sound processor. The cable, the cable plug, and strain reliefs have been carefully modeled

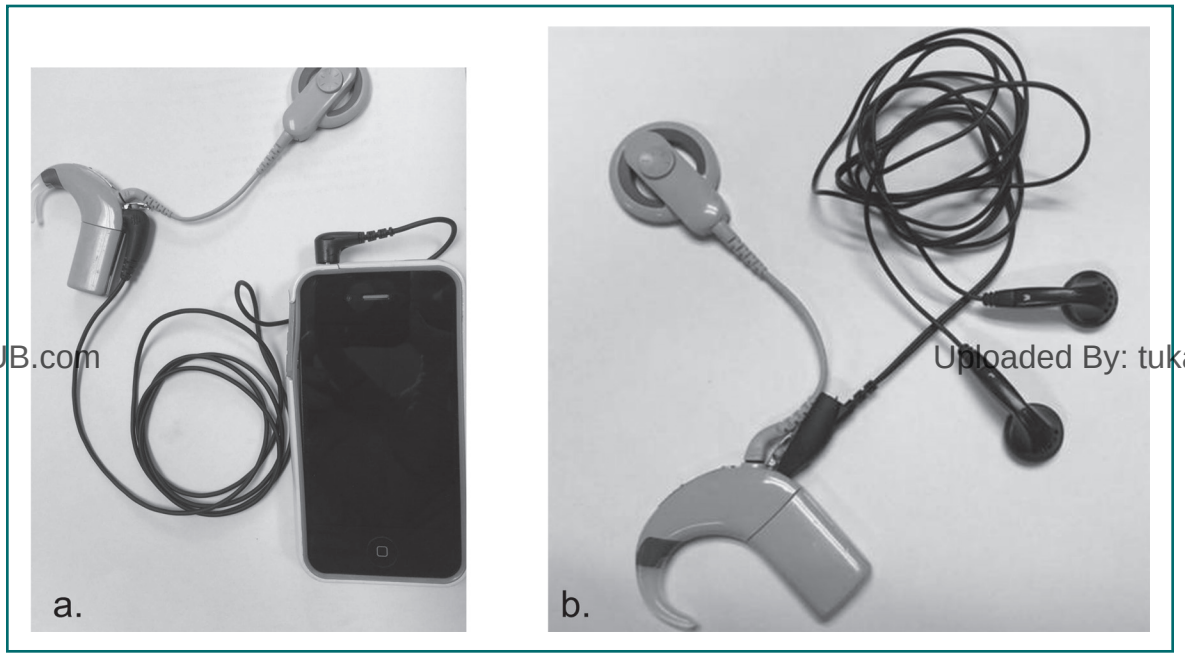



**FIGURE 1–41.** Cochlear Nucleus 6 with Personal Audio Cable coupled to a mobile phone (**a**) and monitoring earphones (**b**).

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**FIGURE 1–42.** Cochlear Nucleus 6 monitoring earphones for listening to the signal from a personal frequency modulation(FM)/ radio frequency (RF) receiver.

and reconfigured to ensure that the flexation near each of the cable connectors is displaced over a considerable length of the cable and plug in order to increase the fatigue life of the product. Also, the angle at which the cable plugs into the processor has been redesigned relative to previous processors and is intended to enhance coil retention, to facilitate a close fit of the cable to the contour of the head, and to accommodate a larger range of head sizes and implant orientations.

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### *Wireless Remote Assistants*

The Nucleus 6 processor is designed for use with either the CR210 or CR230 wireless remote assistant (see Figure 1–24 and Table 1–4). The CR230 remote assistant is essentially the new generation replacement for the CR110 remote assistant of the Nucleus 5 cochlear implant system. Similar to the CR110, the CR230 features two-way communication between the cochlear implant system and the diagnostic interface of the remote assistant, which allows the user access to a wide range of system controls and diagnostic capabilities. The CR230 remote must be paired to a Nucleus 6 sound processor by placing the processor transmitting coil on the back of the CR230 remote assistant. Not only does the CR230 allow the user to have control over a number of processor settings, but it also provides comprehensive information pertaining to processor function to the user via a full-color LED screen and an audible alarm (Figure 1–43).

The CR230 allows for control over a wide array of Nucleus 6 sound processor settings including, volume, sensitivity, program, audio accessory mixing ratio, telecoil mixing ratio, the LED and audible alarm settings, locking of the processor settings. The CR230 remote also may be used to evaluate the status of Nucleus 6 sound processors. The recipient or caregiver may check the battery life, the integrity of the processor, cable/coil, microphone, and RF coupling (i.e., Link/lock) between the external and internal components. If a problem is identified, the CR230 remote




**FIGURE 1–43.** Color screen on the Cochlear CR230 remote assistant. *Note.* Provided courtesy of Cochlear™ Ltd.



**FIGURE 1–44.** LCD screen on the Cochlear CR210 remote assistant. *Note.* Provided courtesy of Cochlear™ Ltd.

will offer a series of suggestions to the user to attempt to mitigate the problem. Additionally, the CR230 remote assistant has a rechargeable battery. Finally, the CR230 remote assistant has an audible alarm that provides an audible warning when the status of the processor is compromised (e.g., battery is low, transmitting coil falls off of the head). With the provision of control over a wide range of processor settings along with the diagnostic and troubleshooting features, the authors consider the CR230 remote assistant to be the preferred remote for use with infants and young children. Table 1–4 provides more information regarding the function of the CR230 remote assistant.

The CR210 remote assistant is significantly smaller than the CR230 remote and also possesses fewer capabilities. It provides a simple user interface that allows for the adjustment of the program, volume or sensitivity (but not both), and telecoil (enable/disable). The aforementioned settings are displayed on an LCD screen (Figure 1–44). Additional settings (accessory mixing ratio, processor button lock, LED light, etc.) cannot be controlled by the CR210 remote. The CR210 also does not allow for diagnostic assessment of cochlear implant function, and it does not possess a troubleshooting guide. It has a disposable CR2032 or a 5004LC 3v standard lithium coin cell battery, which is estimated to have a life of about 3 months if the CR210 is used about 15 times a day. The CR210 also possesses an attachable ring that allows the CR210 to be carried on a keychain. The authors consider the CR210 remote assistant to be a good choice for adult recipients and older children (particularly elderly adults who may become confused with new technology) because it is small, simple to use, and provides control over most settings an adult would prefer to adjust. Table 1–4 provides additional information about the CR210 remote assistant.

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### *Nucleus 6 Wireless Accessories*

One of the most novel additions of the Nucleus 6 cochlear implant system is the advent of a variety of wireless streaming digital hearing assistance

technology accessories that may be used with the Nucleus 6 sound processors. Cochlear Ltd. has developed a partnership with the GN ReSound hearing aid company to provide wireless streaming capability. GN ReSound has developed an array of wireless hearing assistance technologies for its hearing aid product line (Figure 1–45). These wireless accessories include the Unite Phone Clip+, the Unite Mini Microphone, and the Unite TV Streamer 2. Each of these accessories wirelessly streams audio signals directly to a receiver (i.e., antenna) housed within the Nucleus 6 sound processor via proprietary digital radio frequency transmission at the 2.4 GHz ISM (Industry, Science, Medical) band, which is globally license-free. GN ReSound reports that its proprietary system possesses a shorter delay (approximately 20 msec) from the transmitter to the receiver when compared to Bluetooth devices, and it also reports that the power requirements are significantly lower than Bluetooth transmission. These advantages are attributed to the fact that the system is proprietary and dedicated for use with GN ReSound hearing aids or Cochlear Nucleus 6 sound processors, whereas Bluetooth is an open platform used with a variety of commercial electronics. As a result, the proprietary system may be designed to be more efficient in its operation.

The Unite Phone Clip+ (see Figure 1–45) allows for streaming from a Bluetooth-enabled mobile telephone to the Unite Phone Clip+ via Bluetooth and then from the Phone Clip+ to the Nucleus 6 sound processor via GN ReSound's proprietary wireless transmission. The Unite Phone Clip+ has a lithium-ion rechargeable battery that provides up to 6 hours of talk time and 80 hours of standby use (with ReSound hearing aids) on a full charge. The Phone Clip+ must be paired to the user's mobile telephone (or other Bluetooth-enabled device) and to the user's Nucleus 6 sound processor. Pairing to the telephone is accomplished by pressing a small blue button under the case at the base of the Phone Clip+ and then activating the Bluetooth pairing process of the telephone. Once the telephone identifies the Phone Clip+, the user must enter the code "0000," and pairing to the telephone is complete. The user then presses the white button at the base of the Phone Clip+ and places to battery onto the Nucleus 6 sound processor. When the Nucleus 6 processor powers on, it enables pairing mode and will connect to the Phone Clip+.

After pairing is complete, the user should clip the Phone Clip+ next to his or her collar to allow for capture of the user's voice and delivery to the mobile telephone. Likewise, the audio signal from the telephone is also sent from the

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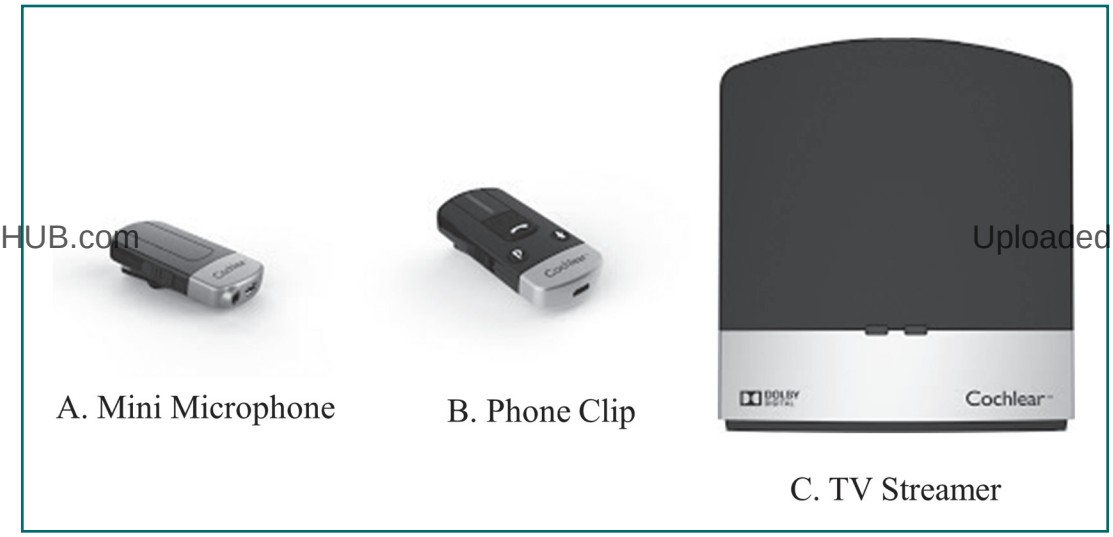



**FIGURE 1–45.** Wireless hearing assistance technologies from GN ReSound that can be paired to the Nucelus 6 sound processor. *Note.* Images provided courtesy of GN ReSound.

Phone Clip+ to the mobile telephone allowing for hands-free telephone use. The user may remain connected between the Phone Clip and two Bluetooth-enabled devices at the same time. In order to switch between streaming devices, the recipient simply presses the button on the front, center of the Phone Clip+ (shaped like a phone handset). The same button also is used to answer and end telephone calls.

The Unite Mini Microphone (see Figure 1–45) is designed to be clipped onto the shirt collar of a person with whom the Nucleus 6 user wishes to hear. Because of the close proximity to the talker's mouth, the Unite Mini Microphone captures the speech signal of interest and wirelessly streams it to the Nucleus 6 sound processor via ReSound's proprietary digital wireless transmission. The Unite Mini Microphone also possesses an audio input port allowing for a hard-wired connection to the headphone jack of any audio device (e.g., MP3 player, handheld video game, notebook computer). The audio signal may then be streamed to the Nucleus 6 sound processor.

The Unite Mini Microphone possesses a lithium-ion rechargeable battery, which lasts up to 8.5 hours on a full charge (charging time is approximately 3 hours). It has a reported range of 5 to 20 feet depending upon the characteristics of the environment and situation of use. Like the Phone Clip+, the Mini Microphone must be connected to the Nucleus 6 sound processor. This is accomplished by removing the silver cover from the base of the device and pressing the connecting button. Then, the Nucleus 6 sound processor is powered on and the connecting process should automatically complete. It should be noted that the Mini Microphone has a volume control to allow the user to find a comfortable listening loudness.

The Unite TV Streamer 2 (see Figure 1–45) allows for wireless streaming (ReSound's proprietary digital radio transmission at 2.4 GHz) from the accessory device directly to the Nucleus 6 sound processor. As with the aforementioned wireless accessories, the Nucleus 6 sound processor must be connected to the Unite TV Streamer 2. This is accomplished by pressing the pairing button on the TV Streamer 2 (Figure 1–46) and then powering on the Nucleus 6 sound processor. The pairing process should occur automatically. One Nucleus 6 sound processor may be connected to as many as three Unite TV Streamer 2 devices, and each Unite TV Streamer 2 may be connected to multiple sets of Nucleus 6 sound processors. The range of transmission of the unite TV Streamer 2 is up to approximately 21 feet. If the user leaves the area of transmission for less than 5 minutes, then the TV Streamer 2 will automatically reconnect to the recipient's Nucleus 6 sound processor when the user returns to the room. The TV Streamer 2 also possesses a volume control, which allows the

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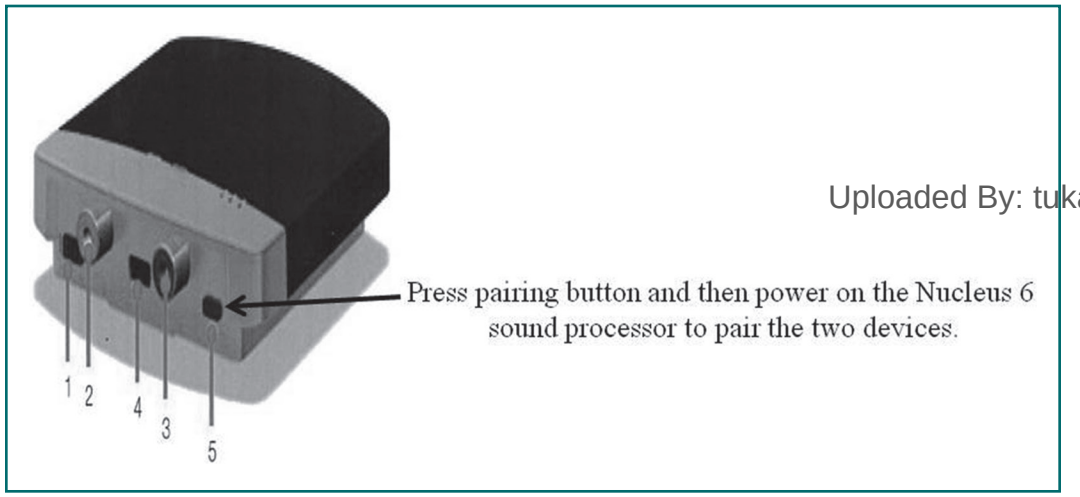



**FIGURE 1–46.** Pairing the GN ReSound Unite TV Streamer 2 to the Nucleus 6 sound processor.

user to adjust the volume of the signal he or she receives independent of the volume setting on the television. At the time of this writing, these accessories were not yet available in the United States for use with the Nucleus 6 sound processors.

## Nucleus Freedom Internal Device

The Nucleus Freedom internal device (CI24RE) (see Figure 1–25) was introduced to the market in the United States in April of 2005. It features a sophisticated digitally driven stimulator encased in titanium. The DSP of the Freedom internal device has an asynchronous architecture to enhance signal processing efficiency and effectiveness. Similar to the Advanced Bionics device, the entire internal device package is hermetically encased in biocompatible silicone. The Nucleus Freedom has an excellent record of device reliability and durability.

The device has a maximum stimulation rate of 32,000 pulses per second and a DSP featuring an asynchronous architecture. This asynchronous architecture allows, among other things, the Freedom implant to produce stimulus pulses at any point in time rather than confining stimuli to a fixed pulse grid. This capability will allow the Freedom to execute peak-derived timing signal coding strategies, which aim to stimulate at rates consistent with the temporal fluctuations in the input signal and harbor potential for improvement of speech recognition. Additionally the Freedom implant has 22 stimulating leads and 22 intracochlear stimulating platinum, half-banded electrode contacts, which are positioned toward the modiolus. Similar to the other cochlear implants on the market, the Nucleus Freedom implant has a telemetry system that facilitates two-way transmission of information from the external and internal devices. A carrier frequency of 5 MHz is used to transmit the signal in both directions. The amplifier used to record responses from the auditory nerve for diagnostic purposes (i.e., the amplifier used to record NRT) has a low noise floor (less than 1 microvolt) and, as a result, enables acquisition of robust neural responses at relatively low stimulation levels.

An internal magnet is positioned within a Silastic sleeve in the center of the internal receiver (telemetry) coil, allowing for temporary removal of the internal magnet if the patient needs an MRI. The internal stimulator is positioned just below the telemetry coil. It should be noted that the Physician's Packet included with Nucleus Freedom implants distributed in the United States indicates that an MRI may be conducted with the magnet removed so long as the strength of the MRI is 1.5 tesla or less. It is likely that the metal in the implant will cast a "shadow" on the MRI in the area around the implant (up to 2.5 inches away from the implant).

It should also be noted that different guidelines exist in other parts of the world. For instance, in some countries (e.g., Australia, New Zealand), an MRI may be completed without removing the magnet or taking any other special precautions if the strength of the MRI strength is 0.2 tesla or less. When the MRI scan is conducted at a strength of more than .02 tesla and up to 1.5 tesla, then the magnet may be left in the implant, but a compression bandage must be tightly adhered to the head over the implant site to prevent excessive implant movement during the scan. Leaving the magnet in place during the MRI scan will result in a larger shadow around the device (up to 4.5 inches). The magnet must be removed when an MRI scan of greater than 1.5 tesla and up to 3.0 tesla is conducted. MRI scans of greater than 3.0 tesla are not recommended, even when the magnet is removed.

One unique characteristic of the Nucleus Freedom internal device is a reference/return electrode (known as the "Monopolar 1" or "MP1") located at the tip of a separate lead (i.e., separate from intracochlear electrode array). This arrangement permits broader stimulation fields and a reduction in the amount of current needed for a given level of stimulation. The Nucleus Freedom also has an extracochlear return electrode located on the case of the stimulator (referred to as the "Monopolar 2" or "MP2"). A wide variety of stimulation modes are available and include the use of both return electrodes in isolation and in tandem, along with a special mode known as common ground stimulation (Chapter 2).

The Nucleus Freedom implant possesses 22 intracochlear stimulating electrode contacts (i.e., physical stimulation sites), which is a greater number than offered by competing implant manufacturers. Although it is not currently available for

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clinical use, it also offers dual electrode stimulation, which allows for fast, sequential stimulation of neighboring electrodes to elicit a maximum point of stimulation between two physical electrode contacts (with significantly improved battery life due to impedance drop).

At the time of this writing, the conventional electrode array of the Nucleus Freedom implant was the Contour Advance array (Figure 1–47) with a perimodiolar design and a "Softip," which is designed to reduce trauma to delicate cochlear structures during insertion. Insertion of the device using the Advanced Off-Stylet technique, along with a consistent method for identification of the cochleostomy site (just anterior and inferior to the round window), is recommended by Cochlear Ltd. These techniques reduce lateral wall force and increase the likelihood that the electrode array is not only inserted in the scala tympani, but also remains in the scala tympani throughout insertion. The Contour Advance electrode array, which is inserted approximately 18 to 20 mm into the cochlea, is intended to be inserted to an angular insertion depth of 450 degrees. The distance between the tip of the electrode array and the most basal electrode is 15 mm. The cross-sectional diameter of the Contour Advance (5 mm at the tip of the array and 0.8 mm at the base) most often prevents a round window insertion. Additionally, the Contour Advance electrode also possesses a marker to guide the surgeon in achieving the proper insertion depth (approximately 18 mm). An emerging line of research indicates that better speech-recognition performance is associated with complete electrode insertion in the scala tympani. The electrode contacts in the Contour Advance array are spaced in a nonlinear manner to maximize the uniformity of spectral coverage for a given number of electrodes and to promote a higher probability of good speech perception. It should be noted that less than 1 mm typically exists between electrode contacts.




**FIGURE 1–47.** Cochlear Nucleus Freedom Contour Advance Electrode Array. *Note.* Provided courtesy of Cochlear™ Ltd.

The Straight electrode array is also available with the Freedom implant (see Figure 1–25). The Straight array (an older generation electrode array) is the predecessor to the Contour Advance and is not a perimodiolar electrode. With a cross-sectional diameter of 0.4 mm at the apical end and 0.6 at the basal end, it is thinner than the Contour Advance but not quite as thin as the CI422. Furthermore, the Straight Array features 22 intracochlear, full-band electrode contacts, 2 extra-cochlear references electrodes, and 10 inactive stiffening rings that aid in the insertion of the array. Furthermore, the Straight Array is almost 17 mm, and because it is typically inserted along the lateral wall of the cochlea, the angular insertion depth approaches 360 degrees. The Straight array is recommended for patients who may have a common cavity deformity of the cochlea. Additionally, because it is more rigid, the Straight Array may be considered for recipients with fibrous tissue growth in the cochlea (i.e., bacterial meningitis), because its increased rigidity compared to the Contour Advance and CI422 arrays enhances the likelihood that it will pass through soft fibrous tissue. For similar reasons, it may be considered for those who are undergoing revision surgery and for whom a Contour Advance or CI422 cannot be inserted.

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It should be noted that Cochlear also offers the Double Array electrode array, but at the time of this writing, it is only available for the previous generation Nucleus 24 internal device (see Table 1–2). The Nucleus Double Array possesses two electrode leads to be placed in two different cochleostomies, one in the traditional basal location and another at a more apical locale. The Double Array is intended for postmeningitic recipients who have significant cochlear ossification preventing a full insertion of a conventional, single electrode array. At the time of this writing, the

Double Array may not be used with the Nucleus 6 sound processor, as the Nucleus 6 is only compatible with the Nucleus.

## CI422 Implant

Cochlear also offers the CI422 cochlear implant (commercially released in the United States in 2012). In essence, the electronics and housing of the internal device of the CI422 (see Figure 1–25) are identical to that of the Freedom implant. The differences between the Freedom and CI422 implants reside in the electrode arrays of the two devices, as well as in the surgical procedures recommending for inserting the two electrode arrays. The CI422 was developed to reduce the likelihood of physically encountering the scala media and the neural elements of the cochlea in order to facilitate a higher likelihood of hearing preservation. The CI422 possesses a "straight" (lateral wall) electrode array rather than the perimodiolar Contour Advance array in order to avoid neural elements and the organ of Corti. Furthermore, the CI422 possesses two depth markers, so that the surgeon may insert the electrode array at a depth of 25 mm (approximately 450 degrees in the typical recipient) or 20 mm (about 360 degrees in the typical recipient). The shallower insertion depth is intended for recipients who have better preoperative low-frequency hearing in order to preserve the likelihood of hearing preservation after implant surgery. Also, the diameter of the CI422 electrode array is thinner than the Freedom Contour Advance. Specifically, the cross-sectional diameter of the CI422 electrode array is 0.3 mm at the tip of the array and .6 mm at the base of the array.

## Nucleus Hybrid Cochlear Implant

In November of 2013, Cochlear received FDA approval to commercially market the Cochlear Nucleus (L24) Hybrid cochlear implant in the United States (Figure 1–48). The Nucleus Hybrid cochlear implant was designed with the primary intent of preserving low-frequency hearing and is intended for use with recipients who have signifi-

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**FIGURE 1–48.** Cochlear Nucleus Hybrid cochlear implant. *Note.* Provided courtesy of Cochlear™ Ltd.

cant low-frequency hearing preoperatively (better than 60 dB HL at 500 Hz). The Nucleus Hybrid implant possesses a similar electronics and housing package as the Nucleus Freedom implant. The primary differences between the Nucleus and the Freedom cochlear implant lie in the electrode lead, the electrode array, and the recommended surgical procedures to insert the electrode arrays of the two devices. The Nucleus Hybrid electrode lead possesses a relatively large fantail, which is intended to facilitate ease of handling while the surgeon inserts to the electrode array into the cochlea. Additionally, the section of the electrode lead adjacent to the fantail is rather stiff, which also enhances the ease of handling during the insertion process. Further, the electrode lead possesses a "stopper," which serves a clear guide as to how deeply the electrode array should be inserted. Also, the electrode array itself is designed to be inserted to a depth of approximately 230 degrees (about 16–17 mm), and the cross-sectional diameter of the Nucleus Hybrid electrode array is 0.25 by 0.35 mm at the apical end of the array and 0.4 by 0.55 mm at the basal end of the array. The array is often thin enough to be inserted into the round window or into a cochleostomy (the choice is typically dependent on surgeon's preference and recipient anatomy). Finally, the Nucleus Hybrid electrode array is intended to be inserted with the use of a "soft" surgical insertion procedure, which is designed to protect the cochlea from injury during insertion. Additional information pertaining to soft surgical techniques is provided in Chapter 10 of this book.




**FIGURE 1–49.** Cochlear Nucleus CI512 cochlear implant. *Note.* Provided courtesy of Cochlear™ Ltd.

## CI512 Implant

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Cochlear commercially introduced the CI512 cochlear implant (Figure 1–49) in September of 2009. At the time of its commercial release, the CI512 was the thinnest implant on the market. For the most part, the CI512 had the same electronics package as the Freedom. The CI512 was voluntarily withdrawn from the market in September of 2011 after it was discovered that a relatively large number of the devices were failing to function after a few months of use by recipients. An internal review by Cochlear Ltd. identified a fault in the brazing process (i.e., the process by which the titanium case is adhered to the ceramic feedthru components) compromised hermiticity allowing moisture into the electronics of the implant. A reliability report published in May of 2014 indicated that 7.1% of CI512 implants had failed within 5 years of use. In comparison, only .7% of Freedom implants failed after 5 years of use. At the time of this writing, the CI512 implant had not returned to the market. It should be noted that the thin casing of the device made it quite attractive for young children and other recipients with thin skull densities, so it is likely that a similarly thin package will be reintroduced to the market in the future. Additionally, the CI512 was only manufactured with the Contour Advance electrode array.

# MED-EL CORPORATION

The MED-EL Corporation manufactures the MAESTRO Cochlear Implant system, which currently offers two external sound processor options

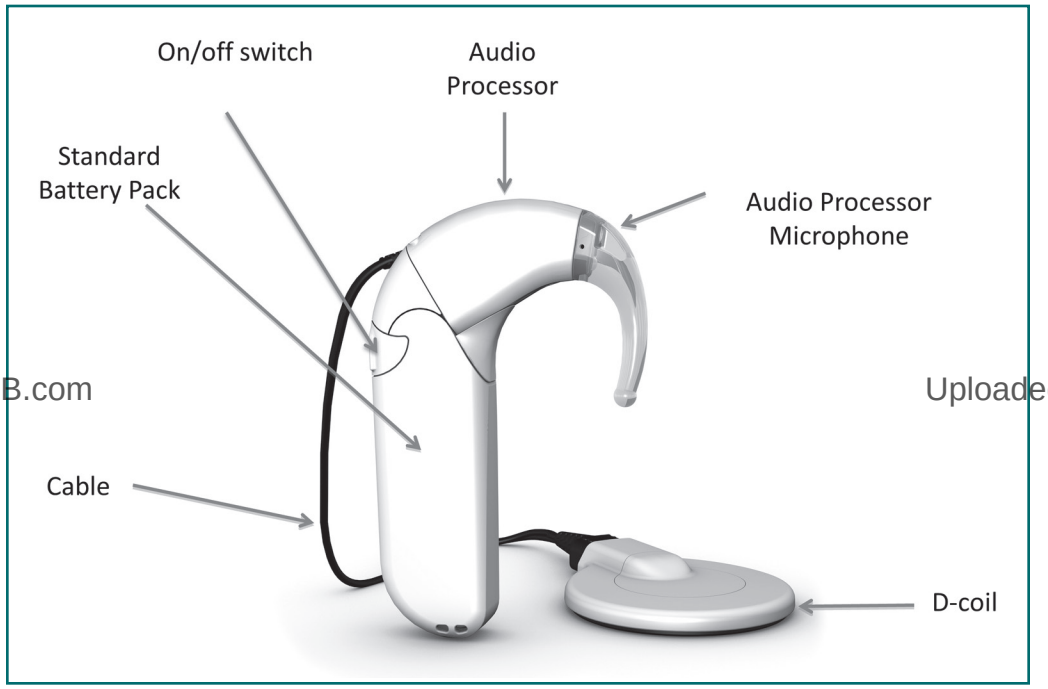
for recipients: the OPUS 2 and the RONDO™, and several internal devices.

## OPUS 2 Sound Processor

The OPUS 2 external sound processor, introduced in 2008, is typically used as a BTE sound processor (Figure 1–50). It is a small, lightweight processor that has several enhanced features when compared to the previous generation processors (re. TEMPO+ and OPUS 1). The OPUS 2 has a fully digital sound processor that uses a Hilbert transformation to analyze incoming acoustic inputs. The input frequency range of the sound processor extends from 70 to 8500 Hz, and the sound processor can accommodate up to four user programs. It has an omnidirectional microphone located at the top of the processor just behind the earhook. It should be noted that the TEMPO+ and OPUS 1 sound processors also feature the same input frequency range as the OPUS 2, and they also utilize a Hilber transformation in the analysis of the input signal.

Additionally, an LED light can be used to alert the user or caregiver of the processor status, and there is a manually accessible telecoil. The earhook of the OPUS 2 sound processor is attached to the apex of the processor by a small pin. This purpose of this pin is to prevent young children from removing and ingesting the earhook. The earhook is available in multiple sizes to accommodate different ear sizes. A unique feature of the OPUS 2 sound processor is that it does not include any user controls (other than an on/off switch). Instead, the user may adjust processor settings via a remote control, which will be discussed later.

MED-EL offers a full portfolio of wearing options for the OPUS 2 (which is discussed below) to facilitate excellent retention and comfort for recipients with varying ear anatomy, developmental needs, and communication lifestyles. In the ear-level configuration, the OPUS 2 sound processor (see Figure 1–50) is composed of three primary components: (1) the sound processor (i.e., control unit), (2) the angled battery pack, and (3) the cable and coil. Another excellent feature of the OPUS 2 sound processor is that it may be used




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**FIGURE 1–50.** MED-EL OPUS 2 external sound processor. *Note.* Image courtesy of MED-EL Corporation.

with every MED-EL cochlear implant that has ever been introduced to the commercial market (e.g., CONCERT, SONATA TI$^{100}$, PULSAR CI$^{100}$, COMBI 40+).

The coil was updated in 2011 with the release of the thinner and lower-profile D-coil that is engineered to be 50% more energy efficient than its predecessor. The D-coil features a removable cable, which comes in several lengths to allow for variation in head size and wearing configurations (battery estimates with D-coil below). The magnet strength of the D-coil may be adjusted by the programming clinician to accommodate a range of adhesive needs across recipients. This is accomplished by removing a twisting and lifting the magnet out of the D-coil and replacing it with a magnet of the desired strength. Magnet strengths are available in soft (one dot on the side of the magnet), standard (two dots), strong (three dots), and super strong (four dots). The radio frequency used to transmit the signal from the D-coil to a MED-EL cochlear implant is 12 MHz.

### *Battery/Wearing Options*

MED-EL currently offers two battery options for use with the OPUS 2 sound processor when worn on the ear. For each ear-level option, a "connecting piece" is required to couple the battery pack to the processor.

The Standard Battery Pack (Figure 1–51) may be used with one of the DaCapo rechargeable batteries or three disposable zinc-air 675 batteries. The DaCapo rechargeable battery is 20% lighter than three zinc-air 675 batteries, and MED-EL reports that up to 16 hours of use time may be achieved with the Standard Battery Pack and the DaCapo rechargeable battery. Additionally, up to 90 hours of battery life may be achieved with the use of the Standard Battery Pack with three zinc-air 675 batteries. As is discussed in Chapters 3 and 6, battery life varies across recipients and is primarily influenced by skin flap thickness.

In addition to the standard ear-level battery pack, MED-EL also offers the OPUS 2 XS Battery Pack, which was released in 2012 (see Figure 1–51). The OPUS 2 XS Battery Pack uses two disposable zinc-air 675 batteries and is smaller and lighter than the Standard Battery Pack. In fact, when the OPUS 2 processor is used with the XS Battery Pack, it is one of the smallest ear-level sound processors that is commercially available in the cochlear implant industry at the time of this writing. The XS option may be particularly attractive for young children as it reduces processor size and weight while allowing for BTE use. This facilitates a more favorable microphone location than what would be obtained with body-worn options, such as the BabyBTE/ActiveWear configurations mentioned below. MED-EL reports that up to

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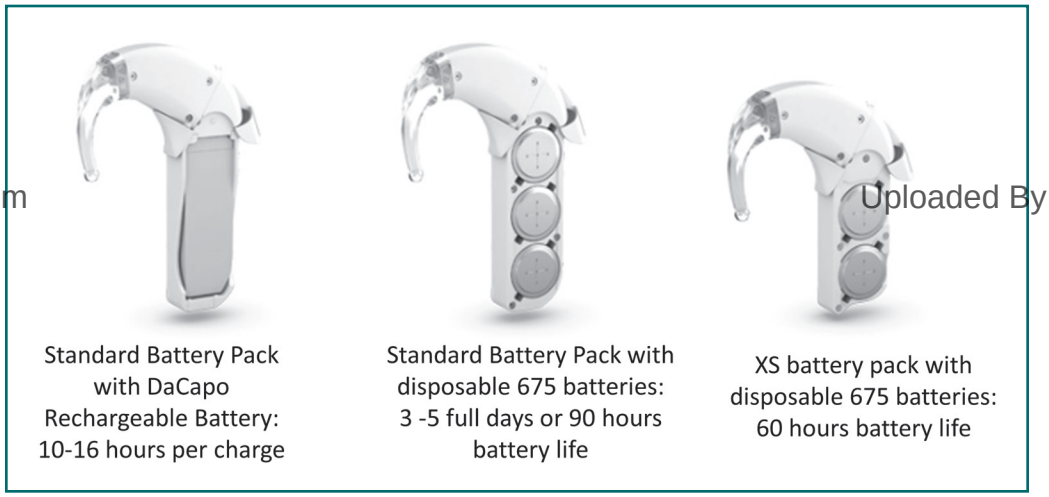



**FIGURE 1–51.** Comparison of MED-EL OPUS 2 battery options and battery life. *Note.* Images courtesy of MED-EL Corporation.

60 hours of battery life may be obtained with the XS Battery Pack (see Figure 1–51) for a comparison of size between the standard rechargeable and nonrechargeable options and the XS Battery Pack).

Additional options are available to cater to unique patients' needs. As shown in Figure 1–52, a pediatric configuration (i.e., BabyBTE™ or ActiveWear) with a straight battery pack, a fixation piece, and a long cable attaching the coil to the processor allows the OPUS 2 system to be clipped to a user's clothing without anything on the ear. This option may be particularly attractive for very young children or people who play sports; however, care should be taken to position the processor microphone appropriately (i.e., toward typical sound source at 0 degrees azimuth). However, this option is not ideal for bilaterally implanted children, where the intra-aural level differences need to be accurately maintained to help with localization, speech recognition in noise, and other bilateral processes that require preservation of inter-aural cues.

MED-EL also offers a Children's Battery Pack (see Figure 1–52), which possesses a long cable connecting the battery to the sound processor. This configuration allows young children to wear the processor on the ear and the battery on the body. The Children's Battery Pack is similar in size to the Standard ear-level Battery Pack and uses three disposable zinc-air 675 batteries. MED-EL reports that the Children's Battery Pack provides up to 90 hours of use with three zinc-air 675 batteries. This is an attractive option for young, bilaterally implanted children because interaural cues will be preserved more effectively than if the sound processor microphone was worn on the body. It should be noted, however, that the Children's Battery Pack does not have DAI port for use with FM/RF systems. MED-EL suggests that the recipient should use the OPUS 2 telecoil with an induction neckloop when use of a personal digital RF/FM system is desired.

MED-EL also offers the Mini Battery Pack option (see Figure 1–52). Like the Children Battery Pack, the Mini Battery Pack uses a long cable that connects the sound processor to the battery unit. Again, this configuration allows the sound processor microphone to be positioned at the ear, whereas the battery, which possesses a great deal of the sound processor's bulk, may be clipped to the body. The Mini Battery Pack may be used with the DaCapo rechargeable battery (up to 16 hours of use) or with an alkaline or rechargeable AAA battery. The Mini Battery Pack does include a three-prong EuroPort to allow DAI.

The rechargeable batteries are charged in the DaCapo charger, which was specifically developed by MED-EL to charge DaCapo rechargeable batteries (Figure 1–53). MED-EL recommends that the DaCapo charger be plugged into a wall outlet, not a surge protector/strip. The DaCapo charger possesses an array of LED lights that indicate

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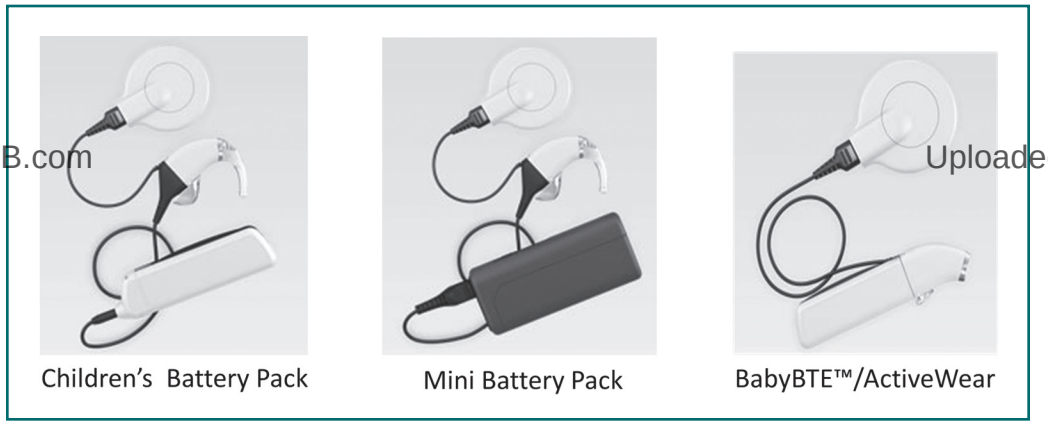



**FIGURE 1–52.** MED-EL OPUS 2 Pediatric Wearing options: Children's Battery Pack, Mini Battery Pack, and BabyBTE™/ActiveWear. *Note.* Images courtesy of MED-EL Corporation.

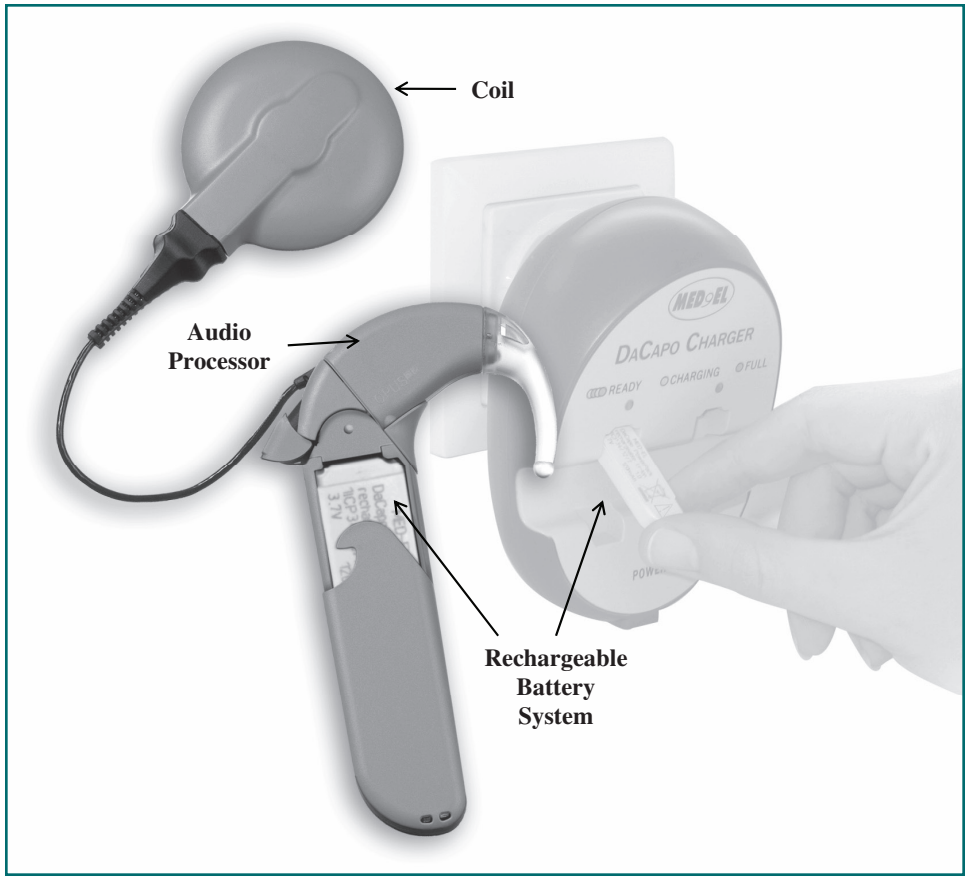



**FIGURE 1–53.** MED-EL OPUS 2 and DeCapo battery charger. *Note.* Image courtesy of MED-EL Corporation.

the status of the charging cycle (e.g., "charging," "full"). Four hours is required to charge a dead battery to a full charge. The DaCapo rechargeable battery has a nominal voltage capacity of 3.7 volts and may be charged over 500 times before replacement is necessary.

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## RONDO

The RONDO™, which was introduced in 2013, is the first-ever, single-unit external sound processor (Figure 1–54). The control unit, battery pack, and transmitting coil are integrated into one device. The RONDO™ was not designed to replace the OPUS 2 sound processor. It is an option that allows the recipient to wear the entire device on the head, but off of the ear. It was built on the same electronics/signal processing platform as the OPUS 2 (discussed below) and includes Automatic Sound Management and FineHearing Technology. In short, the RONDO™ can provide the same types of signal coding strategies and sound processing as the recipient uses with the OPUS 2 processor.

The RONDO™ has an omnidirectional microphone, and MED-EL recommends that the processor is placed so that the microphone is pointing straight up or straight down. That same placement is also recommended to promote optimal use of the telecoil. The RONDO™ does not contain external controls, such as a program switch or volume control. Instead, the recipient may use the FineTuner remote control to adjust the settings on the processor. Connectivity with accessories can be completed via the integrated telecoil or via direct connect (DAI) with the Mini Battery Pack and connection cable. For children, the authors do not recommend the RONDO be used during

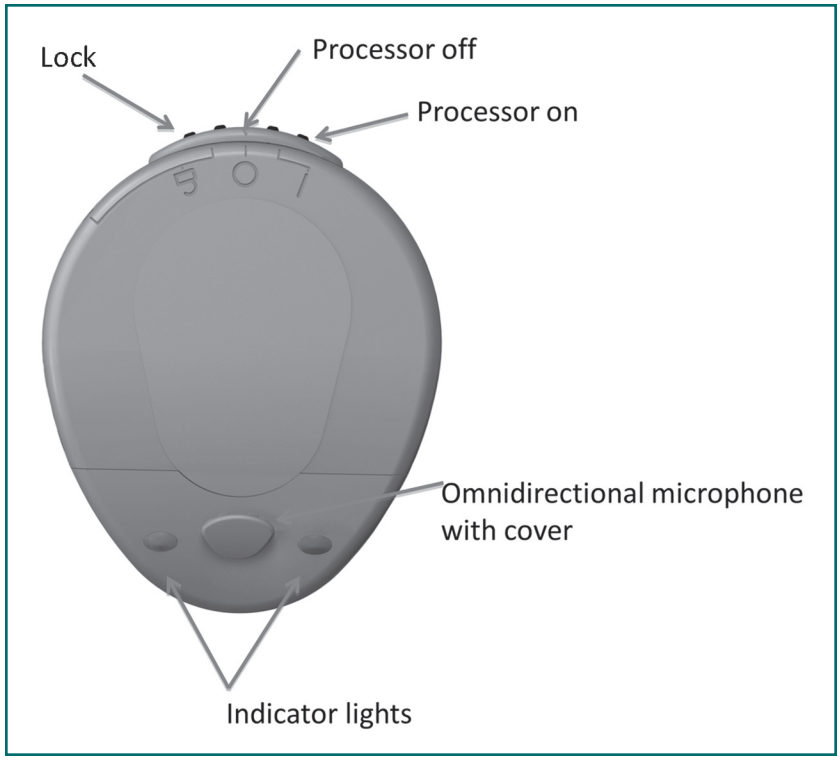



**FIGURE 1–54.** MED-EL RONDO sound processor. *Note.* I = activate, O = deactivate (plus lock button). Must unlock to change batteries or get to magnet. *Note.* Image courtesy of MED-EL Corporation.

sports or heavy activity due to the possibility of the device falling off during strenuous activity. However, the company recently announced the release of a Sports Headband that has a pocket for the RONDO to improve device retention when patients are active (Figure 1–55). Increasing the magnet strength may help with retention, but may be uncomfortable or cause complications to the skin flap if used for extended periods of time. The programming clinician may adjust magnet strength by removing a cap on the RONDO processor and inserting a magnet with one of four different strengths. The RONDO also includes a protection sleeve and a place to which a retention cord may be attached to prevent the processor from being lost if it falls from the head.



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**FIGURE 1–55.** MED-EL RONDO Sports Headband. *Note.* Image courtesy of MED-EL Corporation.

### *Battery Options*

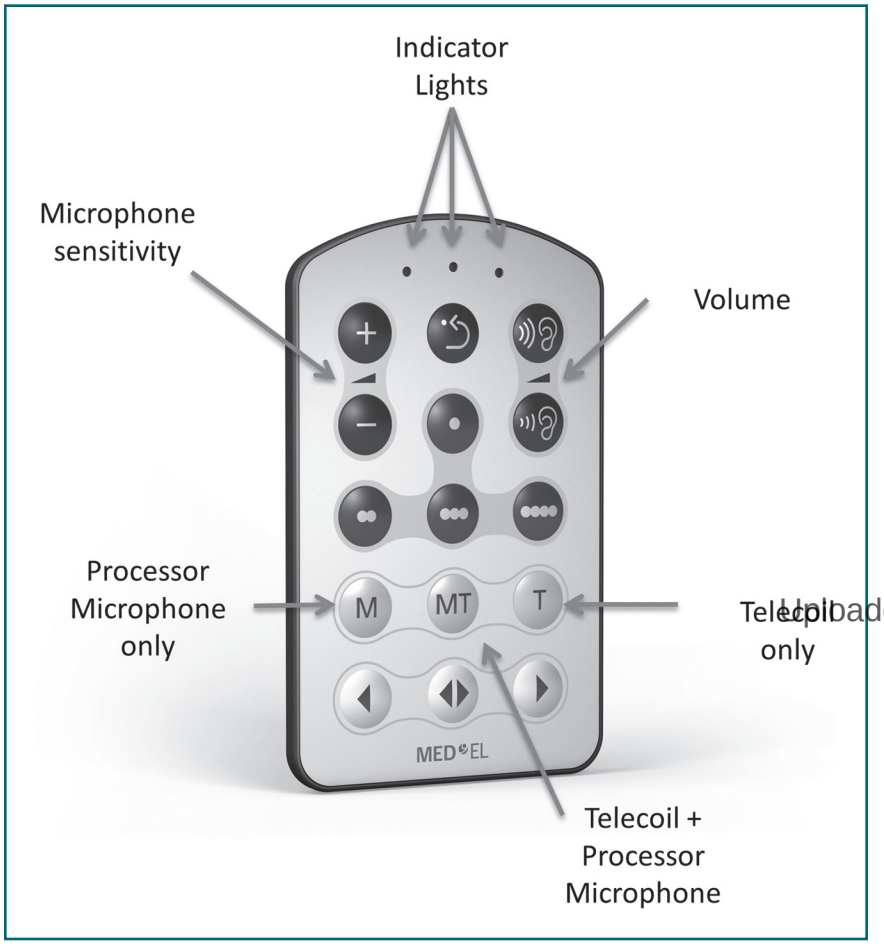
The RONDO uses three #675 batteries that will provide up to five days or 90 hours of use. Currently, there is not a rechargeable battery option or DAI/FM receiver option unless the device is configured with an off-the-processor battery pack (mini battery pack) that uses one AAA alkaline battery or a DaCapo Rechargeable battery.

### *FineTuner*

As previously discussed, both the RONDO and OPUS 2 sound processors are unique because they do not contain user controls on the body of the processor, with the exception of a power (on/off) switch. Instead, the user may adjust the processor settings with a FineTuner remote control, which was the first remote control to be introduced in the cochlear implant industry. The FineTuner (Figure 1–56) allows the recipient to adjust the volume, change the microphone sensitivity, and select up to four user programs from up to approximately 2.5 feet away. The FineTuner uses near-field magnetic induction at a carrier frequency of approximately 9000 Hz to transmit information from the remote to the processor. Furthermore, the recipient may use the FineTuner to activate the integrated telecoil in a telecoil-only mode (T) or a telecoil-plus processor-microphone mode (MT). The FineTuner can be used to make unilateral or bilateral adjustments to the processor.

It is quite simple to synchronize the FineTuner to one both of a recipient's processors. The clinician or recipient must simply turn off the RONDO or OPUS 2 sound processor, place the OPUS 2 D-coil/RONDO over the MT key of the FineTuner, and then turn the processor on. Two amber indicator lights will blink briefly to indicate successful synchronization.

When using the OPUS 2, the FineTuner only needs to be synced for the first fit, when using a new processor, or when receiving a replacement FineTuner. When using the RONDO processor, the remote needs to be resynced any time a new configuration is programmed into the device. For




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**FIGURE 1–56.** MED-EL FineTuner remote control. *Note.* Image courtesy of MED-EL Corporation.

purposes of ease of use and safety, the buttons of the FineTuner can also be locked manually or programmed (within the programming software) to disable certain features/settings.

### *Additional Features in MED-EL Systems*

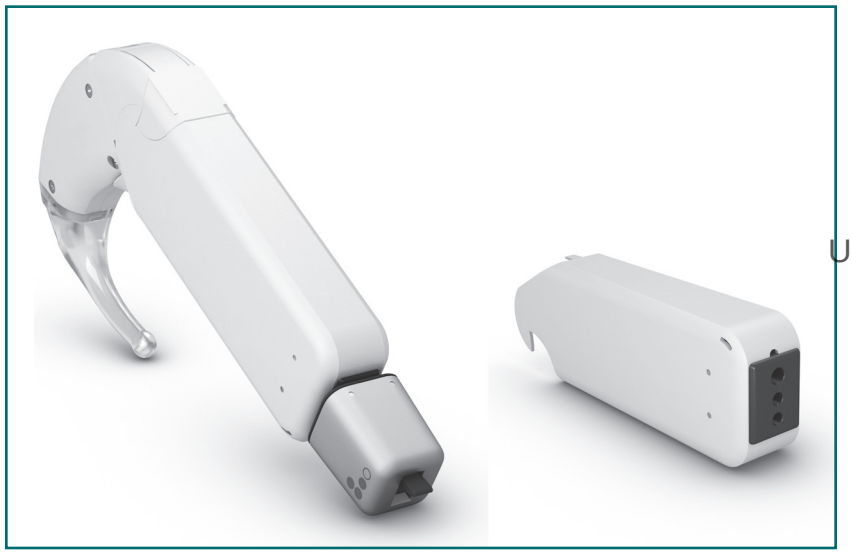
There are several additional features in MED-EL external sound processors, such as private alerts, SoundGuard, and IRIS™. Private alerts may be enabled by the audiologist to signal that setting changes have been made or to alert the patient that the battery is low. SoundGuard monitors against electrostatic discharge (ESD) and when detected, will discontinue stimulation and the LED light will flash. The processor can be rebooted and the device will be restored to its settings, usually with no need for reprogramming. The IRIS™, which stands for Individual Recognition of Implant System, ensures that only the correct ear is stimulated.

Another novel aspect to this processor is the OPUS 2 FM Battery Pack Cover with a three-pin Europlug port (Figure 1–57) When this battery cover is in use, the Europort plug is located at the base of the processor. It enables wireless coupling of a personal digital RF/FM receiver (e.g., Phonak Roger X, Phonak MLxi, Oticon Amigo R2) to the processor. Additionally, the recipient may connect a specialized audio cable to the Europlug port to allow connection to personal audio devices, such as MP3 players, laptop computers, handheld video game consoles, mobile telephones, and so forth (Figure 1–58). Two different specialized audio cables are available: a yellow cable is used



**FIGURE 1–58.** Specialized audio cable to the Europlug port to allow connection to personal audio device. *Note.* Image courtesy of MED-EL Corporation.

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**FIGURE 1–57.** OPUS 2 FM Battery Pack Cover with a three-pin Europlug port and universal RF receiver. *Note.* Image courtesy of MED-EL Corporation.

to obtain a 50/50 mixing ratio (50% input from processor microphone and 50% input from auxiliary device for use with an MP3 player or phone), and a red cable is used when the recipient chooses to primarily focus on the auxiliary input (90/10 mixing ratio with 90% input from the auxiliary device and 10% from the sound processor microphone). A bilateral audio Y-cable is also available to allow bilateral recipients to connect personal audio devices to both of their implant processors.

An additional hearing assistance technology option is use of an induction neckloop via the T-coil. The recipient may use the FineTuner to select between telecoil only (T) use or a 50/50 mix of the telecoil and microphone (MT). MED-EL offers an interactive assistive listening device database on its website.

## CONCERT Internal Device

The CONCERT internal device (referred to as CONCERTO in countries outside the US, but is referred to here as the CONCERT) was released in 2011 and is MED-EL's newest internal device (Figure 1–59). At the time of this writing, it is the industry's smallest and thinnest internal device. The electronics package is identical to that of its predecessor, the MED-EL SONATA TI$^{100}$, but the casing of the CONCERT is 25% thinner than the casing of the SONATA TI$^{100}$. Like the SONATA TI$^{100}$, the case of the CONCERT is comprised of titanium with a silicone covering. In 2013, MED-EL began offering the option of a CONCERT internal device with a PIN positioned on the back of the CONCERT case. This PIN is designed to be inserted by the surgeon onto the recipient's skull to facilitate better retention and to prevent lateral/rotation at the point at which the CONCERT case contacts the skull. The CONCERT PIN is optional and selected based on the preference of the surgeon.

## SONATA TI$^{100}$ Internal Device

The SONATA TI$^{100}$ was the first MED-EL internal device encased in titanium with a silicone covering over the receiving coil and case (Figure 1–60). The

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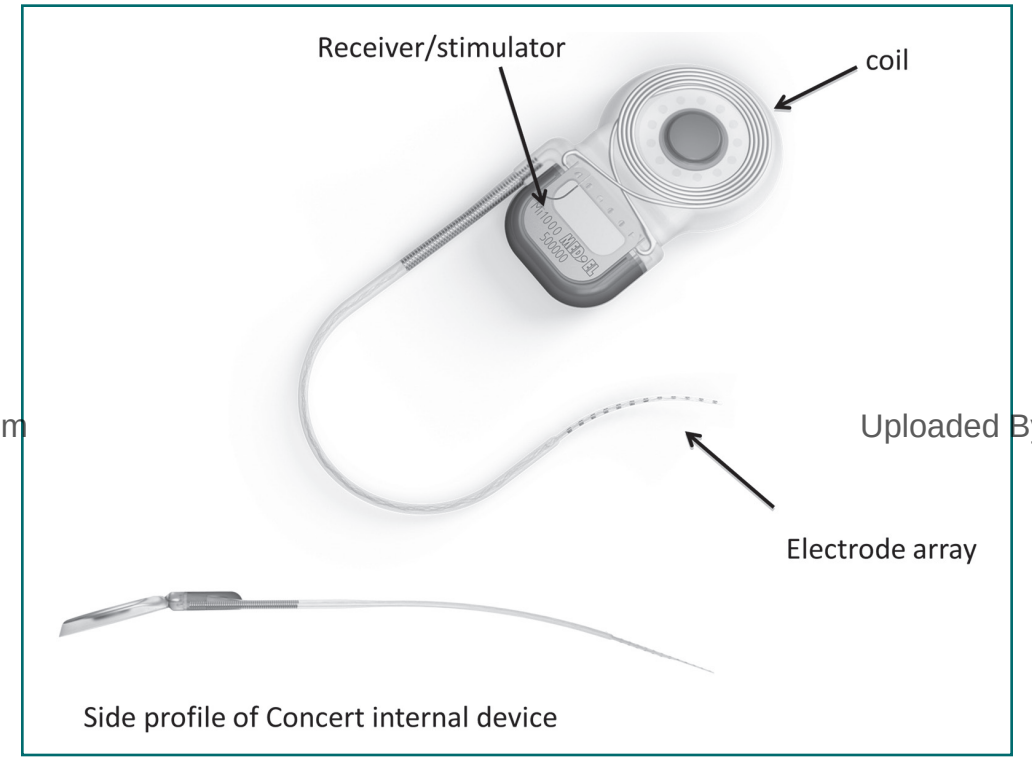



**FIGURE 1–59.** MED-EL CONCERT internal device. *Note.* Image courtesy of MED-EL Corporation.

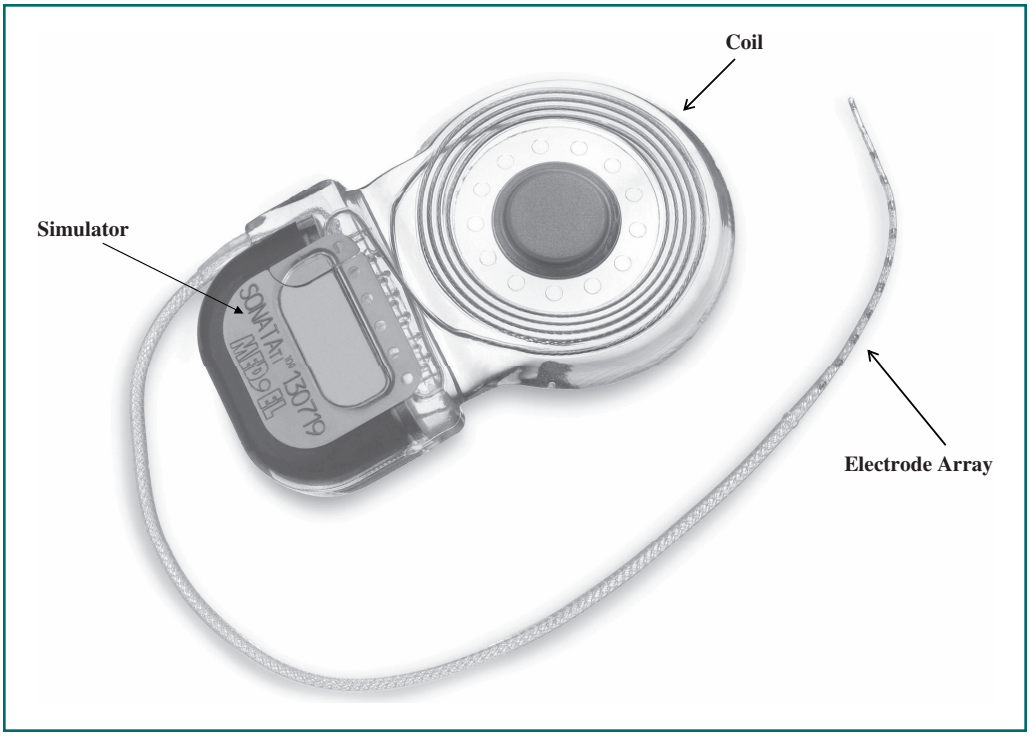



**FIGURE 1–60.** MED-EL SONATA TI$^{100}$ internal device. *Note.* Image courtesy of MED-EL.

electronics package of the SONATA TI$^{100}$ is identical to that of the CONCERT. The only difference between the two implants is that the CONCERT is thinner/smaller than the SONATA. Previous internal devices were encased in a ceramic shell. All three internal devices: CONCERT, SONATA, and PULSAR were recently FDA approved for 1.5T (tesla) MRI scans without surgical removal of the internal magnet (MED-EL, 2013). Health care providers should follow MED-EL's recommendations for MRI scanning in order to ensure recipient safety and to uphold the manufacturer's warranty. These recommendations are available on MED-EL's website.

## PULSAR CI$^{100}$ Internal Device

Additionally, MED-EL continues to offer the PULSAR CI$^{100}$, which is the CI industry's only implant with a ceramic casing. The PULSAR CI$^{100}$ is slightly smaller than the SONATA TI$^{100}$. Although the titanium casing may improve durability and reduce the chance of failure due to impact trauma, the primary reason MED-EL introduced the titanium casing was to answer surgeons' requests. It should be noted that the MED-EL ceramic case of the PULSAR CI$^{100}$ has a relatively low rate of failure and case fracture.

### *Digital Signal Processing*

The electronics of the PULSAR CI$^{100}$ are identical to the electronics in the CONCERT and SONATA TI$^{100}$. The CONCERT features a sophisticated DSP with a powerful microchip and several electrode array options. For all MED-EL cochlear implants, a 12 MHz carrier frequency is used to transmit the signal to the internal device.

The DSP of the CONCERT (as well as for other MED-EL implants) allows for a total maximum stimulation rate of 50,704 pulses per second. It has independent output circuits for each channel, enabling precise delivery of stimulation and providing the capability for simultaneous activation of channels. In addition, each chan-

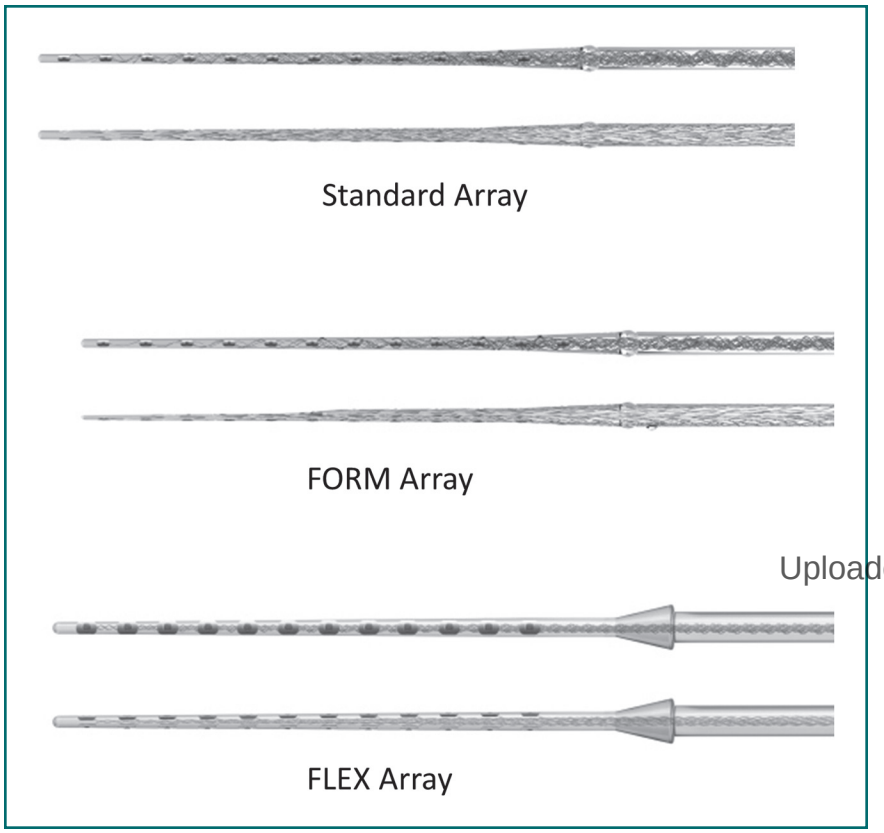
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nel has independent output capacitors for safety purposes. Twenty-four oval electrode contacts are arranged in pairs to form 12 stimulating sites. The paired configuration allows for a broader stimulation site and allows the same electrode array to be used for either the right or left ear. A primary return (or reference) electrode and a secondary reference electrode (i.e., ECAP/ART measures) are located on the titanium casing. (For the PULSAR CI$^{100}$, the reference electrode is a remote lead positioned under the temporalis muscle). The ECAP reference electrode is used for recording the electrically evoked compound action potential via Auditory nerve Response Telemetry (ART).

## MED-EL Electrode Array Options

MED-EL offers the largest selection of electrode options within the implant industry, some of which are shown in Figure 1–61. These electrode options are designed to accommodate different cochlear duct lengths and to provide complete cochlear coverage for all recipients. Traditional options include the Standard, Medium, Compressed, and Split electrodes. The newest options include the FLEX 24 and 28, released in 2012, as well as the FLEX SOFT, which is not commercially available in the United States at the time of publication but is available in most countries. All FLEX options are discussed below.

The unique and potentially beneficial aspect of the Standard MED-EL electrode array is that it is designed to be inserted much deeper than electrode arrays developed by other manufacturers. For example, the Standard Array has a 31.5-mm array length from the apical tip to the seal placed in the cochleostomy and is intended to be inserted to an angular insertion depth of about 630 degrees. Conventional electrode arrays offered by other




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**FIGURE 1–61.** Three MED-EL electrode arrays. *Note.* Image courtesy of MED-EL.

manufacturers typically are inserted to an angular insertion depth of 450 to 540 degrees, which corresponds to a length of about 18 mm (in the case of perimodiolar electrodes) to 25 mm (in the case of lateral wall electrodes). For electrode arrays inserted to an angular insertion depth of 450 to 540 degrees, low-frequency inputs (150 to 500 Hz) are likely delivered to a place in the cochlea that is innervated by nerve fibers with characteristic frequencies that are several hundred Hz higher than the frequency of the original sound. Deeper insertion of the electrode array (i.e., 31.5-mm/630-degree angular insertion depth) may provide more favorable access to low-frequency sounds because low-frequency stimulation will be provided to auditory nerve fibers that have characteristic frequencies corresponding to those of the input sounds. The longer length and deeper insertion of the Standard Array is one of the components of MED-EL's attempt to provide complete cochlear coverage along with fine structural cues.

The Standard Array is a straight, relatively flexible electrode array that was designed to be inserted along the lateral wall of the cochlea. The flexible nature of the electrode array and the lateral wall insertion is designed to facilitate hearing preservation and a reduction in trauma to the delicate structures in the scala media. The cross-sectional diameter of the Standard Array is relatively narrow (0.5 mm at the tip and 1.3 mm where the electrode is designed to rest in the round window or cochleostomy). The narrow diameter allows for a round window insertion. In fact, MED-EL promotes the use of a round window insertion to facilitate delivery into the scala tympani and to reduce cochlear trauma. Other characteristics of the Standard Array include 24 oval-shaped platinum electrode contacts that are arranged in pairs to form 12 stimulating sites (see Figure 1–61). As stated previously, the Standard Array is 31.5 mm in length from the apical tip of the electrode array to the seal, which resides at the round window or cochleostomy; however, the 12 stimulating sites are arranged 26.4 mm across the array (beginning approximately 1 mm from the tip). This allows for 2.4 mm of spacing between each stimulating site, which is greater than what exists for other electrode arrays in the implant industry. As a result, it is quite possible that less channel interaction is present for the Standard Array relative to other implant electrode arrays.

MED-EL has also developed a new line of electrode array technology referred to as FLEX electrode technology. The unique characteristics of FLEX electrode technology are as follows:

1. The seven most basal stimulating sites each consist of pairs of oval-shaped electrodes, which is similar to what is found in the Standard Array. However, the five most apical stimulating sites consist of just one electrode contact, which reduces the cross-sectional diameter and enhances the flaccidity of the apical end of the array.
2. The diameter of the FLEX Array is narrow at the apical tip of the array and increases at the basal end where the electrode array resides in the round window or cochleostomy. This tapered design, which is shown in Figure 1–61, allows for increased mechanical flexibility at the tip of the array, insertion into the narrow lumen of the most apical region of the scala tympani, and ease of handing at the basal end on the part of the surgeon.
3. The wires in the FLEX series of MED-EL electrode arrays are configured in a wave-shaped pattern, which further enhances flexibility and promotes atraumaticity.

Again, at the time of this writing, the FLEX SOFT Array was available for use in Europe but not in the United States.

Also at the time of this writing, MED-EL offers two arrays with the FLEX technology in the United States. The FLEX 24 and 28 electrode arrays each possess the same FLEX characteristics as discussed in the FLEX SOFT Array: (1) seven most basal stimulating sites consist of pairs of oval electrode contacts, whereas the five most apical stimulating sites are composed of single electrode contacts, (2) the diameter of the apical tip is tapered to promote mechanical flexibility, and (3) the platinum–iridium electrode leads (i.e., wires) are arranged in a

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wave-shaped pattern to promote flexibility/atraumaticity.

The FLEX 28 specifications are as follows: the diameter at apical end is 0.5 mm × 0.4 mm, the diameter at the basal end is 0.8 mm, and electrode contacts are spaced 2.1 mm apart with an active stimulation range of 23.1 mm. The FLEX 28 may be considered for recipients whose cochlear duct is not long enough to accommodate the FLEX SOFT or Standard arrays. It may also be beneficial for recipients who have substantial residual hearing at 500 Hz and below with severe to profound hearing loss at higher frequencies. In this case, the FLEX 28 may allow for a better opportunity for hearing preservation in the low frequencies.

For the FLEX 24, the diameter at the apical end is 0.5 mm × 0.33 mm, the diameter at the basal end is 0.8 mm (same as FLEX 28), and electrode contacts are spaced 1.9 mm apart with an active stimulation range of 20.9. The FLEX 24 was formerly known as the FLEX EAS array. The FLEX 24 is generally the electrode of choice for recipients for whom hearing preservation at 1000 Hz and below is an important priority. It should be noted that the FLEX 28 and FLEX 24 options are available only in the CONCERT implant, in the United States. Additional discussion of electroacoustic stimulation (i.e., hybrid cochlear implants) is provided in Chapter 9.

Although not commercially available in the United States, MED-EL offers the FLEX SOFT electrode array in most other parts of the world. The FLEX SOFT is a novel alternative to the Standard Array and is intended for cases in which a deep insertion is required but hearing and cochlear preservation are also desired. The FLEX SOFT is essentially the same length as the Standard Array (31.3 mm from the apical tip to the basal location at which the electrode array resides at the round window or cochleostomy), but as the name implies, the FLEX SOFT Array possesses FLEX electrode technology, which was designed to reduce cochlear trauma and enhance hearing preservation. The diameter of the FLEX Array is narrow (0.36 × 0.48 mm) at the apical tip of the array and increases to 1.3 mm at the basal end. The narrower tip of the apical end of the FLEX SOFT array (relative to the Standard Array) facilitates an atraumatic, deep insertion.

MED-EL's three special array options include the Medium Electrode Array (HM), the Split Electrode Array (HGB), and the Compressed Electrode Array (HS). They are primarily considered for recipients with atypical cochlear anatomy. The Medium Array's electrode contacts span 20.9 mm across the array with a 1.9 mm space between each stimulating site. This array may be used when a deep insertion is not possible (e.g., shorter cochlear duct length, abnormal cochlear anatomy, such as a Mondini's dysplasia, fibrous tissue growth secondary to bacterial meningitis, or reimplantation after previous use of another brand). Furthermore, the Compressed Array electrode contacts are placed 12.1 mm across the array with 1.1 mm spacing between sites. This array is intended for recipients who have ossification or fibrous tissue growth in the cochlea.

Finally, the Split Array is used for cases in which significant ossification exists in the cochlea and is only available as coupled to the PULSAR internal device. The 12 contact pairs are arranged on two separate electrode arrays that use the same spacing as the compressed array; seven channels are on one array and five channels are on the other. When using this array, the surgeon must make two cochleostomy sites to insert one array into the upper-basal turn of the cochlea and the other into the lower-basal region. To be precise, the two cochleostomy sites are essentially two straight "tunnel drill-outs" that the surgeon makes in an ossified cochlea. With the Split Array, the audiologist must reorder the tonotopic arrangement of the bandpass filters at the initial programming, depending on which of the two electrode arrays was placed in the lower basal turn and which was place in the upper turn of the cochlea.

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## MED-EL SYNCHRONY Cochlear Implant System

Coincident with the time this book was headed to press, MED-EL announced the release of the SYNCHRONY cochlear implant system. At the time




**FIGURE 1–62.** MED-EL. SYNCHRONY Cochlear Implant System. (At the time of this writing, this device has not been approved by the FDA.)

of this writing, this device has not been approved by the FDA. The MED-EL SYNCHRONY system, which is shown in Figure 1–62, is composed of the SYNCHRONY cochlear implant and the SONNET BTE sound processor. The SYNCHRONY cochlear implant possesses the same electronic capability and packaging as its predecessor, the CONCERT, but the former is approved in Europe for MRI up to a 3.0 tesla. Additionally, the SYNCHRONY features a removable magnet that may be temporarily taken out by a cochlear implant surgeon in the event that an MRI is needed for the head.

The SONNET sound processor is similar in appearance and shape to the MED-EL OPUS 2 sound processor, but the former has been completely redesigned to allow for greater ease of use, better aesthetics, and improved reliability. The case of the SONNET contains fewer seams than the OPUS 2, and as a result it is water resistant with a 54 IP rating. Also, the SONNET does not contain any control switches (power button, program switch, etc.), and it does not contain the connector pins that were used to couple the battery case to the OPUS 2 processor. Instead, a battery cover may be removed from the base of the processor to expose a battery frame that houses two 675 zinc-air batteries. At the time of this writing, a rechargeable battery was unavailable. The SONNET processor is powered on and off by connecting and disconnecting the battery cover to or from the processor, respectively. The battery frame may also be removed from the SONNET by pushing a level at the base of the SONNET processor. This configuration allows the SONNET processor to be modular in design. According to the manufacturer, the SONNET allows for up to 60 hours of use time with two 675 zinc-air battery. It should be noted that the conventional battery case may be replaced with an FM battery case to allow the recipient to connect to an FM/RF receiver or any other device that uses a three-pin Euro connection.

The SONNET sound processor also allows for use of Automatic Sound Management 2.0 (ASM 2.0) sound processing, which includes several new processing features. For example, the SONNET two features two microphones to provide advanced directional processing. The clinician may select from four different microphone modes: (1) omnidirectional; (2) adaptive directional, which adaptively positions to null of the microphone response to coincide with the direction from which the most intense noise is arriving; (3) natural, which implements a polar plot pattern that mimics the modest positive directivity of the natural ear; and (4) automatic, which automatically switches between the aforementioned modes. Furthermore, the ASM 2.0 features wind noise reduction and an automatic volume control. The SONNET processor also allows for datalogging.

Additionally, the SONNET sound processor has a new transmitting cable that connects to the D coil. The new transmitting cable is reported to be more durable than the cable of the OPUS 2 sound processor. Also, the SONNET features an integrated telecoil for telephone and induction loop use. Further, the SONNET is wireless ready, which refers to the fact that it possesses a 2.4 GHz antenna that will allow for direct streaming of audio signals from MED-EL designed wireless accessories to the SONNET via digital RF transmission. These wireless accessories were not available at the time of this writing.

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Finally, the SONNET sound processor is capable of delivering electroacoustic stimulation. An acoustic receiver is present at the apex of the SONNET processor. The conventional earhook may be removed and replaced with an acoustic earhook that may be coupled to any standard earmold. The acoustic receiver of the SONNET is capable of an overall gain of 48 dB SPL and a maximum power output of 118 dB SPL. It also possesses 6 channels to allow for flexible programming of the low-frequency acoustic signal. The SONNET is actually available in two variations, an electric-only version and an electric-acoustic version, which contains the acoustic receiver within the processor.

## KEY CONCEPTS

Following this chapter, the reader should recognize and understand the following concepts:

- The basic operation of cochlear implants is similar across the three manufacturers.
- The basic components of cochlear implant systems are also similar across manufacturers.
- The internal devices and sound processors across manufacturers each have several distinctive characteristics, programming parameters, and user controls.

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# 2

# Basic Terminology of Cochlear Implant Programming

Although several parallels exist between cochlear implants and hearing aids, numerous differences occur in the terminology used in the programming of each device. With that in mind, the goal of this chapter is to familiarize the reader with fundamental terms and concepts that are encountered when programming a cochlear implant. The first half of the chapter discusses how various implant programming parameters influence how the acoustical signal is coded into an electrical signal in each of three domains: (1) intensity, (2) frequency, and (3) time. The second half of the chapter addresses basic terminology associated with cochlear implant programming and signal coding strategies.

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## PARAMETERS AFFECTING SIGNAL CODING IN THE INTENSITY DOMAIN

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A summary of the parameters that influence coding in the intensity domain are provided in Table 2–1.

### Stimulation Levels

In most cases, the most important parameters a clinician determines when programming a recipient's cochlear implant is the magnitude of stimulation provided from the implant to the auditory nerve. The fundamental goal of programming is to restore audibility for a range of speech sounds extending from soft speech to loud speech. Ideally, stimulation levels also are set to optimize identification of speech sounds. Finally, it is desirable to set stimulation levels so that normal loudness percepts are restored for speech in addition to environmental sounds. Sounds that are perceived as soft to a person with normal-hearing sensitivity also should sound soft to a cochlear implant user, while sounds that are perceived as loud for a person with normal-hearing sensitivity also should be loud, but not uncomfortable to the user.

Achieving the aforementioned goals is challenging because speech and everyday acoustical sounds have a wide range of intensities of approximately 100 dB. For most cochlear implant users, this wide range of intensities must be coded into a relatively small electrical dynamic range (often around 20 dB when converted from clinical units to dB; Nelson, Schmitz, Donaldson, Viemeister, & Javel, 1996; Zeng & Galvin, 1999). The electrical dynamic range is defined as the difference between the cochlear implant user's perceptual threshold and most comfortable level (i.e., loud but not uncomfortable) for electrical stimulation. Cochlear implant manufacturers use various

**Table 2–1.** Summary of Factors Affecting Signal Coding in the Intensity Domain

| *Term* | *Brief Description* |
|---|---|
| Threshold (T level, THR) | Softest electrical input level detectable by user for each electrode contact |
| Upper-Stimulation Level (C Level, M Level, MCL) | Electrical input level that is perceived as loud, but comfortable to user |
| Amplitude | Current amplitude per phase of a biphasic electrical pulse |
| Pulse Width | Duration or length of one single phase (typically in microseconds) |
| Electrical Dynamic Range | Difference in electrical units between threshold and upper-stimulation level |
| Input Dynamic Range (IDR) | Range of acoustic inputs mapped into electrical dynamic range |
| Sensitivity | Determines gain of processor microphone, which subsequently determines the quietest sound picked up by the microphone |
| Channel Gain | Controls per-channel location of stimulation within the electrical dynamic range |
| Volume | Determines the upper-stimulation level setting used by the recipient, which subsequently determines the loudness level of the incoming acoustic signal |
| Compression | Codes the acoustic inputs into the measured electrical dynamic range |

types of compression to code the acoustical inputs of interest into the recipient's measured electrical dynamic range. This section introduces programming parameters related to the electrical dynamic range, whereas the following section discusses how these parameters dictate the electrical dynamic range.

## Threshold of Stimulation

The threshold of electrical stimulation refers to the least amount of stimulation a recipient can detect when electrical signals (typically biphasic electrical pulses) are delivered to individual electrode contacts. In practice, the exact definition and name of the electrical threshold of stimulation varies across programming software manufacturers. For Advanced Bionics cochlear implants, the electrical threshold is comparable to the audiometric threshold and is best defined as the lowest amount of electrical stimulation a user can detect with 50% accuracy. For Cochlear Ltd. Nucleus (heretofore referred to as Nucleus implants) implants, the electrical threshold is defined as the minimum amount of electrical stimulation the recipient can detect 100% of the time. In contrast, MED-EL defines electrical threshold as the highest level at which a response is not obtained (i.e., the amount of electrical stimulation just below that which elicits an audible percept). Abbreviated terms, such as "T level" for Advanced Bionics and Nucleus and "THR" or "threshold" level in MED-EL devices, often are used to describe the electrical threshold.

In adults and older children, the measurement of electrical threshold is relatively straightforward and involves psychophysical measures frequently used in diagnostic audiology. Most commonly, recipients are instructed to raise their hands or say "yes" when they hear the pulsed programming signal. To a cochlear implant user, this signal is usually perceptually similar to a beep or tone used in pure tone audiometric assessment, although in relatively rare cases, the recipient may describe the signal as possessing a scratchy or static noise-like percept. A persistent

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(i.e., throughout the several weeks of implant use) report that the programming possesses a static noise-like quality may suggest that stimulation is being delivered to an area of the cochlea with substantial neural degeneration or a void of neural innervation. Additionally, it may be possible that stimulation is eliciting a response in an area of the auditory nervous system that has long been deprived of auditory stimulation.

The clinician usually employs a modified Hughson Westlake ascending/descending adaptive procedure to determine threshold (Carhart & Jerger, 1959). The ideal step size for testing likely varies by manufacturer because stimulation scales are different. Larger step sizes may be used for young children who have limited attention spans. For some recipients with tinnitus, determination of threshold may be difficult because the signal tends to "blend" with the ongoing tinnitus. This problem is especially prevalent in newly implanted users.

An alternative approach to the measurement of electrical threshold is to ask recipients to count the number of beeps they hear. When using the procedure, the clinician randomly alters the number of signals presented at each trial (typically between two to five beeps at a time), and the user's task is to report accurately the number of beeps heard. This method assists recipients in focusing on a pulsed signal rather than the continuous tinnitus and enhances their likelihood of responding to the signal rather than the tinnitus. Furthermore, it assists the clinician in obtaining valid T level responses for recipients who provide numerous false positive responses because of difficulty with distinguishing between the programming stimulus and their tinnitus. Use of a count-the-beeps approach often results in higher thresholds than thresholds based on conventional audiometric threshold procedures (Skinner, Holden, Holden, & Demorest, 1995). According to research, a higher threshold level may improve the salience of soft speech and environmental sounds (Holden, Reeder, Firszt, & Finley, 2011; Skinner, Holden, Holden, & Demorest, 1999).

Psychophysical loudness scaling is an alternative approach that is used to set the threshold of electrical stimulation. With this method, the clinician typically begins presentation of the signal at a sub-threshold level and gradually increases the presentation level over time (i.e., ascending approach). Recipients are asked to acknowledge their loudness percept of the signal by pointing to categories on a loudness scale similar to the scale pictured in Figure 2–1. Electrical threshold typically is set at a level that corresponds to "barely audible" or "very soft" on the scale. Similar to the count-the-beeps approach, this method likely results in higher threshold levels than what is obtained with conventional audiometric procedures. In addition, it results in better detection for low-level speech and environmental sounds (Skinner et al., 1999).

Obtaining threshold levels may be difficult in young children with cochlear implants. The techniques for establishing threshold levels vary by age and are similar to behavioral testing approaches used by pediatric audiologists to evaluate hearing sensitivity. A detailed discussion on the measurement of thresholds in children is provided in Chapter 3.

In an effort to expedite the programming process and eliminate an audible noise that may occur when T levels are set too high in CIS-type signal coding strategies, Advanced Bionics and MED-EL have introduced programming strategies that do not require the measurement of electrical threshold. Instead, the electrical threshold is set at 0 units of stimulation or a certain percentage (e.g., 10%) of the maximum stimulation level. The magnitude of stimulation provided for low-level inputs is determined by compression processing and logarithmic mapping functions, which place the acoustic input range into the upper portion of the typical recipient's electrical dynamic range. Previous research initially suggested that speech recognition and audibility for soft sounds are similar when using maps with measured threshold levels versus estimated threshold levels (Spahr & Dorman, 2005). In contrast, Holden et al. (2011) showed that detection thresholds, for warble-tone stimuli presented in the sounds field, were often lower (i.e., better) and recognition of low-level speech was better when subjects used programs with measured threshold levels. As a result, in some cases it may be beneficial to measure electrical-threshold levels rather than rely on the program's attempt to use estimated threshold levels,

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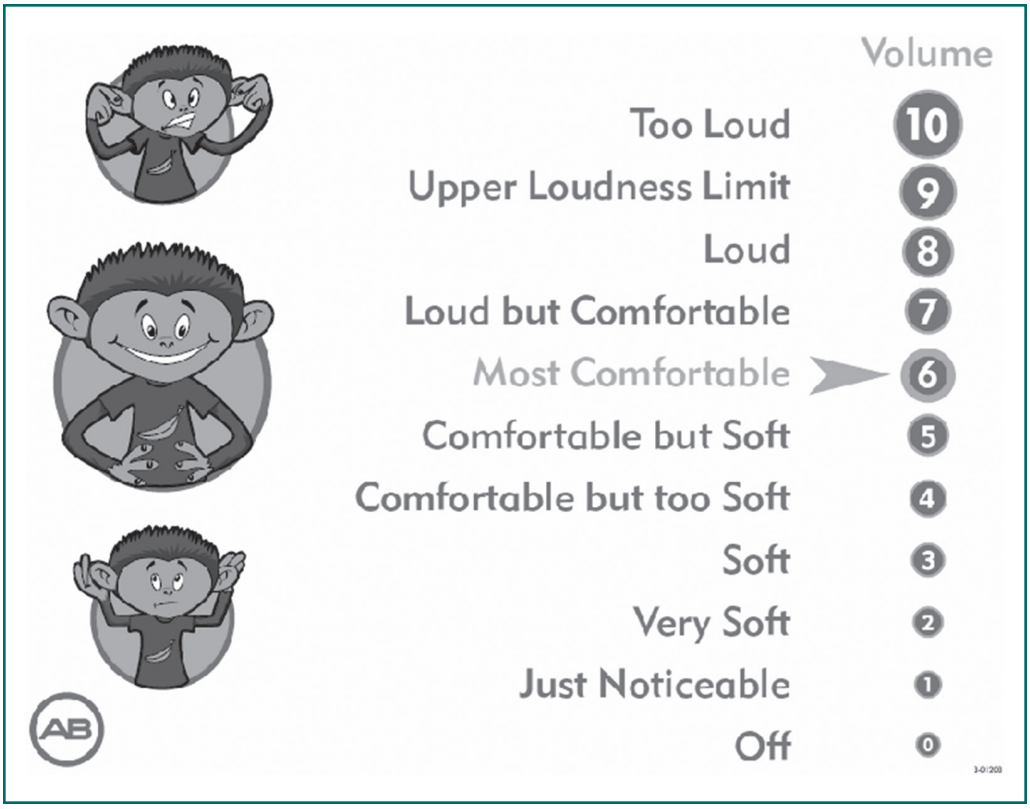



**FIGURE 2–1.** Example of a loudness scale for determining electrical threshold. *Note*. Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

particularly for children who require the audibility of low-level speech sounds for speech recognition and speech and language development. At the very least, clinicians should evaluate a recipients' ability to hear low-level sounds in the sound field. The audiologist should consider increasing the electrical threshold when behavioral hearing thresholds exceed 30 dB HL for adults or 25 dB HL for children (or adjust other manufacturer-specific parameters that enhance audibility for low-level inputs; in particular, MED-EL recommends adjusting other parameters, such as the maplaw, to improve audibility for low-level sounds (See Chapters 6 and 11 for more detail).

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## Upper-Stimulation Levels

The programming parameter related to the upper level of stimulation also varies by terminology and definition across manufacturers. In the Advanced Bionics system, the upper limit of electrical stimulation is set at a level the user perceives as "most comfortable." This parameter is similar to the most comfortable listening level frequently measured in hearing aid evaluations and, in Advanced Bionics cochlear implants, is commonly known as the "M level." In the MED-EL system, upper-stimulation levels are known as "maximum comfort levels" (i.e., MCL) and are defined as the amount of electrical stimulation considered to be "loud, but not uncomfortable." For Nucleus implants, upper-stimulation levels are known as C levels and are set to a level of stimulation the user considers to be "loud, but comfortable."

A cochlear implant user's upper-stimulation levels are critically important because they influence speech recognition, sound quality, and in the case of prelingually deafened children, the ability to monitor one's own voice and produce intelligible speech. If upper-stimulation levels are set inappropriately, the recipient will likely experience poor outcomes. Clinicians typically set upper-stimulation levels via psychophysical

loudness scaling methods or through behavioral observation. Additionally, several researchers report that the electrically evoked stapedial reflex threshold (ESRT) serves as a useful guide for determining upper-stimulation levels (Brickley et al., 2005; Buckler & Overstreet, 2003; Wolfe & Kasulis, 2008). A detailed discussion of setting upper-stimulation levels is provided in Chapter 3.

## Current Amplitude and Pulse Width

As previously mentioned, contemporary cochlear implants deliver biphasic electrical pulses (Figure 2–2) to electrode contacts across the electrode array. Increases in the stimulus intensity can be obtained in two ways. First, the amplitude (or height of each phase of the pulse) of the current (in amperes) can be increased. Figure 2–3 provides an example in which the intensity of the original stimulus (Figure 2–3A) has been doubled by increasing the current amplitude by a factor of two (Figure 2–3B). Second, the stimulus intensity can be increased by lengthening the pulse width. The pulse width describes the duration of each phase of the biphasic programming stimulus and is typically measured in microseconds. Figure 2–3C illustrates a situation in which the stimulus intensity has been doubled by increasing the pulse width by a factor of two. Therefore, the total magnitude of a biphasic, electrical pulse is determined by the amplitude of the current and the width of each pulse. The total energy in the stimulus is specified in charge units (Current amplitude * Pulse width = Total charge in nanocoulombs [unit of charge]). Figure 2–3D provides a description of how pulse width is measured.

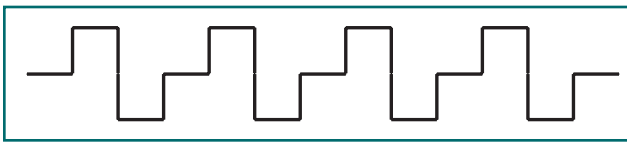
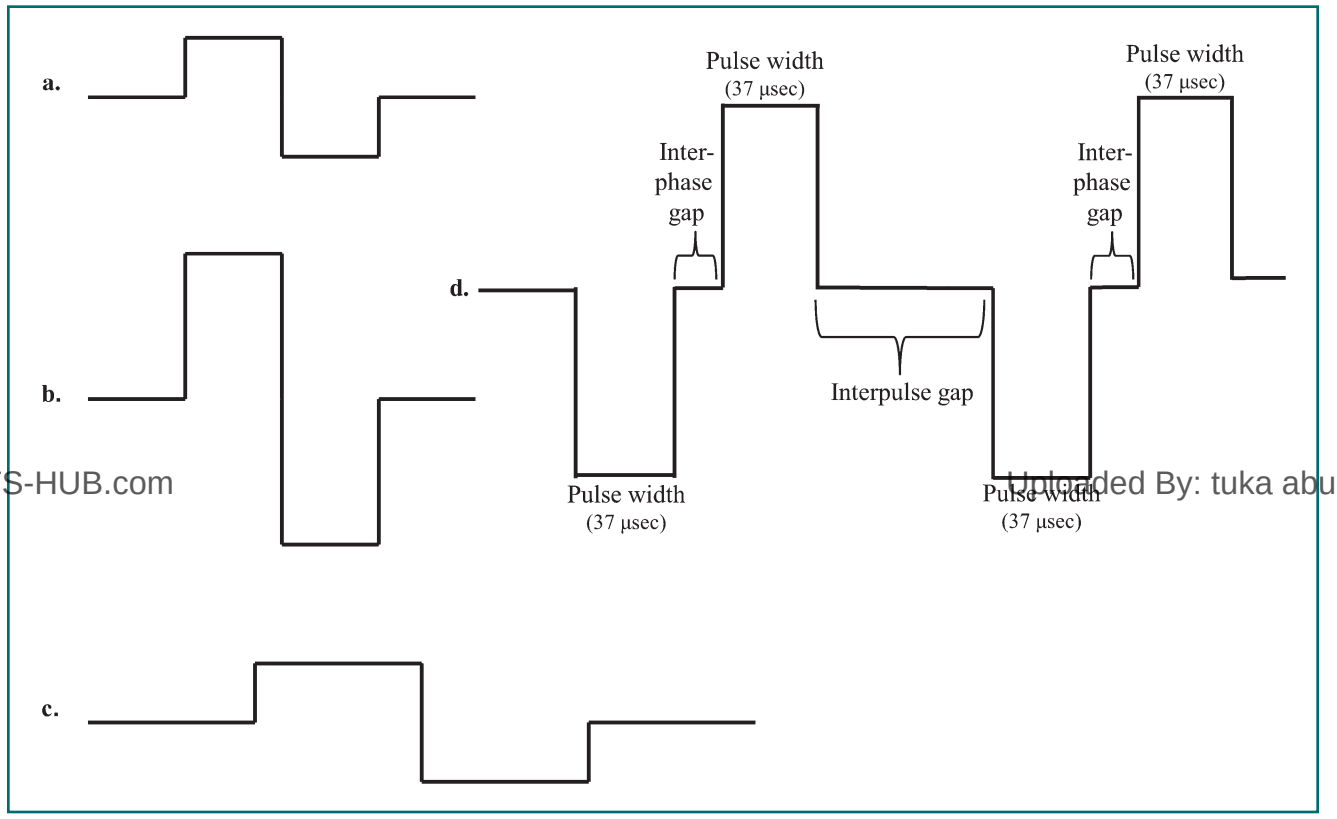


**FIGURE 2–2.** Illustration of a biphasic electrical pulse.




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**FIGURE 2–3.** Increase in intensity from an original signal (**a**) achieved with increased current amplitude (**b**) and increased pulse width (**c**) and a description of how pulse width is measured (**d**).

## Mapping Acoustical Inputs into the Electrical Dynamic Range

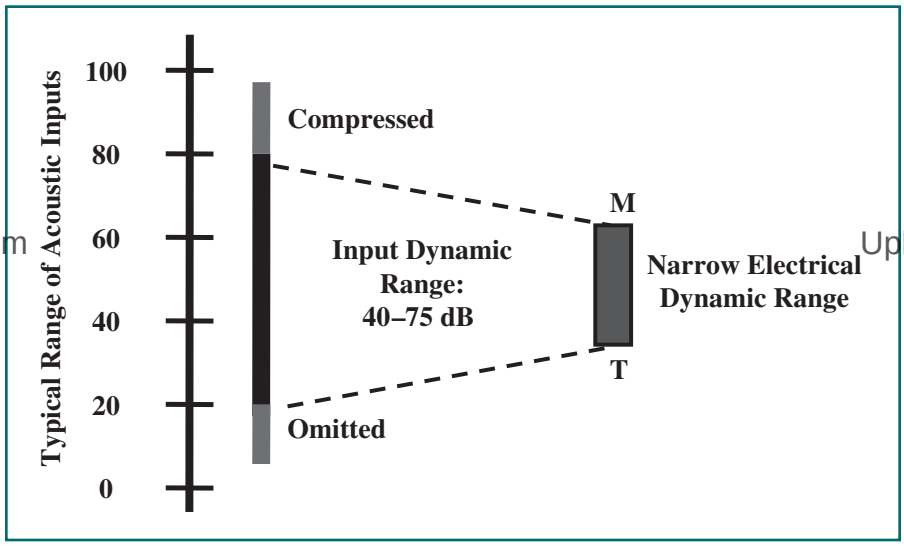
The three cochlear implant manufacturers in the United States take different approaches to directing desirable acoustic inputs into the recipient's electrical dynamic range. The following parameters influence the provision of acoustical inputs into the recipient's electrical dynamic range: input dynamic range, sensitivity, compression, channel gain, and volume control. All of these parameters, as described here, influence the way the signal is coded in the intensity domain. A more detailed explanation of how these parameters are used during signal coding and programming is provided in Chapters 4, 5, and 6.

### *Input Dynamic Range (IDR)*

The input dynamic range of a sound processor is a clinician-adjustable parameter that, along with sensitivity, determines the range of acoustic inputs that are mapped into the recipient's electrical dynamic range. As shown in Figure 2–4, the default IDR of contemporary cochlear implant systems ranges from 40 to 60 dB. The lower end of the IDR determines the acoustic input level that is mapped near the threshold of electrical stimulation, whereas the upper end of the IDR determines the acoustic input level that is mapped near the maximum electrical stimulation level (i.e., M level, MCL, or C level).

In current cochlear implants, the lower end of the IDR usually is set between 20 to 30 dB SPL, while the upper end of the IDR is set between 65 and 85 dB SPL depending on the manufacturer. Acoustic inputs that fall below the lower end of the IDR are mapped below the recipient's electrical dynamic range and are presumably inaudible to the user. Acoustic inputs that exceed the IDR are subjected to infinite compression. It should be noted that the implementation of IDR in signal coding is a bit more complicated than what has been previously described. For example, Nucleus also uses a parameter known as *instantaneous input dynamic range* (IIDR), which refers to the range of short-term fluctuations that are mapped without compression (or other types of attenuation, such as Autosensitivity Control) into the recipient's electrical dynamic range. The IIDR is typically set to 40 dB SPL in order to capture the range of ongoing speech amplitudes, from peaks to valleys, at a given input level.

Cochlear defines IDR as the total range of input levels the processor can analyze without clipping or significant distortion. In Nucleus processors, a change (either manually or automatically with Autosensitivity Control [ASC]) in




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**FIGURE 2–4.** Illustration of electrical dynamic range as measured in the sound field and input dynamic range (IDR) in dB SPL. *Note*. *T*/*THR* = threshold; *C*/*M*/*MCL* = most comfortable level.

another parameter, known as sensitivity, results in a shift of the 40 dB IIDR window. As a result, the range of inputs that are mapped into the Nucleus recipient's electrical dynamic range is actually much wider than 40 dB. This concept is discussed in detail in Chapter 5 of this book.

### Sensitivity

The parameter known as sensitivity controls the gain provided by the sound processor microphone. For some sound processors (e.g., MED-EL OPUS 2 and Nucleus Freedom), the sensitivity parameter only adjusts the gain for the signal received by the sound processor microphone, while for other processors (e.g., Nucleus 5 and Nucleus 6), the sensitivity parameter affects the gain of signals from the microphone and the direct auditory input source. The numerical value of the sensitivity parameter varies by manufacturer and is discussed more thoroughly in Chapters 4, 5, and 6.

Sensitivity interacts with the IDR/IIDR parameter to determine how acoustic inputs are mapped into the recipient's electrical dynamic range. In essence, a reduction in microphone sensitivity adjusts the IDR in a predictable manner. As shown in Figure 2–5, when the sensitivity of the Nucleus 6 sound processor is reduced from a default setting of 12 to a lower setting (e.g., closer to zero), the acoustic input level required to engage the high-level compressor increases. Also, the floor of the IDR/IIDR increases to reduce the audibility of low-level sounds. Conversely, if the microphone sensitivity is increased above the default setting, audibility for low-level sounds is improved and acoustic inputs below the default ceiling of the IDR/IIDR are subjected to infinite compression. Overall, the sensitivity control makes adjustments to the input signal prior to frequency analysis; therefore, its effect is similar across the frequency range of the processor.

### Compression

All contemporary cochlear implant systems use compression to direct a wide range of acoustic inputs into the implant user's narrow electrical dynamic range. In essence, the compression acts as an automatic gain control, but the gain adjustment is determined as a function of input level and the time constants that govern the responsiveness of the compression vary widely across manufacturers. For example, all three U.S. commercial cochlear implant systems use a feature known as Automatic Gain Control (AGC), but the manner

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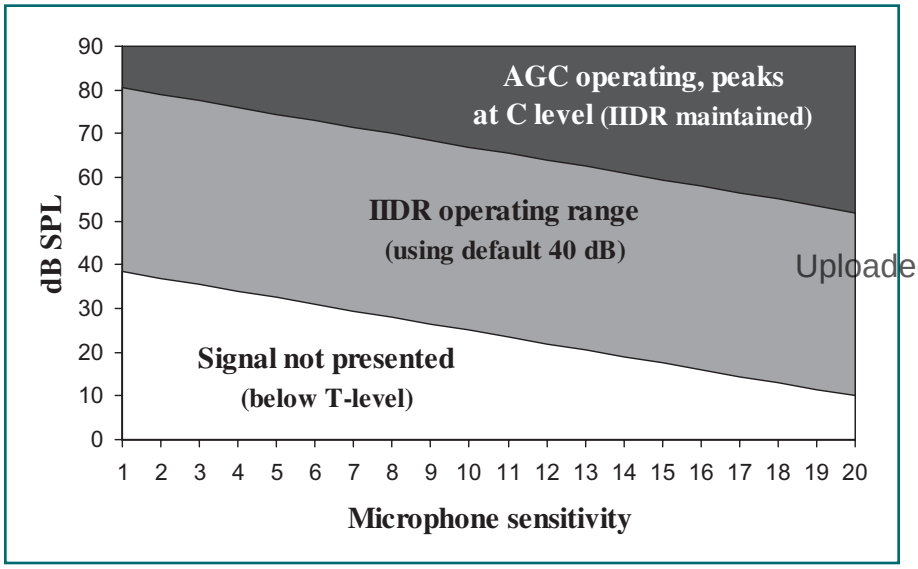



**FIGURE 2–5.** Relationship of microphone sensitivity setting to compression and audibility of high- and low-level sounds. *Note*. Provided courtesy of Cochlear™ Ltd., © 2005 Cochlear Ltd.

in which this feature is implemented is very different among the three systems. Certain characteristics of the AGC in the Advanced Bionics and MED-EL systems can be adjusted by the clinician (see Chapters 4 and 6). However, Nucleus systems offers a special type of input processing known as *ASC*, which basically operates like a slow-acting compressor in response to moderate to high-level input noise. Additional compression is also provided in the Nucleus system for moderate to high-level sounds that exceed the upper end of the IIDR. Please see Chapter 5 for additional information pertaining to signal processing within the Nucleus sound processor.

### Channel Gain

Adjustments to channel gain controls the amplification provided to the signal in a channel-specific and, therefore, frequency-specific manner. Similar to the other parameters, the effect of changing the channel gain varies by manufacturer. In cochlear implant systems using CIS-type strategies, an increase of the channel gain results in a frequency-specific increase to the input signal. An increase in channel gain should enhance the salience of the signal in a frequency-specific manner without the need for increases in the upper-stimulation level. This may be appealing when increases in the upper-stimulation level produce undesirable effects, such as poor sound quality, facial nerve stimulation, discomfort, and so forth.

For Nucleus devices, adjustment of the channel gain also affects the signal in a frequency-specific manner as it leaves the sound processor microphone. In other words, the effect of gain occurs prior to the processing of the signal. As a result, the effect of channel gain on the signal varies by the signal coding strategy and may result in certain sounds being more or less likely to be delivered to the electrode contacts for stimulation of the auditory nerve. The effect of channel gain is discussed further in Chapter 5.

### Volume Control

In general, the volume control parameter is fairly straightforward. As one would expect, adjustments to the volume control affect the user's perception of signal loudness. For all three systems commercially available in the United States, a change in the volume control setting produces a change in the upper level of stimulation the recipient receives from the implant (i.e., C levels, M levels, or MCL). Depending on the programmed settings, however, adjusting the volume control to the minimal position may result in little to no sound percept for the patient, or this action may result in a minimal change in overall stimulation relative to what is provided with the volume control at the maximum position. Furthermore, changing the volume control setting to the maximum position may result in only a small change to stimulation levels, or this action may result in a dramatic increase in the upper-stimulation level. Therefore, it is very important that the clinician and recipient share a thorough understanding of the effect of the volume control on the stimulation provided to the user. The exact effect of the volume control setting varies by cochlear implant manufacturer, and as a result, the volume control parameter is discussed in detail when manufacturer-specific information is discussed in Chapters 4, 5, and 6.

## PARAMETERS AFFECTING SIGNAL CODING IN THE FREQUENCY DOMAIN

### Electrode Contact Versus Channel

As described in Chapter 1, the term *electrode contact* describes a physical contact in the internal device electrode array where stimulation is delivered to the auditory nerve fibers. The term intracochlear electrode contact describes an electrode contact that is located inside the cochlea and one that is typically used for the delivery of stimulation to auditory nerve fibers. The number of intracochlear electrode contacts varies across manufacturers (Advanced Bionics: 16, Nucleus: 22, and MED-EL: 24 electrode contacts arranged in pairs to create 12 separate stimulation sites). Frequency information in the incoming acoustic signal is primarily conveyed by the place of stimulation within the cochlea. Cochlear implant signal

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coding strategies take advantage of the tonotopic organization of the typically functioning cochlea by delivering low-frequency inputs to electrode contacts located in the apical end of the cochlea and high-frequency inputs to electrode contacts located in the basal end of the cochlea.

A *channel* describes a discrete frequency range over which sound is analyzed for eventual delivery to an electrode contact. Channels are defined by bandpass filters with a relatively steep roll-off (although some overlap does exist between neighboring channels). In many cases, electrode contacts and stimulation are similar. For instance, as seen in Figure 2–6, inputs that are processed through channel 1 are delivered to electrode contact number 1. In other rare cases, the channel number and electrode number may be different. As seen in Figure 2–7, inputs processed through

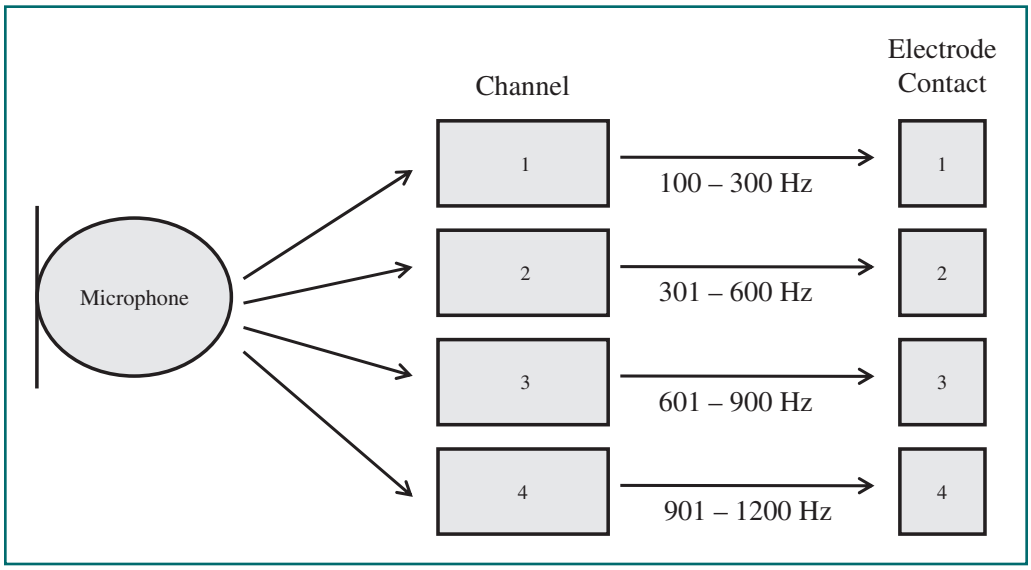
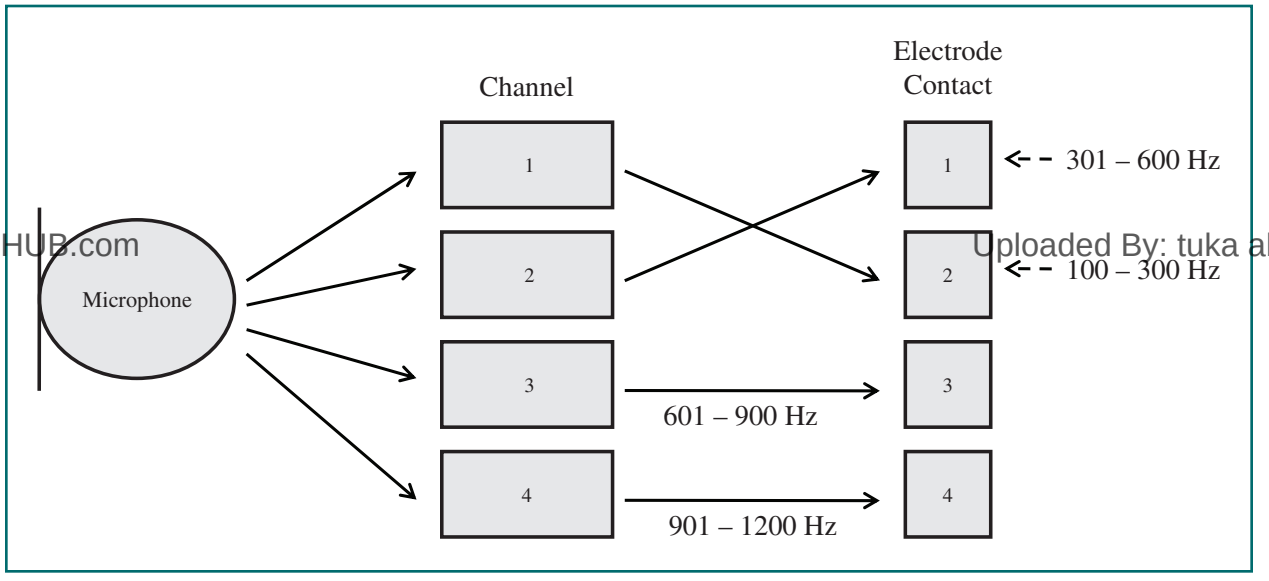



**FIGURE 2–6.** Matching channel to electrode assignment.




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**FIGURE 2–7.** Switching channel to electrode assignment to restore tonotopicity.

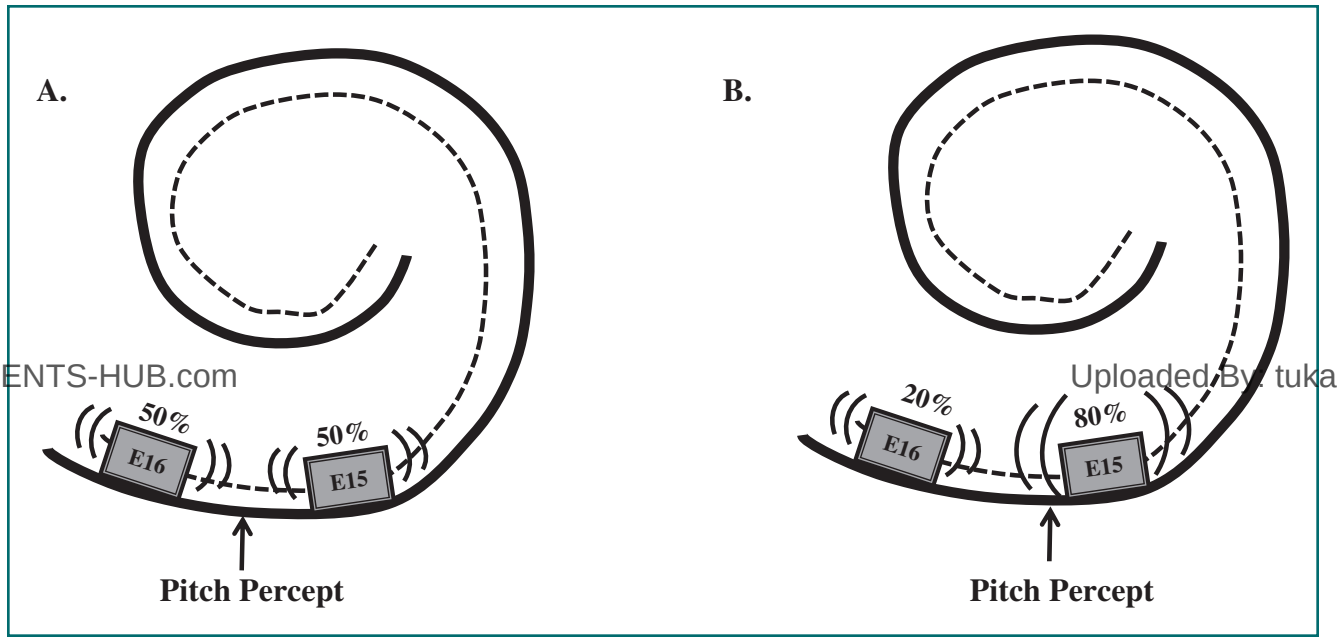
channel 1 are actually delivered to electrode 2, and inputs processed through channel 2 are delivered to electrode 1. In the latter case, adjustments were made to improve the patient's tonotopic organization. The subject's pitch percept did not transition in the expected low-to-high manner when stimulation was swept across the electrodes. Therefore, switching the channel to electrode assignment restored the desired tonotopic organization. Double channel mapping is another example in which the inputs from two channels are delivered to one electrode, but again, the need for double channel mapping is more the exception rather than the rule.

## Virtual Electrodes (Current Steering)

In an attempt to provide a greater number of sites of stimulation across the cochlea, Advanced Bionics and MED-EL both use an alternative approach known as virtual electrodes or current steering. In the Advanced Bionics current-steering approach, two neighboring electrodes are stimulated simultaneously, resulting in a focal provision of stimulation between the two contacts. As seen in Figure 2–8A, when two neighboring electrodes are stimulated simultaneously with equal magnitude, the focus of stimulation falls within the middle of the two electrodes. If the magnitude of stimulation is greater at one of the electrodes, then the focus of stimulation will occur closer to that electrode (Figure 2–8B).

Intermediate pitches also can be created using sequential stimulation, as implemented in the current MED-EL approach, and require the use of bell-shaped overlapping bandpass filters to elicit pitch percepts that shift gradually within and across channels along the electrode array. The Cochlear Nucleus implant also has the capability of applying current steering through sequential stimulation, but at this point, the approach is not used in clinically available signal coding strategies.

The objective of the virtual electrode/current steering approach is to increase the spectral-resolution abilities of recipients. The normal cochlea contains about 3,500 inner hair cells, each of which is tonotopically tuned to one individual frequency and one corresponding afferent auditory




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**FIGURE 2–8.** Illustration of current steering. **A.** Two neighboring electrodes stimulated simultaneously with equal magnitude. **B.** Two neighboring electrodes stimulated simultaneously, but one with greater magnitude.

nerve fiber. This exquisite design of the normal peripheral auditory system allows for discrimination between acoustic components that differ by approximately 2 Hz (Wier, Jesteadt, & Green, 1977). Adept spectral resolution is critical for discrimination of consonants that differ on the basis of place cues, optimal recognition of melodies and enjoyment of music, pitch perception, and speech recognition in the presence of competing noise. Given the fact that cochlear implants provide far fewer transduction sites than the normal ear (12 to 22 physical electrode contacts versus 3,500 inner hair cells), the concept of virtual electrodes (current steering) is an attempt to increase the spectral fidelity of the signal delivered to the auditory nerve. The clinical implementation of this concept along with expected clinical benefit is discussed in Chapters 4 and 6.

### Frequency Allocation

As the name implies, the parameter, "frequency allocation," controls how frequencies are delivered across the active channels. In the Advanced Bionics system, frequency allocation is determined automatically by the programming software. If an electrode is disabled, the frequencies then are reallocated to existing active electrodes. In the Nucleus devices, adjustment of the frequency allocation parameter affects the higher limit (or the highest frequency to be delivered to the most basal electrode), as well as the width of bandpass filters in the mid-frequency channels. An additional parameter in the Nucleus programming software referred to as "edit frequency boundaries," allows the clinician to specify the exact cutoff frequencies of each channel. In the MED-EL device, clinicians may change the lower (70 to 350 Hz) and higher (3500 to 8500 Hz) frequency limits in 10- and 500-Hz intervals, respectively. Also, clinicians can select from four different frequency allocation tables that are each based on different psychoacoustically derived approaches. Additional information on the frequency allocation for the Nucleus and MED-EL devices is provided in Chapters 5 and 6.

## PARAMETERS AFFECTING SIGNAL CODING IN THE TIME DOMAIN

### Stimulation Rate

As stated previously, contemporary cochlear implant systems deliver trains of biphasic electrical pulses to the intracochlear electrode contacts. The stimulation rate typically refers to the number of biphasic pulses that are delivered to an individual electrode contact within 1 second and is specified in pulses per second (pps). The earliest multiple channel cochlear implants had relatively slow stimulation rates (i.e., 250 pps or less), but contemporary systems allow for much higher stimulation rates (i.e., up to about 5,000 pps). Aside from coding temporal cues, changes in stimulation rate also can result in changes in the user's pitch and intensity percepts. Specifically, faster stimulation rates often result in a louder signal due to temporal summation and a higher pitch percept, particularly as rate is increased from 50 to 500 Hz. In fact, one of the theoretical underpinnings of the MED-EL Fine Structure Processing (FSP) signal coding strategy is the attempt to enhance both spectral and temporal resolution in the low-frequency channels by adjusting the stimulation rate to coincide with the frequency of the input signal.

Theoretical benefits can be gained by providing higher rates of stimulation. First, the incoming acoustic signals change in amplitude over time. Slow changes in amplitude (i.e., 2 to 50 Hz) provide the user with an amplitude envelope. The envelope provides temporal cues used to understand speech. Conversely, rapid fluctuations in amplitude carry information about the fine temporal structure of the signal. Acquisition of fine temporal structure is critical for speech understanding in noise, music recognition and appreciation, and potentially, recognition of vocal pitch. Finally, the amplitude envelope and spectrum of some speech sounds are similar; therefore, the listener must rely on differences in fine temporal structure to discern among sounds.

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Additionally, stimulation of the auditory nerve with electrical pulses results in a highly synchronous compound response from auditory nerve fibers adjacent to the electrode contact. In other words, for an initial electrical pulse, all of the auditory nerve fibers in the vicinity of the stimulating electrode contact fire in response to the stimulus, and then they all enter their refractory period at the same time. In contrast, the normal auditory system responds in a stochastic manner to acoustic stimulation. A stochastic response occurs when some (but not all) auditory nerve fibers are excited in response to an individual acoustic stimulus. This stochastic response results in a refractory state for some fibers and an active state for others. As a subsequent acoustic stimulus arrives, the nerve fibers that did not initially fire will be able to respond, whereas the fibers that responded to the original stimulus are in their refractory or recovery period.

It has been suggested that this stochastic behavior of the auditory nerve to acoustic stimuli allows for the coding of the pitch of a sound (Moller, 2001, 2006). The normal auditory system is capable of following changes in amplitude of an acoustic signal in a time-locked manner (Moller, 2001, 2006; Moore, 2003). This property of the normal auditory system may be beneficial because it theoretically allows the system to follow high-frequency stimuli in a per-cycle manner and provides for the acquisition of fine temporal structure. Moller (2001, 2006) also contends that the temporal response of the auditory nerve plays a critical role in the processing of pitch and frequency. He notes that an auditory nerve fiber cannot respond to each individual cycle of a stimulus because some fibers are still in a refractory state. Fibers likely respond to multiples of the incoming stimulus rate. However, the auditory nerve fibers collectively are able to follow the stimulus on a cycle-by-cycle rate. This phenomenon has previously been referred to as the "volley approach."

When using electrical stimulation rates below 2000 pps, the auditory nerve fires in a highly synchronous manner. Because the auditory nerve fibers are firing in unison, and subsequently possess a simultaneous refractory period, the auditory system is unable to follow rapid changes in a stimulus (i.e., fine temporal structure). At stimulation rates above 2,000 pps, the auditory nerve is more likely to assume stochastic firing, and as a result, a higher probability exists that the nerve fibers will respond in a volley type fashion to follow fine temporal structure cues.

In reality, a paucity of research demonstrates consistent improvements in patient performance when stimulation rates exceed 2,000 pps. There are only a handful of published studies that suggest that rates exceeding 2,000 pps provide improvement in speech recognition. For instance, two groups of researchers reported that subjects who converted from slower (800–1,600 pps) to faster (5,000 pps) coding strategies achieved better speech recognition (Dunn, Tyler, Witt, & Gantz, 2006; Koch, Osberger, Segel, & Kessler, 2004). However, the subjects in the Koch et al. study had a longer duration of experience with the higher rate coding strategy; therefore, the improvement may be attributed to learning effects. Also, in both studies, the faster strategy provided twice as many stimulation sites (i.e., electrode contacts) and had a more sophisticated AGC. As a result, other signal processing variables may have been responsible for performance gains.

Most researchers who have explored performance changes as a function of stimulation rate have concluded that the optimal stimulation rate varies on an individual basis (Arora, Dawson, Dowell, & Vandali, 2009; Balkany et al., 2007; Buchner, Frohne-Buchner, Battmer, & Lenarz 2004; Vandali, Whitford, Plant, & Clark, 2000; Verschuur, 2005; Weber et al., 2007). Verschuur (2005) evaluated speech recognition at stimulation rates of 400, 800, and approximately 1,500 pps for a group of MED-EL recipients. He reported that speech recognition generally improved as stimulation rate was increased from 400 to 800 pps, but no additional improvement was typically observed with an increase from 800 to 1,500 pps. However, Verschuur did note that two subjects did show evidence of better performance with the fastest stimulation rate. He concluded that individualized optimization of rate may be helpful to achieve the best outcomes for CI recipients.

Balkany et al. (2007) and Weber et al. (2007) each evaluated speech recognition in quiet and in noise at rates ranging from 250 to 3,500 pps and showed no mean differences in performance

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as a function of stimulation rate. However, both groups of researchers did show individual differences in performance across the various rates with some subjects performing better at slower rates and others experiencing optimal performance at faster rates. Additionally, Balkany et al. (2007) reported that subjects possessed a wide variety in their preference for which stimulation rate elicited the best sound quality, but they did note that most subjects preferred rates lower than 1200 pps. Like Verschuur (2005), Balkany et al. (2007) and Weber et al. (2007) concluded that differences in optimal stimulation rate are likely to exist across recipients, but in most cases, optimal performance is achieved with stimulation rates of 1200 pps and lower.

Clearly, stimulation rates up to several hundred pulses per second seem to be beneficial for all cochlear implant users. Some recipients, however, achieve maximum performance and prefer rates of 500 pps, and others prefer and perform optimally with rates of up to 5,000 pps (Buchner et al., 2004). Again, most recipients do not show improvements in performance with rates exceeding 1,000 to 2,000 pps. As a result, the recipient probably is best served when the clinician identifies the individual's optimal stimulation rate.

The optimal stimulation rate may vary on the basis of cochlear implant hardware as well. To optimize the delivery of high-rate electrical pulses, the current sources of the implant's internal device must be able to generate narrow pulse widths with high current amplitudes. Also, the spacing between electrode contacts may influence a recipient's performance with higher stimulation rates as closer spacing may increase the likelihood of temporal channel interaction. Indeed, reduced channel interaction seems to play a role in better performance with high-rate stimuli as some studies have shown that performance with very fast stimulation rates (e.g., 5,000 pps) is more likely with 8-channel maps (every other electrode contact disabled) versus 16-channel programs in which all active electrode contacts are enabled (Buchner et al., 2004). Furthermore, the sampling rate of the sound processor must be fast enough to capture the fast fluctuations occurring in the input signal (i.e., the sampling rate must be at least twice as high as the highest frequency of interest in the incoming sound). More research is needed to explore the potential benefits and limitations of high stimulation rates across the variety of cochlear implant systems available on the commercial market. The optimal stimulation rate may vary not only across recipients but also by device characteristics.

It should be noted that fast stimulation rates (e.g., several thousand pps) may occasionally be associated with undesirable results. Vandali et al. (2000) evaluated speech recognition in quiet and in noise at stimulation rates of 250, 807, and 1,615 pps. They reported that speech recognition actually decreased as the stimulation rate was increased from 807 to 1,615 pps. Additionally, they reported that higher stimulation rates were associated with a harsher sound quality for some subjects. Furthermore, one subject developed severe tinnitus at the highest stimulation rate.

Anecdotally, the authors of this text have encountered some recipients who experience similar negative effects with faster stimulation rates. For example, some recipients have reported the impression of an echo, harshness, a tinny quality, and distortion when rates of several thousand pps are employed. A common belief amongst CI programming audiologists is that faster stimulation rates are more likely to be detrimental for the very elderly, recipients with an aplastic/deficient auditory nerve, recipients' neural dysfunction (e.g., auditory neuropathy neuropathy spectrum disorder—ANSD), and those with long (greater than 30 years) duration of deafness. Although there is some evidence to suggest that slower stimulation rates may be beneficial for recipients with the aforementioned characteristics (Pelosi et al., 2012; Peterson et al., 2005), there are no large, well-designed published studies that conclusively corroborate this hypothesis.

In summary, the authors of this book tend to select stimulation rates between 900 to 2,000 pps. Specific recommendations pertaining to stimulation rate for cochlear implant systems of different manufacturers are provided in Chapters 4, 5, and 6. For recipients who experience difficulties with rates within this range, the authors will attempt to determine the optimal stimulation rate to maximize the recipient's performance. In cases in which stimulation rates are individually optimized, the

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authors will typically allow the recipient to listen to programs with a variety of stimulation rates (e.g., 250 pps, 500 pps, 1,000 pps, 1,500 pps, 2,000 pps, 3,000 pps) to determine the rate that yields the best sound quality and speech recognition. In cases in which the recipient is uncertain of which stimulation rate is the best, the authors will provide multiple programs (each with a different stimulation rate) to allow the recipient to experiment with a variety of stimulation rates in real-world settings. Finally, despite the presence of conclusive evidence supporting the tactic, the authors tend to gravitate toward slower stimulation rates for recipients who have evidence of auditory nerve dysfunction (e.g., ANSD, auditory nerve dysplasia) as well as for recipients who are very elderly or who have a long duration of deafness and are struggling to make satisfactory progress with their cochlear implant.

It should be noted that manufacturers also frequently refer to an implant's total stimulation rate, or the overall maximum rate of stimulation possible, across all active electrodes within one second. Total stimulation rate usually is calculated by determining the product of the per channel stimulation rate and the number of active channels stimulated for an incoming stimulus (per channel stimulation rate * number of active channels stimulated = total stimulation rate). For example, if an incoming sound is delivered to 10 channels, and the per channel stimulation rate is 1,200 pps, the total stimulation rate is 12,000 pps.

The maximum total stimulation rate usually is determined by the number of pulses per second the implant is capable of delivering. For example, the total stimulation rate for the Advanced Bionics 90K implant is just over 80,000 pps (i.e., approximately 5,000 pps for each channel * 16 channels = approximately 80,000 pps total stimulation rate). There are, however, exceptions to this rule. For instance, the Nucleus Freedom and CI512 devices are capable of a total stimulation rate of 32,000 pps. However, when the default ACE signal coding strategy is selected, the total stimulation rate is 14,400 pps. This limited total stimulation rate for ACE is historically based. When ACE was developed for commercial use, the Nucleus 24 implant was the internal device available at the time, and it was capable of a total stimulation rate of only 14,400 pps. To access higher total stimulation rates in the Nucleus CI512 device, the clinician must select the more recently developed "High-ACE" (ACERE) coding strategy.

## BASIC COCHLEAR IMPLANT TERMINOLOGY

Basic definitions for cochlear implant programming terminology are provided in Table 2–2.

### Electrode Coupling Strategy/ Stimulation Mode

The electrode coupling strategy (or stimulation mode) indicates how channels are connected to form an electrical circuit through which current can be delivered to the auditory nerve. In a complete electrical circuit, current travels from the power source to a resistive component and then to a return location. Cochlear implant stimulation must also be delivered through a complete circuit.

The "electrical circuit" in a cochlear implant is made up of (1) the signal generator in the internal processor (i.e., current source/output circuit), (2) the active electrode lead and intracochlear electrode contact, and (3) the return electrode(s). Depending on the application, the clinician's preferred terminology, and the manufacturer, the return electrodes also may be referred to as ground, indifferent, or reference electrodes. The electrode contact, cochlear fluids, and other tissues adjacent to the active electrode contact serve as the resistive elements in this circuit. The active electrode is always an intracochlear electrode and ideally is located in the scala tympani. In contemporary cochlear implants, the return electrode(s) usually is an extracochlear electrode (located outside of the cochlea). The return electrode typically is located on the case of the internal stimulator. Alternatively, it may be located on the electrode lead, such as the ring electrode in the Advanced Bionics 90K implant, or within a separate lead that terminates in an electrode contact at a remote location from the cochlea, such as the MP2 reference

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**Table 2–2.** Summary of Basic Cochlear Implant Programming Terminology

| *Term* | *Brief Description* |
|---|---|
| Electrode Coupling Strategy/ Stimulation Mode | How electrodes are electrically connected to form electrical circuit; all consist of active and reference electrodes |
| Sequential Stimulation | Presentation of biphasic pulses to electrodes occur one after another |
| Simultaneous & Partially Simultaneous Stimulation | Presentation of biphasic pulses are directed to several or 2 (partial) electrodes at one time |
| Sweeping | Sequential presentation of a programming stimulus across the electrode array (typically at C/M/MCL level to ensure user comfort) |
| Loudness Balancing | Successive presentation of stimulus to 2 or more electrodes, ensures equal loudness |
| Radio Frequency | The carrier frequency for transmission of signals across scalp: from processor to internal device and/or vice versa |
| Bidirectional Telemetry | Transmission of signals to and from the implant. Used for normal stimulation, device programming, and to check integrity of internal device |
| Electrical Impedance | Measurement of electrode function: the amount of current that returns to a reference electrode |
| Voltage Compliance | Maximum amount of current delivered to electrode for a given power source |
| Mixing Ratio | Relative strength of signals from processor and auxiliary input source input source |

*Note*. *C*/*M*/*MCL*= most comfortable level of electrical stimulation for given electrode.

electrode located at the end of the second lead of the Nucleus Freedom implant (see Chapter 1).

*Monopolar stimulation* describes electrode coupling in which stimulation is provided to an active intracochlear electrode and an extracochlear electrode serves as the return (Figure 2–9). All contemporary cochlear implants use monopolar stimulation as the default mode. An alternative approach is *bipolar stimulation*, which employs bipolar electrode coupling whereby electrical current is delivered to an active intracochlear electrode and a neighboring intracochlear electrode serves as the return (Figure 2–10). Bipolar coupling specifically is referred to as bipolar stimulation (BP) when the return electrode is positioned immediately adjacent to the active electrode. When the return and active electrodes are separated by one electrode contact, the bipolar stimulation is referred to as bipolar +1 (BP+1). BP+2 refers to two electrode contacts separating the active and return electrodes; BP+3 refers to three electrode contacts separating the active and return electrodes; and so forth.

Although bipolar stimulation should provide more focused stimulation, monopolar stimulation does allow for the provision of a tonotopic signal across the cochlea. Additionally, the relatively narrow spread of electrical current inherent with bipolar stimulation requires higher current levels (i.e., higher threshold and upper-stimulation levels) to reach a given sound percept when compared to monopolar stimulation. Thus, the higher current levels required for bipolar stimulation may reduce battery life. Furthermore, the broader stimulation pattern provided by monopolar stimulation results in a relatively gradual change in stimulation levels from one electrode to the next (i.e., T levels and C/M/MCL levels), so interpolation is a viable option in the monopolar mode (see explanation of interpolation later

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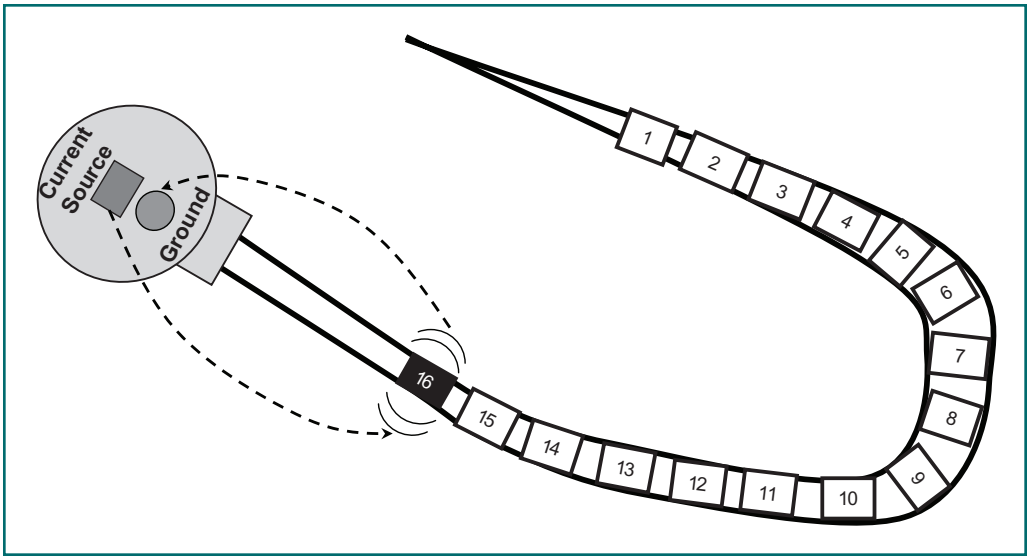
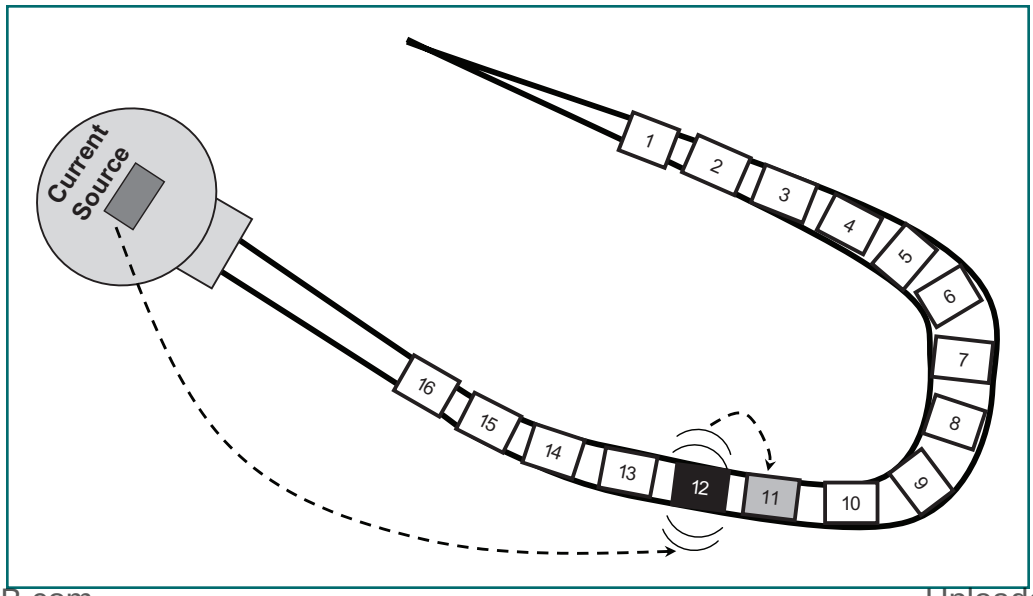



**FIGURE 2–9.** Illustration monopolar electrode coupling where the single reference/ground electrode is at a distance from the active electrodes.




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**FIGURE 2–10.** Illustration of bipolar electrode coupling where the reference/ground electrode is in close proximity to the active electrode.

in this chapter). In contrast, stimulation requirements may change drastically from one electrode to the next with bipolar stimulation; therefore, the clinician must measure T and C/M/MCL levels for all active electrodes.

Common ground coupling is another electrode coupling mode used for diagnostic purposes but is not used for stimulation in contemporary signal coding strategies. In common ground coupling, electrical current is delivered to an active intracochlear electrode, and all of the remaining intracochlear electrodes serve as the collective return (Figure 2–11). Because all of the electrodes are coupled to one another electrically and electrical resistance across these electrodes can be referenced to an extracochlear electrode serving as the

ground (defined as zero voltage potential/reference potential), common ground coupling is the most sensitive mode to detect shorted electrodes during electrode impedance assessment.

Finally, cochlear implant manufacturers are exploring more advanced electrode coupling strategies, such as tripolar coupling and other higher order coupling modes. The potential advantage of higher order coupling modes is to provide more focused stimulation to spiral ganglion dendrites and cell bodies relative to monopolar or bipolar stimulation (Bierer & Middlebrooks, 2002; Bonham & Litvak, 2008). An example of tripolar stimulation is shown in Figure 2–12. With tripolar stimulation, current is delivered to one intracochlear electrode and its two neighboring

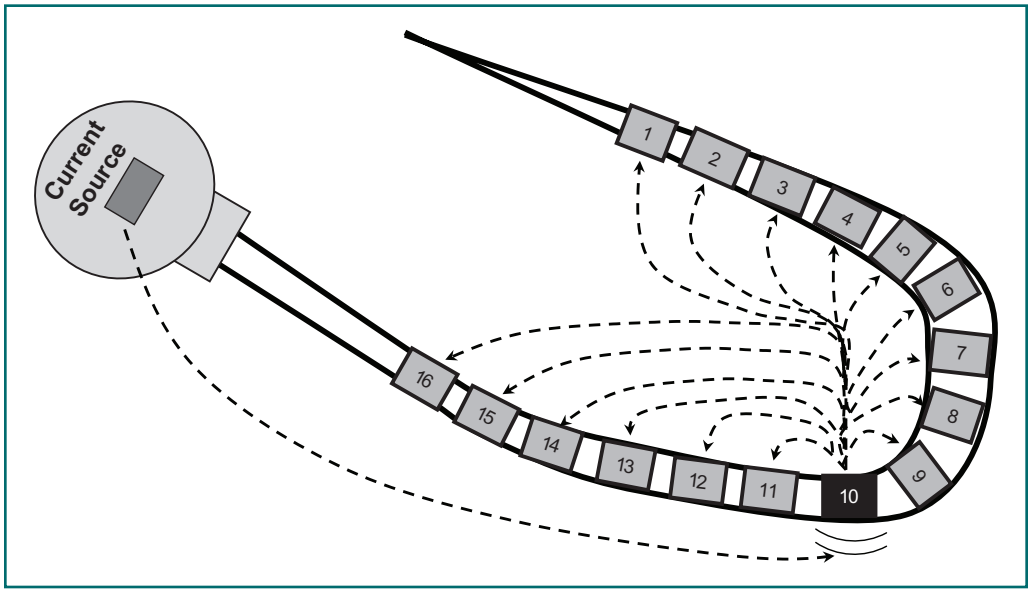



**FIGURE 2–11.** Illustration of common ground coupling where all electrodes adjacent to the active electrode serve as the reference/ground electrode.

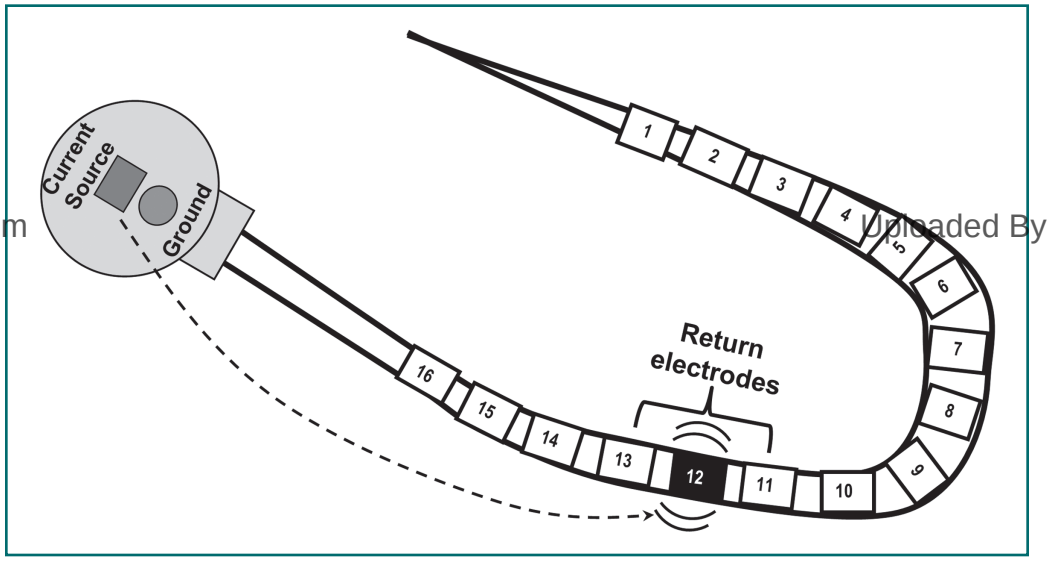



**FIGURE 2–12.** Tripolar stimulation where current is delivered to one intracochlear electrode and its two neighboring electrodes serve as the return electrodes.

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electrodes serve as the return electrodes. Each of these return electrodes receives half of the return current. At the time of this writing, however, no commercial cochlear implant systems allow for these advanced methods of electrode coupling.

## Sequential Versus Simultaneous Stimulation

Most contemporary cochlear implant systems deliver electrical biphasic pulses in a sequential fashion, in which the presentation of pulses occurs one after another. In other words, a simultaneous presentation of two pulses never occurs. This approach is designed to reduce or prevent channel interaction, which may result in distortion or smearing of speech sounds. The previously mentioned concept of virtual channels is a variation of simultaneous stimulation that is intended to elicit one percept or "channel" of stimulation. This approach differs from the traditional concept of simultaneous stimulation, which sought to stimulate across several channels at one time to create multiple percepts.

In contrast to sequential stimulation, simultaneous stimulation refers to the presentation of an electrical signal to several electrodes at one time. An early signal coding strategy called Simultaneous Analog Stimulation (SAS) used simultaneous stimulation. In the SAS signal coding strategy, the input sound was filtered into multiple channels and the acoustic output of each channel was converted into a continuous electrical analog, which was delivered to intracochlear electrode contacts based on the frequency of the respective channel and the position of the contact within the cochlea. Zwolan et al. (2001) showed that Advanced Bionics users with low stimulation level needs (i.e., low T and M levels) frequently performed well and preferred the SAS strategy, whereas recipients with higher stimulation level needs preferred sequential stimulation. Simultaneous stimulation typically is implemented with bipolar electrode coupling in an attempt to reduce channel interaction. Current signal coding strategies do not use totally simultaneous stimulation, primarily to avoid channel interaction.

Advanced Bionics cochlear implants are also capable of using an approach that is best referred to as partially simultaneous stimulation, which is essentially a combination of the two aforementioned approaches. With this approach, biphasic electrical pulses are provided simultaneously to two different electrode contacts, but these electrode contacts are separated by a relatively large physical distance (Figure 2–13). The objective of this approach is to allow for faster rates of stimulation per channel (because two channels are stimulated at the same time) while limiting channel interaction (by ensuring that two electrodes that are simultaneously stimulated are separated by a relatively great physical distance) historically associated with simultaneous stimulation.

## Interpolation

As the number of cochlear implant recipients has grown, cochlear implant clinicians have struggled to manage their expanding patient caseloads. In response, manufacturers have developed streamlined programming strategies to expedite the programming process. A streamlined strategy is basically one that is designed to allow for a program that may be created quickly without sacrificing sound quality or performance. Interpolation refers to the estimation of stimulation levels on the basis of measured values from neighboring channels. For instance, if the measured T levels for electrodes 6 and 8 were 70 and 90, respectively, the estimated T level for electrode 7 would be 80.

In practice, interpolation often is used to estimate the stimulation levels for several electrodes that are between two measured electrodes (e.g., stimulation levels of electrodes 21 through 17 are estimated on the basis of measured values at electrodes 22 and 16). The measured values are sometimes referred to as "anchor channels." In the aforementioned example, the stimulation levels of channels 21 through 17 are estimated based on the stimulation levels measured on the anchor channels, 22 and 16. Research demonstrates that, on average, user performance for a large group of recipients did not decline when programs were based on interpolated versus individual channel measures (Plant et al., 2005). However, because of individual subject variation, the authors recommend that clinicians measure stimulation levels on most active channels when time allows (i.e.,

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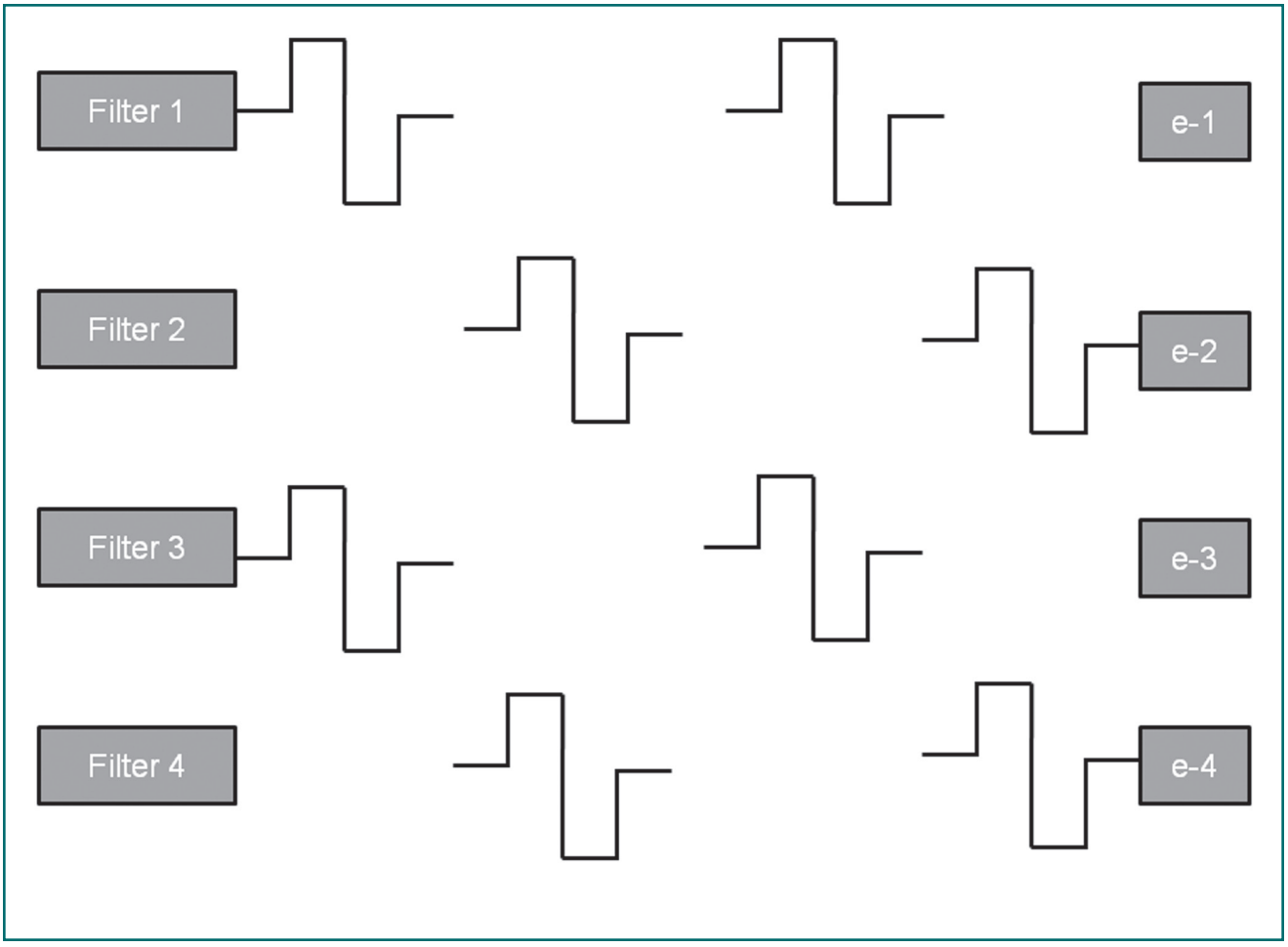



**FIGURE 2–13.** Illustration of partially simultaneous stimulation.

all even-numbered electrodes and any odd electrodes that are adjacent to two even electrodes with stimulation levels that differ substantially). In particular, the clinician should consider measuring stimulation levels when a large difference exists in the measured values of two anchor channels. Also, because the quantity and integrity of auditory nerve elements can vary considerably in the more basal region of the cochlea, the authors recommend measurement of stimulation levels for the last three electrode contacts within the electrode array.

## Sweeping

Sweeping refers to the sequential presentation of the programming stimulus across the active electrodes at a given stimulation level. Sweeping often is performed at the upper-stimulation level (i.e., C/M/MCL level) to ensure that none of the electrodes are too loud or result in unfavorable sound quality. Sweeping, however, may be performed at lower stimulation levels (i.e., 50% of the dynamic range) to ensure equal loudness for lower level inputs, particularly when the recipient complains that certain soft sounds are too loud or that soft speech is difficult to understand. Additionally, the perceived pitch should become progressively higher if sweeping is performed from the apical to basal direction. As such, the prudent clinician may choose to sweep at the upper-stimulation level and ensure that the pitch changes in the expected tonotopic manner. Sweeping may also be conducted to determine whether there are any channels that possess undesirable sound qual-

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ity or to confirm that there are no channels that produce unwanted nonauditory side effects (e.g., facial nerve stimulation). Finally, sweeping may be performed across all active electrodes in succession or just across a couple or a few electrodes at a time to allow for more precise responses.

## Loudness Balancing

Loudness balancing refers to the successive presentation of a programming stimulus to two or more active electrodes to ensure that the subject perceives the stimuli to be equally loud when each is presented at a specified percentage of the dynamic range. Researchers have reported that the provision of stimulation resulting in equal loudness percepts across the electrode array results in optimal sound quality and speech recognition (Dawson, Skok, & Clark, 1997; Sainz, de la Torre, Roldan, Ruiz, & Vargas, 2003). The authors of this book contend that the provision of upper-stimulation levels that are equally loud across the electrode array is one of the most important facets of establishing a program that will optimize sound quality and speech recognition.

Most commonly in clinical practice, two or more electrodes are stimulated at or near (e.g., 80% of MCL) the upper-stimulation level (i.e., C/M/MCL level), and the subject reports whether the stimuli are equally loud or whether one electrode possesses a louder percept. If the latter is the case, the clinician makes an adjustment to achieve balanced loudness between the electrodes. Typically, the clinician begins with presentation to an active electrode at the most apical location in the electrode array and progresses across electrodes in a basal direction. The number of electrodes that are balanced in a given trial often varies across clinicians (i.e., balancing two, three, four, or more at a time), but the authors prefer to balance loudness across two electrodes at a time. For example, if the clinician begins by asking the recipient to judge loudness at C level for electrodes 22 and 21, and electrode 21 is louder than 22, then the stimulation level for electrode 21 would be decreased slightly. The loudness of the two then would be reevaluated. After the recipient reports that the loudness is balanced between the electrodes, the clinician compares loudness at C level between electrodes 21 and 20 and eventually for the rest of the electrode array. The adjustment to the stimulation level always should be made to the second electrode in the pair because the first electrode is balanced to the preceding electrodes in the array.

## Radio Frequency (RF)

As mentioned in Chapter 1, cochlear implants transmit signals across the scalp via near-field magnetic induction at a given radio frequency (RF). The term RF is used to express the carrier frequency of the electromagnetic induction (radio) signal on which the signal of interest is modulated. Some cochlear implant systems use a single RF to transmit signals both to and from the internal device, but the Advanced Bionics Corporation uses one RF for forward transmission and a second RF for backward transmission. This electromagnetic (RF) signal is also used to power the cochlear implant.

## Telemetry

When programming cochlear implants, telemetry refers to the transmission of data through radio frequency communication to and from the internal implant via the electromagnetic induction link. In other words, telemetry is used to send the signal of interest to the internal device for eventual stimulation. Additionally, telemetry is used during the programming session to deliver signals (i.e., T level stimuli) to the implant. Finally, backward telemetry may be used to transmit signals from the internal device to the external coil, sound processor, and programming computer. Backward telemetry allows the clinician to evaluate the integrity of the internal device and measure responses from the auditory nerve. It also may allow for an estimation of the amount of power the internal device needs from the processor.

The terms "lock" and "link" are often used to specify successful radio communication (i.e., telemetry) between the internal and external processors. "Failure to obtain lock/link" happens when the signal cannot be transmitted from the

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external processor to the internal processor or vice versa.

## Impedance

Impedance refers to the resistance to the flow of energy through any medium. Electrode impedance is a measure of the opposition to electrical current flow across an electrode. It is influenced not only by the electrode contact and the electrode lead that is coupled to the contact but also by the medium surrounding the electrode lead and contact. This medium includes cochlear fluids, surrounding tissues (fibrous, bony, etc.), electrolytes, macrophages, proteins, and so forth. The electrode impedance measurement should be conducted at the beginning of every programming session. During the measurement, a small amount of electrical current, which is typically inaudible to the recipient, is delivered sequentially to each active intracochlear electrode. Ohm's law is used to determine the impedance measured at each electrode contact. Briefly, Ohm's law states that:

Voltage (V) = Current (I) * Resistance (R)

Because the voltage used to deliver the stimulus and the current of the return signal are both known, the impedance may be calculated from Ohm's law. The actual impedance measure results are indicated by the current that returns to a reference electrode (also referred to as a ground or indifferent electrode) and how voltage is distributed to the stimulated electrode contact (and in some cases, the remaining intracochlear electrode contacts) relative to a reference electrode. Ideally, electrode impedances will be greater than 1 kohm and less than 15 kohms. An electrode with an excessively low impedance (e.g., <1 kohm) is sometimes referred to as a *short*, whereas an electrode with an excessively high impedance (e.g., >30 kohms) is called an *open circuit*. A short circuit refers to an establishment of relatively low resistance between two points in a circuit, which typically are separated by a much higher electrical resistance. Essentially, the term "short" means that two electrode leads/contacts are electrically coupled to one another in an unintended manner. As a result, a shorted electrode is identified as two electrodes that are electrically connected and consequently possess an identical voltage that is distributed to each electrode when only one is stimulated.

Short circuits may be caused by physical contact between two electrode contacts or electrode leads, an electrical fault within the electrode lead/contact, or excessive distortion or tension on the array. Short circuits should be considered as persistent anomalies and are generally permanently disabled.

Open circuits may be caused by anomalies (e.g., ossification) or by an air bubble or protein buildup in the electrode-tissue interface. Electrode impedances are often relatively high when the electrode contacts have not been stimulated for quite some time (i.e., at initial activation of the cochlear implant). This is most likely due to the presence of proteins/macrophages that have accumulated on the electrode contact and/or to the organization of the fibrous tissue surrounding the electrode contact. In most cases, electrode impedances will decrease substantially after electrical stimulation is delivered to the electrode contacts. As a result, the clinician should not disable electrode contacts with modestly high electrode impedance values (e.g., 20 to 30 kohms) because these values will likely decrease with implant use.

Furthermore, open circuits may be caused by broken electrode leads or faulty electrode contacts. In this case, the electrode impedance will likely be measured at an infinitely high value. From a practical standpoint, the most recent version of each manufacturer's software identifies the presence of abnormal electrodes and "flags" them for management by the programming clinician. Electrodes marked as open circuits should also be disabled but reevaluated after a period of implant use.

Along with the absolute impedance value of each electrode, clinicians also may observe the morphology of electrode impedances (i.e., shape or change of impedances across the electrode array) and changes in impedance values across time. Ideally, electrode impedance values should be relatively similar across the electrode array or vary gradually. Erratic electrode impedance patterns should be evaluated more closely.

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It should be noted that electrode impedance is typically quite low at surgery because the electrode contacts are surrounded by cochlear fluid (e.g., perilymph). Electrode impedance often increases in the first few weeks following surgery due to fibrous tissue growth in the cochlea following surgery (Hughes et al., 2001; Newbold et al., 2004). As previously discussed, it is normal for electrode impedances to be relatively high (e.g., 10 to 20 kohms) after periods of nonuse (e.g., at initial activation, after several days of nonuse). In these cases, a significant decrease in electrode impedance should occur within the first few days or even hours of implant use. Electrode impedances will eventually stabilize during the first few weeks of electrical stimulation (Hughes et al., 2001; Newbold et al., 2004). If electrode impedance values continue to fluctuate after several months of use, the programming clinician should monitor recipient performance closely and consider a referral to an otologist for medical evaluation. In this case, a representative of the manufacturer also should be notified. Persistently fluctuating electrode impedances may be attributed to changes within the cochlea (e.g., fibrous tissue growth), a change in the integrity and/or the electrochemical properties of the recipient's cochlear fluids and tissues (i.e., inflammation, hormonal changes, autoimmune reaction), or a fault in the cochlear implant. The reader is encouraged to refer to Chapters 4, 5, and 6 for important and clinically relevant information pertaining to electrode impedance measurement specific to each cochlear implant manufacturer.

Electrode conditioning is a feature available in some implant systems that allows for the presentation of low-level current to each electrode to remove air bubbles, protein buildup, and so forth. Electrode conditioning is typically only needed prior to testing in the operating room, at initial activation, at a programming session preceded by a prolonged period of nonuse, or when activating electrodes were previously disabled.

Abnormal electrode impedance values may compromise sound quality or produce nonauditory sensations, such as facial nerve stimulation. Other detrimental effects associated with the use of abnormal electrodes include poor speech recognition performance, inadequate loudness growth, sudden changes in loudness, and potential discomfort. Numerous reasons for changes or fluctuations in impedance exist. For instance, the electrode array may become displaced over time. Also, ossification or fibrous tissue may arise in the cochlea after implantation and alter the electrode-neural tissue interface. A Stenver's x-ray or a computed tomography (CT) scan (conventional or ideally a 64-slice three-dimensional image) may be useful in evaluating changes in electrode locations or anatomy of the cochlea. In general, imaging to evaluate the position of the electrode array should be considered if unexplained abnormal electrode results exist or if a sudden unexplained change in electrode impedance occurs. Additionally, changes in a recipient's hormonal levels, which may occur during adolescence, pregnancy, menopause, or hormonal therapy (e.g., testosterone supplementation), may result in a change in electrode impedances. The clinician should consider these conditions in light of fluctuating or changing electrode impedance results.

Finally, a "normal" electrode impedance value does not necessarily imply that all is well. Chapter 11 provides examples of cases in which the absolute electrode impedance values were within normal limits, but a significant problem existed. Furthermore, it should be noted that an absolute electrode impedance value that is within normal limits does not even indicate that the electrode contact is within the cochlea. The clinician should remember that the electrode impedance indicates how electrical current flows across an electrode contact to surrounding tissues or fluid. A "normal" electrode impedance value may be obtained when an electrode is in contact with middle ear tissue or fluids or body tissue remote from the cochlea. Again, it is important for the programming clinician to review the cochlear implant surgeon's postoperative report and/or postoperative x-ray to determine whether all of the stimulating electrode contacts are inserted into the cochlea. Indeed, the authors of this book have encountered recipients who have had normal electrode impedance values across the array but have had electrode contacts residing in the middle ear space or even in one exceptional case, in the cerebellopontine angle. In cases in which the recipient reports unusual or undesirable effects associated

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with stimulation of certain electrodes, the clinician should refer the recipient for imaging to evaluate the placement of the electrode array.

## Voltage Compliance

Each cochlear implant system has a finite voltage capacity (specified by the type, number, and size of battery cells used to power the processor), which determines the maximum amount of current it can deliver. In practice, the amount of power available to operate the implant and the current that will be delivered to an individual electrode is determined by the efficiency of the signal delivery from the external coil to the internal device, the voltage capacity of the battery, and the electrode impedance. In essence, the maximum amount of electrical current available to stimulate an electrode contact is determined by Ohm's law, which states:

Voltage (V) = Current (I) * Resistance (R)

The aforementioned equation determines the voltage compliance level of an electrode, or the maximum amount of current that can be delivered to an electrode by a given power source (e.g., battery). Of course, the voltage is determined by the capacity of the battery and is a fixed value. However, the current available at each electrode is variable across recipients and across electrodes within a recipient and is determined by the impedance at each electrode. For a given voltage (i.e., battery module), a higher impedance will result in a lower amount of current available to stimulate the electrode contact (i.e., a lower voltage compliance level). In contrast, a lower electrode impedance value results in a higher amount of current available to stimulate the electrode contact (i.e., a higher voltage compliance level).

After stimulation reaches the voltage compliance level, additional increases in current level cannot be provided. As a result, increasing the current level in the programming software will not result in loudness growth. If voltage compliance levels have been reached at a current level value that is lower than what is desired or required for adequate loudness growth or performance, then the clinician should increase the stimulus pulse width in order to provide more stimulation. If a recipient uses a map with electrodes that are out of voltage compliance, then it is likely that speech recognition and sound quality will be compromised. Furthermore, it is probable that loudness will not be balanced across the electrode array, and the user may experience some nonauditory side effects, such as facial nerve stimulation, because the current amplitude is at a maximum. The programming software for contemporary cochlear implants provides an indication that voltage compliance levels have been exceeded.

It should be noted that voltage compliance levels may also be avoided by using a battery with a higher voltage capacity. For example, the Nucleus Freedom behind-the-ear sound processor may be operated by a small battery module with two zinc-air 675 batteries or a larger module with three zinc-air 675 batteries. Because the larger module possesses more batteries and a larger voltage capacity, it allows for higher voltage compliance limits (see Figure 2–14 for an example). The Freedom sound processor may also be operated by a body-worn controller that uses two AAA alkaline batteries in order to provide even higher voltage compliance limits if needed.

## Mixing Ratio

The mixing-ratio parameter controls the relative strength of the signal from the speech processor microphone and the direct auditory input (DAI) source. All contemporary cochlear implant sound processors have a DAI port (or a special battery case that facilitates a DAI connection) in order to allow for the direct delivery of an external input, such as a signal from a personal FM/RF system or MP3 player, to the sound processor. In some instances, the user may desire to listen only to the signal from the DAI source (e.g., listening to music from an MP3 player). In many situations, however, a cochlear implant user will want to hear the signals from both the DAI source and sound processor microphone. For example, children using a personal digital RF/FM system in a classroom should hear the teacher's voice from the digital RF/FM transmitter, but also their own voice and the voices of other children in the room.

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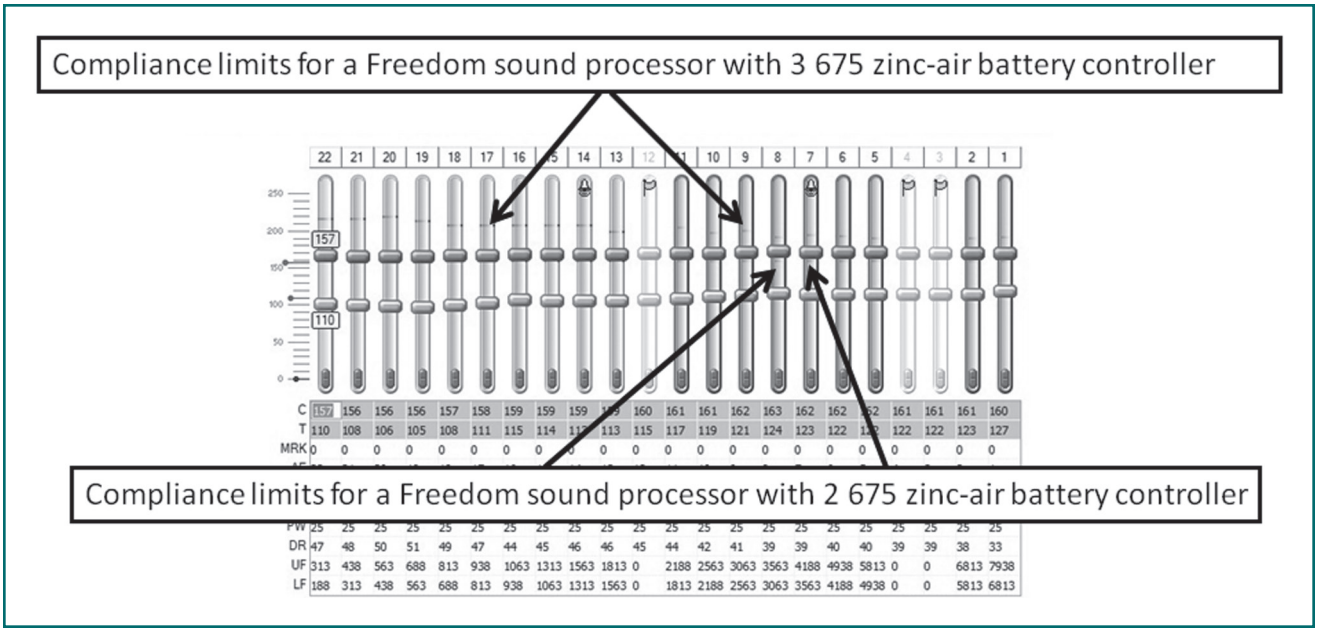



**FIGURE 2–14.** Voltage compliance limits with the Freedom processor using two different battery modules. *Note.* Provided courtesy of Cochlear Ltd.

Typically, the mixing-ratio parameter allows the clinician to choose from a variety of ratios ranging in a graded manner from an equal emphasis on signals from the sound processor microphone and DAI (i.e., no attenuation of signal from sound processor microphone) to DAI only (i.e., complete attenuation of signal from sound processor microphone). The mixing-ratio parameter also may include the telecoil-mixing ratio, which controls the strength of the signals from the telecoil and sound processor microphone. It also ranges from equal emphasis between the telecoil and microphone signals to telecoil-only emphasis. Additional information on adjustment of the mixing ratio is provided in Chapters 4, 5, and 9.

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## BASIC COCHLEAR IMPLANT SIGNAL CODING STRATEGIES

A signal coding strategy describes the algorithm used to transform the important features of the incoming acoustical signal (i.e., amplitude, frequency, and temporal cues) into an electrical code. This code attempts to represent these features in a meaningful manner to the auditory nerve. Although relatively large differences exist in the default signal coding strategies used by recent cochlear implant systems, clinical trials demonstrate comparable performance across the systems of the three different cochlear implant manufacturers. The following section describes contemporary cochlear implant signal coding strategies.

### Continuous Interleaved Sampling (CIS)

The Continuous Interleaved Sampling (CIS) strategy is available in the cochlear implant systems for all three manufacturers, and it served as a precursor for most of the current signal coding strategies. As illustrated in Figure 2–15, the acoustic signal is sent through a bank of bandpass filters that separates the input signal into discrete frequency bands. (Note: In contemporary systems, filtering is accomplished through digital signal processing rather than physical analog filtering.) The tonotopic organization created by the filtering is preserved throughout signal processing and eventually represented to the auditory system via presentation to electrodes located in different positions in the cochlea (i.e., low-frequency filter outputs are eventually delivered to electrode

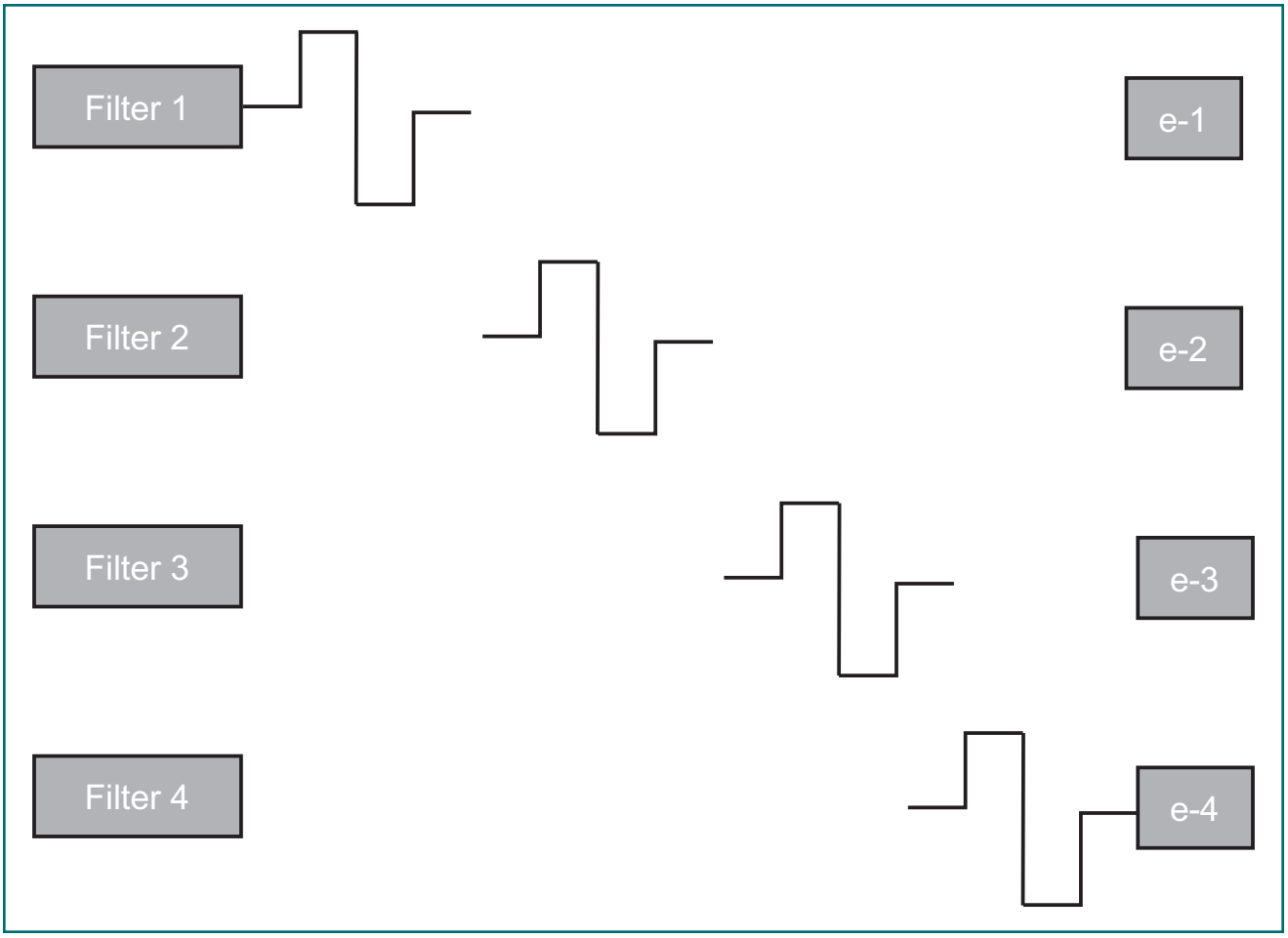



**FIGURE 2–15.** Illustration of the Continuous Interleaved Sampling (CIS) strategy. The acoustic signal sent through a bank of bandpass filters.

contacts located in the apical end of the cochlea, whereas high-frequency filter outputs are delivered to electrode contacts located at the basal end of the cochlea).

The bandwidth of the filters used in CIS varies across the three different manufacturers. The output from these filters is directed to a rectifier that converts the alternating current into a direct current signal. Following the rectifier, the signal is sent through a low-pass filter, where it is transformed into a temporal envelope. This envelope is analogous to the amplitude envelope of the acoustic input.

The CIS pulse rate must ideally be four to five times higher than the output from the low-pass filters to prevent aliasing errors. Because early versions of CIS-type strategies used pulse rates that were less than 2,000 pps, the cutoff frequency of the low-pass filter usually is set between 200 and 400 Hz. A cutoff frequency up to 400 Hz encompasses the frequency at which the nerve can code changes in electrical pulse rate. This cutoff frequency also includes the fundamental frequency of the human voice, which is as low as 125 Hz for males and up to 400 Hz for young children.

The output of the low-pass filters is sent to a compressor that uses a logarithmic function to map the input signal into the patient's electrical dynamic range. This output from the compressor then is sent to a pulse generator, which is set to a fixed frequency (e.g., often between 800 to 1600 pps). The amplitude of the pulse train from the generator is modulated on the basis of the input received from the compressor. Thus, the amplitude of the signal within each band is represented by the amplitude of the pulses within that

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same band. Finally, the modulated pulse trains in each channel are delivered to their respective electrode contacts.

Therefore, in the typical case, the intense low-frequency components of a speech signal are represented by high-amplitude pulses delivered to apical electrodes, whereas the relatively low-level, high-frequency components are represented by low-amplitude pulses delivered to the basal electrodes. All electrodes are stimulated sequentially during each stimulation cycle, and the amplitude of stimulation during each cycle corresponds to the energy present in the respective channel. Again, the frequency of the pulse trains is fixed, and stimulation occurs continuously at that stimulation frequency. Typical CIS pulse rates range from about 800 to 1,600 pps and usually are implemented with 8 to 16 channels.

Advanced Bionics offers a variation of CIS known as *Multiple Pulsatile Sampler* (MPS). MPS provides partially simultaneous stimulation. Specifically, two electrodes that are remotely spaced from one another are stimulated at the same time (i.e., electrodes 1 and 5; then 2 and 6; then 3 and 7; then 4 and 8). This approach allows for a doubling in the stimulation rate (e.g., the stimulation rate increases from almost 800 pps to about 1,600 pps), which should theoretically improve the provision of fine temporal structural cues.

The MED-EL Corporation offers additional variations of CIS known as CIS+ and High Definition CIS (HDCIS). CIS+ operates similarly to conventional CIS, but the frequency is expanded. Also, Hilbert transformation is used in place of wave rectification and low-pass filtering to more accurately estimate the input signal. In HDCIS, sequential stimulation between neighboring stimulation sites also is used to elicit a locus of stimulation somewhere between the two independent stimulation sites (i.e., virtual channels via current steering). This approach is used in an attempt to provide better spectral resolution.

## HiResolution (HiRes) Sound Processing

HiResolution sound processing, developed by Advanced Bionics and commercially released in the United States in 2003, is another variation of CIS processing. The primary differences between HiResolution and conventional CIS include: (1) the provision of 16 active electrodes in HiRes rather than the typical 8 in CIS, (2) considerably higher maximum stimulation rates (e.g., up to 5,156 pps), (3) higher cutoff frequencies for low-pass filters (e.g., 2800 Hz), and (4) a more sophisticated AGC system. Theoretically, the combination of the higher stimulation rates and filter cutoffs will improve the provision of fine temporal structure. An additional advantage may be the facilitation of stochastic firing in the auditory nerve that would more closely represent normal auditory nerve function.

HiResolution sound processing is available in two commercial forms, HiResS and HiResP. HiResS provides completely sequential stimulation, whereas HiResP provides partial simultaneous stimulation similar to the MPS strategy. In other words, HiResP provides simultaneous stimulation at 1 and 9, 2 and 10, 3 and 11, 4 and 12, 5 and 13, 6 and 14, 7 and 15, as well as 8 and 16. HiResS can deliver pulse trains with rates up to 2,900 pps, while HiResP has a maximum stimulation rate of 5,156 pps. In a clinical trial, speech-recognition performance of adult recipients was significantly better with the HiRes processing when compared to conventional strategies (Koch et al., 2004). In addition, the HiRes processing was preferred by the majority of subjects.

As previously mentioned, Advanced Bionics recently released a new version of HiRes known as HiRes Fidelity 120, which incorporates current steering. Current steering attempts to increase the number of perceptual channels in the frequency domain by simultaneously stimulating on two neighboring electrode contacts to create a locus of stimulation that falls somewhere between those two contacts (see Figure 2–8 for a visual illustration of current steering). Through the use of current steering, HiRes Fidelity 120 creates up to 120 virtual channels. Additional channels may increase the number of frequency percepts and spectral resolution and, therefore, should improve speech recognition in noise, sound quality, and music appreciation. Results of a recent clinical trial show better speech recognition, superior sound quality ratings, and higher music appreciation when adults with Harmony processors were

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using the HiRes Fidelity 120 strategy compared to the HiRes strategy (Advanced Bionics Corporation, 2009).

HiRes Optima is the newest strategy in the HiResolution sound processing family. The primary objective of HiRes Optima is to reduce the power requirements for the processor. As previously stated, the Naída CI Q70 processor is substantially smaller than the Harmony sound processor. The reduction in processor size required a reduction in battery size. Specifically, the zinc-air battery module must operate from two 675P zinc-air batteries.

HiRes Optima incorporates a couple of changes from HiRes Fidelity 120 that are all designed to reduce battery consumption while preserving recipient performance. HiRes Optima reduces power requirements by reducing the compliance voltage of the implant from 8 volts to 4 volts, by modifying the current steering paradigm so that stimulation is delivered only via virtual electrodes, and by modifying the pulse width management scheme to maintain the 4-volt compliance limit. In all other aspects (front-end options, frequency analysis, ClearVoice), HiRes Optima is identical to HiRes Fidelity 120 sound processing. A clinical trial conducted in 2012 showed that speech perception benefit was comparable between HiRes Optima and HiRes Fidelity 120 and that average battery life was improved by 53% (Advanced Bionics, 2012). It should be noted that for some recipients with high impedance levels, the pulse width required to achieve adequate loudness may increase and cause a subsequent reduction in the stimulation rate. However, no detriment in benefit is expected to result. HiRes Optima is available for the Naída CI Q70, Neptune, and Harmony sound processors.

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## *n*-of-*m* Strategies

The common denominator among all of the aforementioned CIS-type strategies is that all active electrode contacts are stimulated (either in a sequential or partially simultaneous manner) for each cycle of stimulation. The *n*-of-*m* signal coding strategies utilize a different approach. For a given input sound, the acoustic energy present in each of the *m* channels is determined, and stimulation is administered to only the *n* channels with the highest amplitude inputs (Figure 2–16). The *n* is typically referred to as maxima in these types of signal coding strategies. The *n* (or maxima) typically varies from 8 to 12. Therefore, a program with 22 channels and a maxima of 10 will result in a train of biphasic pulses delivered to the 10 channels with the highest amplitude. The remaining 12 channels are not stimulated during that specific cycle.

The *n*-of-*m* strategies are designed to capture the prominent components present in a speech signal while discarding the relatively low fluctuations in amplitude, which are more likely to be noise. The reduction in the number of active electrodes during each cycle allows for (1) faster stimulation rates, (2) a reduction in channel interaction and masking, and (3) an increase in battery life. The potential limitation of *n*-of-*m* approaches is that desirable information in some channels may not possess the necessary amplitude to be selected. As a result, the user will not have access to this acoustic information. However, research studies conducted with Cochlear Corporation users indicate that they do as well or better with an *n*-of-*m* strategy (i.e., ACE) as compared to CIS strategies (Kiefer, Hohl, Sturzebecher, Pfennigdorff, & Gstöettner, 2001; Skinner et al., 2002).

## Spectral Peak (SPEAK)

SPEAK was one of the first clinically available signal coding strategies that used the *n*-of-*m* approach. The initial components are similar to that of CIS strategies. Inputs are sent through a bank of bandpass filters, in which the complex broadband sounds are divided into narrower frequency bands. The amplitude of the inputs in each channel is determined, and the *n* channels with the highest amplitudes are selected for processing (i.e., a maxima of 8 is selected, then 8 of the 22 active electrodes are stimulated). The outputs from those filters then are rectified and low-pass filtered with the cutoff frequency of 200 Hz. Finally, those outputs are used to modulate a train of biphasic pulses that are fixed in frequency (typically 250 pps) and are delivered to corresponding electrodes. In practice, the SPEAK strategy

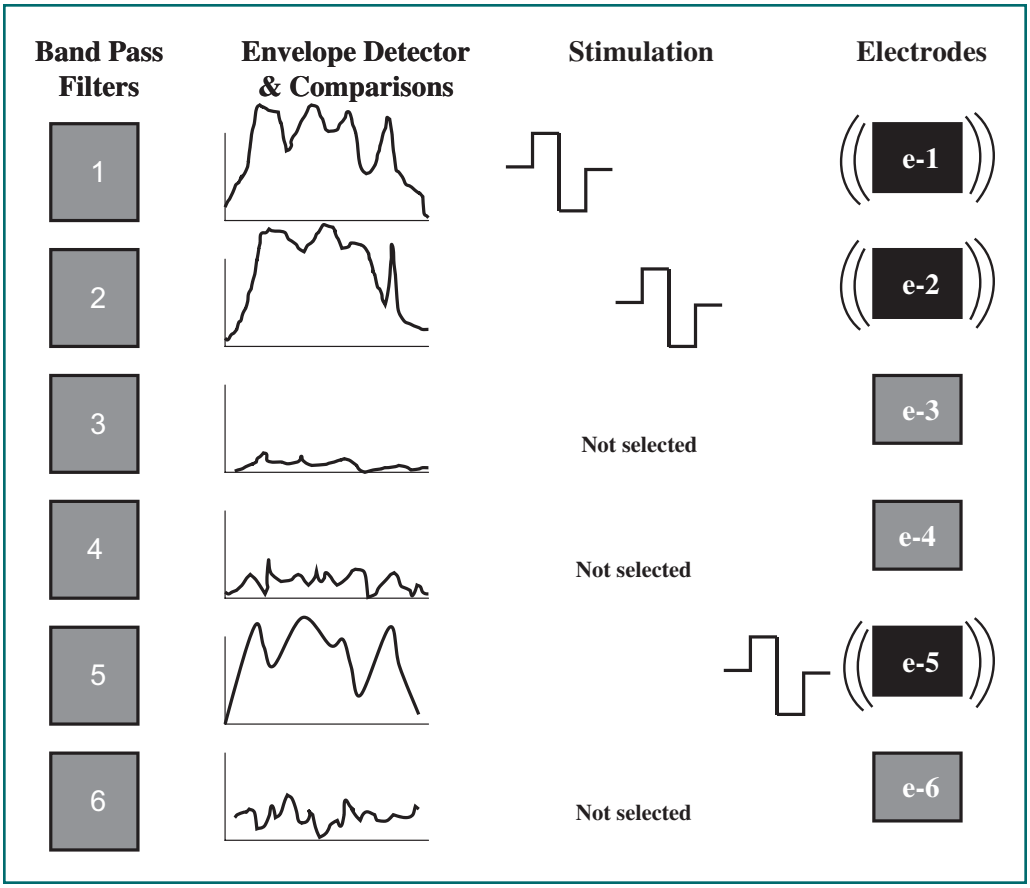



**FIGURE 2–16.** Block diagram of the *n*-of-*m* signal coding strategy (i.e., ACE, SPEAK). The three channels with the greatest amplitude envelopes were selected for sequential stimulation.

is implemented with bipolar electrode coupling in the previous generation Nucleus 22 implant and either bipolar or monopolar coupling in the Nucleus 24 and later implants. A maximum of 20 active channels are available, and the typical maxima is set at 8.

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## Advanced Combination Encoder (ACE)

ACE is also an *n*-of-*m* strategy that focuses on providing fine temporal structure to the user. It is similar to SPEAK, but it incorporates higher stimulation rates. ACE is the current default signal coding strategy used with Cochlear Ltd. implants. The first version of ACE allowed for a total stimulation rate of 14,400 pps. Consequently, the per channel stimulation rate is dependent upon the number of maxima selected. If 8 maxima is selected, then a per channel rate of 1,800 pps can be achieved (8 × 1,800 = 14,400). In practice, ACE uses monopolar electrode coupling. A maximum of 22 channels are available, and the typical maxima ranges from 8 to 12. In the current version of Custom Sound programming software, the ACE signal coding strategy is implemented with a default stimulation rate of 900 pps and 8 maxima. The authors of this book routinely use the ACE signal coding strategy with Nucleus recipients, but the maxima is typically increased to 10.

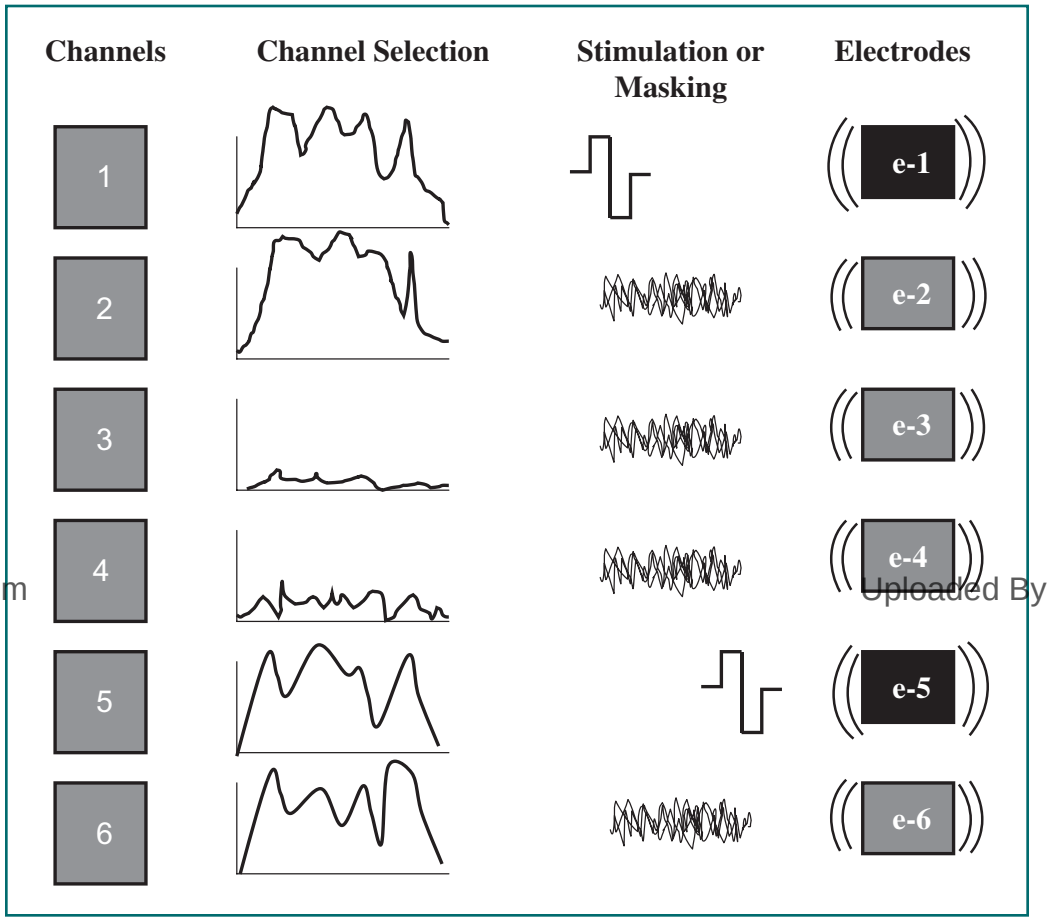
Research studies have shown that most recipients perform better with ACE than SPEAK (Kiefer et al., 2001; Skinner et al., 2002). ACE likely allows for better performance because of the provision of fine temporal structure cues via the higher stimulation rates of ACE. As a result, an overwhelm-

ing number of Nucleus recipients use ACE. The clinician may wish to switch to the SPEAK signal coding strategy if the recipient experiences undesirable effects with monopolar stimulation (e.g., facial nerve stimulation), and bipolar or pseudomonopolar stimulation is required.

Cochlear also offers a variation of ACE known as ACE(RE) or HighACE. ACE(RE) operates exactly like ACE, but the maximum total stimulation rate is 32,000 pps rather than 14,400 pps. The advantages of ACE(RE) over ACE may be negligible for many users because, as mentioned earlier, most recipients do not perform better with stimulation rates exceeding approximately 2,000 pps. In fact, in a large clinical trial, many subjects preferred and performed better with programs using rates at or below 900 pps (Arora et al., 2009; Balkany et al., 2007). It should be noted that the higher stimulation rates associated with ACE(RE) (particularly with the 3,500 pps stimulation rate) require a very narrow pulse width, and as a result, voltage compliance limits are often reached before an adequate loudness percept is obtained.

MP3000 is yet another Cochlear Corporation variation of an *n*-of-*m* strategy. Similar to the approach used in modern MP3 recreational audio players, unimportant information in the input signal (i.e., low-level components that will be masked by adjacent higher level and more important components) is discarded. As a result, the signal is conveyed in a more efficient manner and without a significant compromise in quality or clarity.

Figure 2–17 provides a simplified illustration of the MP3000 strategy in operation. Across these 6 channels, only channels 1 and 5 will be audible for the user because the neighboring channels will be masked. As a result, stimulation is provided to only these two channels.




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**FIGURE 2–17.** Illustration of MP3000 coding strategy.

The primary advantage of the MP3000 strategy is an improvement in signal efficiency, which allows (proportional to the number in maxima reduction) for longer battery life, reduction in battery size, and ultimately, smaller sound processors. Preliminary research findings suggest that speech recognition with the MP3000 strategy is better than that obtained with the ACE strategy (Buchner, Nogueira, Edler, Battmer, & Lenarz, 2008). The MP3000 signal coding strategy has been used in Europe but has yet to receive FDA approval for use in the United States.

## Fine Structure Processing (FSP)

MED-EL also offers another novel CIS-type signal coding strategy referred to as FSP, which has two important differences to the original CIS strategy. First, intermediate pitches are generated using bell-shaped, overlapping bandpass filters to provide better spectral resolution, which is important for the recognition of high-frequency phonemes including consonants identified on the basis of place cues (i.e., /s/, /f/, /t/). This form of current steering also may improve sound quality, speech recognition in noise, music appreciation, and music recognition.

The second difference from the traditional CIS strategy is that the FSP strategy modulates the timing of channel-specific sampling sequences (CSSS), or pulse bursts, in the lowest frequency channels, depending on the input frequency within those channels. In other words, the stimulation frequency present in low-frequency channels adaptively changes in order to be identical to the frequency of the input signal in the low-frequency channels. The conjectural advantage of this type of stimulation in the low-frequency channels is to provide better access to fine temporal structure by stimulating low-frequency auditory nerve fibers at the same rate as the low-frequency input signal. An improved provision of fine temporal structure may improve speech recognition, sound quality, recognition of vocal pitch, music appreciation, and music recognition. The FSP coding (i.e., CSSS analysis and adaptive provision of stimulation rate) is typically administered over the one to two most apical channels. Research on FSP is encouraging. Arnolder and colleagues (2007) report significantly improved speech and music perception with FSP relative to CIS, and most participants preferred the use of FSP over CIS.

At the time of this writing, MED-EL recently introduced a variation of the FSP strategy referred to as FS4. As previously discussed, the FSP strategy usually incorporates the CSSS analysis and adaptive provision of stimulation rate over one or two of the most apical channels. FS4 provides the fine structure processing on up to four of the most apical channels extending up to 1000 Hz. The primary objective is to enhance fine structure processing throughout the low-frequency range. Theoretically, the provision of fine structure is further facilitated by the deep placement of the MED-EL Standard electrode array into the apical area of the cochlea. This deep placement allows for stimulation of auditory nerve fibers that naturally possess a low-frequency characteristic frequency.

FS4-p is a variation of the FS4 signal coding strategy. With FS4-p, fine structure processing is provided up to 1,000 Hz, and parallel stimulation (simultaneous stimulation) is provided across two of those channels at a time in the event that two channels experienced a zero crossing since the last stimulation frame. The FS4-p strategy also possesses a novel algorithm referred to as Channel Interaction Compensation, which ensures that undesirable spatial channel interaction does not occur during the implementation of the parallel stimulation. At the time of this writing, there were no published studies examining the influence of FS4-p on speech recognition.

## Simultaneous Analog Stimulation (SAS)

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Simultaneous Analog Stimulation (SAS) is unique when compared to the aforementioned signal coding strategies. SAS stimulates electrode contacts with continuous electrical waveforms (rather than biphasic pulses) that are analogous to the acoustical inputs to each channel. The microphone converts the acoustic input into an electrical signal, and similar to most signal coding strategies, this signal is subjected to a bank of bandpass filters. The filters separate complex electrical signals into relatively narrow frequency bands. The electrical

output from each channel is subjected to compression based on a logarithmic mapping function, and then the compressed signal from each channel is sent to a dedicated electrode. Each electrode contact is stimulated at the same time (i.e., fully simultaneous stimulation) on each cycle of stimulation.

SAS has been used commercially only in Advanced Bionics cochlear implants, but it is no longer available in the current version of programming software. The typical SAS user will be an Advanced Bionics' recipient who received a cochlear implant prior to 2002. In its clinical implementation, SAS used seven to eight active electrode contacts with bipolar electrode coupling. To counter initial difficulties of avoiding voltage compliance issues, Advanced Bionics developed an internal device/electrode array that featured bipolar electrode pairs that were spaced at a greater distance than the original C1 device. The configuration of the new C1.2 device was referred to as enhanced bipolar electrode coupling.

The primary theoretical advantage of SAS is that it preserves most all of the cues present in the original input signal. The primary disadvantage is the fact that fully simultaneous stimulation likely results in a great deal of channel interaction, which may lead to distortion. Research studies examining performance differences between SAS and CIS produced mixed results (Zwolan et al., 2001). In short, this study showed that some recipients preferred and achieved at least similar performance with SAS when compared to CIS. Subjects who performed well with SAS typically had favorable preimplant histories and lower stimulation needs. Interestingly, Zwolan et al. found that recipients with the HiFocus electrode array and electrode positioner, a special component inserted along with the electrode array to facilitate proximity of the electrode contacts to the modiolus, were more likely to perform well with SAS. This finding may be attributed to the lower stimulation needs and subsequent reduction in channel interaction due to the shorter distance between the stimulating contacts and target neural elements. The electrode positioner was removed voluntarily from the market by the manufacturer in 2002 because it was associated with a higher incidence of bacterial meningitis.

## KEY CONCEPTS

After reading this chapter, the reader should understand the following basic terminology and concepts:

- Basic terminology is similar across cochlear implant manufacturers, but some manufacturer-specific features and signal coding strategies do exist.
- Parameters that influence coding in the intensity domain include threshold and upper-stimulation levels, amplitude of current, pulse width, electrical dynamic range, IDR, sensitivity, channel gain, and volume control.
- Parameters that influence coding in the frequency domain include the number of electrode contacts and channels, current steering, frequency allocation, and stimulation rate.

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# 3

# Basic Principles of Programming

Although some differences in cochlear implant programming exist across the three manufacturers, many more similarities can be found. It is these parallel aspects of programming that serve as the foundation for creating a good program (e.g., MAP). This chapter addresses the basic principles of cochlear implant programming that are necessary to provide a program/MAP that will optimize auditory, speech, and communication abilities as well as user satisfaction.

## PREACTIVATION PROCEDURES

### Realistic Expectations Prior to Activation

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Helping the patient and family to establish realistic expectations of cochlear implant benefit and performance is one of the most important objectives prior to implant activation. A cochlear implant offers an auditory experience that is quite different from a hearing aid, and as a result, most recipients require several days or even months of implant use before they are able to develop satisfactory open-set speech recognition and sound quality. Unfortunately, no matter how thoroughly the clinician discusses realistic expectations, recipients and families often are discouraged with performance during the first few days or even weeks of use. This is especially true for teenagers and adults who have long periods of deafness prior to implantation. However, some adults with residual hearing and families of young children with congenital deafness also are disappointed that the implant does not provide hearing at "normal levels."

Patient dissatisfaction with initial cochlear implant use occurs for several reasons. First, the signal from a cochlear implant is much different from that of a hearing aid. For example, many implants deliver the lowest frequency components through electrode contacts that are located in a place differing from the location for acoustic stimulation (i.e., several hundred Hz higher than the frequency of the input signal). As a result, the pitch of the human voice often is much higher than what the recipient is accustomed. Additionally, electrical pulsatile stimulation elicits an auditory nerve response that is much more synchronous than what is typical for acoustic stimuli. Furthermore, although the signal coding strategies used in contemporary implants are quite sophisticated and are the product of countless hours of research and development, the signal processing available within cochlear implants results in a stimulating

signal that is markedly different from the signal received from a normal peripheral auditory system in response to acoustic stimulation.

Second, a period of acclimatization is required for the auditory system to adjust to the new signal. Most studies show that performance of postlingually deafened adults typically improves throughout the first one to two years of cochlear implant use (Hamzavi, Baumgartner, Pok, Franz, & Gstoettner, 2003; Oh et al., 2003). This may be attributed to the fact that the auditory nervous system has not received stimulation for certain sounds (e.g., high-frequency sounds) for many years, and as a result, the auditory nervous system will lose some of its ability to effectively process these sounds. Also, most cochlear implant systems possess 12 to 22 active stimulating sites (i.e., intracochlear electrode contacts), which pales in comparison to the 3,500 inner hair cells that process spectral information in the normal cochlea. Consequently, the auditory nervous system most likely requires a sustained period of cochlear implant use in order to effectively process the impoverished signal provided by the implant.

Third, some recipients, such as young children and prelingually deafened adults, have little or no exposure to sound prior to implantation. These recipients require intensive exposure to speech and environmental sounds in order to lay a foundation within the auditory nervous system to allow processing of complex acoustic inputs. Thus, a period of auditory-focused speech and language therapy (e.g., Auditory Verbal therapy) normally is required before conversational, spoken-language skills emerge.

The preimplant history often determines the candidate's expectations as well as benefit and satisfaction with the implant. For example, recipients who suddenly lose hearing (e.g., ototoxic medications) have a much higher potential for immediate communicative benefit, open-set speech recognition, and successful telephone use than prelingually deafened adults. Adults who have prelingual deafness, a limited history of hearing aid use, and poor open-set auditory skills are still candidates for a cochlear implant, but they will need to have realistic expectations, such as improved awareness of environmental sounds and speech-reading abilities with little to no likelihood of the development of open-set word recognition.

Regardless of the history, counseling will help the patient understand that implant benefit will take time to manifest. Using a conservative counseling approach, audiologists should prepare adult recipients for limited word understanding and for the possibility that speech and environmental sounds most likely will initially possess a strange and poor sound quality. For preimplant counseling, the authors of this book subscribe to the approach of attempting to "under-promise and then over-deliver." In other words, we tend to conservatively counsel the implant candidate and his or her family with a deliberate intent to suggest that the recipient will not understand speech during the first few days or weeks of implant use and will likely be very dissatisfied with the initial sound quality of speech and environmental sounds. We even suggest that many recipients are unable to hear speech at all during the initial stages of implant use and that some recipients report that speech sounds like noise or ringing bells. In reality, we know that many adults are able to understand speech to at least some extent during the first few days of implant use. In cases in which recipients can hear speech or when recipients first begin to hear voices, they often describe the speech as possessing a robotic or mechanical quality. Many recipients report that speech is distorted in pitch (i.e., it sounds like Mickey Mouse, Donald Duck, Darth Vader, or like someone is speaking after inhaling the gas from a helium-filled balloon). The authors of this book believe that it is important to inform implant candidates and new recipients that many recipients will initially regret their decision to receive a cochlear implant, but after several weeks or months of implant use, they adapt to the new signal they receive from their implant and perform much better than they did with hearing aids. Most specifically, it is imperative to stress that time is required to adjust to the implant and to realize it full benefit.

Additionally, the authors describe, in general terms, the differences in electric and acoustic hearing as well as the physiology and patho-physiology of the normal and impaired peripheral and central auditory systems. The relevance of this information to implant outcomes is discussed. We believe

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that this discussion is especially important in light of the impressive advances that exist in other areas of medicine and health care. For example, ophthalmologists may use laser surgery to restore 20/20 (i.e., normal) vision for patients who are nearsighted. The general public is generally very aware of this medical treatment, and many patients erroneously assume that cochlear implantation should provide a similarly expeditious and effective correction for hearing loss. It is helpful to discuss the anatomical and physiological differences between common visual and hearing impairment to aid the candidate in understanding the differences experienced in outcomes received with a cochlear implant versus laser eye surgery. At this point, we often stress that a cochlear implant is not a substitute for normal hearing.

Furthermore, the authors of this book note that there are no guarantees associated with cochlear implantation. We note that although many recipients are able to converse over the mobile telephone, follow speech on the television without closed caption, and converse in real-world situations (i.e., restaurants), there are some recipients who never acquire these abilities, despite the best intentions and efforts of everyone involved in the process. This discussion is important because we have previously served recipients who have expressed that they understood that an adjustment process was necessary following cochlear implantation, but that they also were under the impression that they would eventually regain the ability to converse on the telephone, understand speech in noisy restaurants, and/or use the television without closed caption. Again, most postlingually deafened adult recipients and prelingually deafened children who receive implants at an early age do succeed in the aforementioned situations. It is simply important to note that these situations tend to be the most challenging for implant users and may continue to be problematic for the candidate under counsel.

Furthermore, the authors of this book often take the opportunity to discuss the use of hearing assistance technology (HAT) (e.g., personal digital RF/FM system) with cochlear implant sound processors. We have found that cochlear implant recipients are generally much more receptive to the use of HAT if the topic is discussed prior to implantation. Doing so also implicitly reiterates the fact that the cochlear implant alone may not sufficiently address the candidate's communication difficulties across every conceivable situation.

The implant manufacturers provide demo kits that include sample or "dummy" examples of their contemporary cochlear implant hardware (e.g., cochlear implant, sound processor, batteries, remote control). It is important to show this equipment to the candidate and his or her family and to discuss the advantages and limitations (e.g., size, ease of use, battery life, water resistance, HAT compatibility, signal processing, telephone use) of each cochlear implant systems offered by the cochlear implant program. This provides the recipient with the opportunity to handle the equipment, ask questions about the equipment, and determine the system (along with accessory equipment and processor color) that best fits his or her needs. Although this may seem like a trivial step in the process, it should be noted that some implant candidates select a particular implant make and model on the basis of a fairly practical or mundane consideration, such as water resistance, battery type/life, processor size/color, and so forth.

The authors of this book also review the various risks and potential negative consequences associated with cochlear implantation. Although medical risks will be covered by the cochlear implant surgeon, it is also important for the audiologist to discuss the surgical risks and considerations associated with implantation. We also discuss the increased risk of bacterial meningitis associated with cochlear implantation and the need to stay current with vaccinations for pneumococcal meningitis.

In short, the goal of preimplant counseling is to establish a relatively conservative or limited outlook pertaining to the benefit and performance a recipient will receive with the cochlear implant, while still convincing the candidate that cochlear implantation is in his or her best interests. In addition to the verbal counseling that is provided for the cochlear implant candidate and his or her family, the clinician should also provide written materials to the family so that they may review a tangible description of the information that has been provided (Figure 3–1). It should be noted

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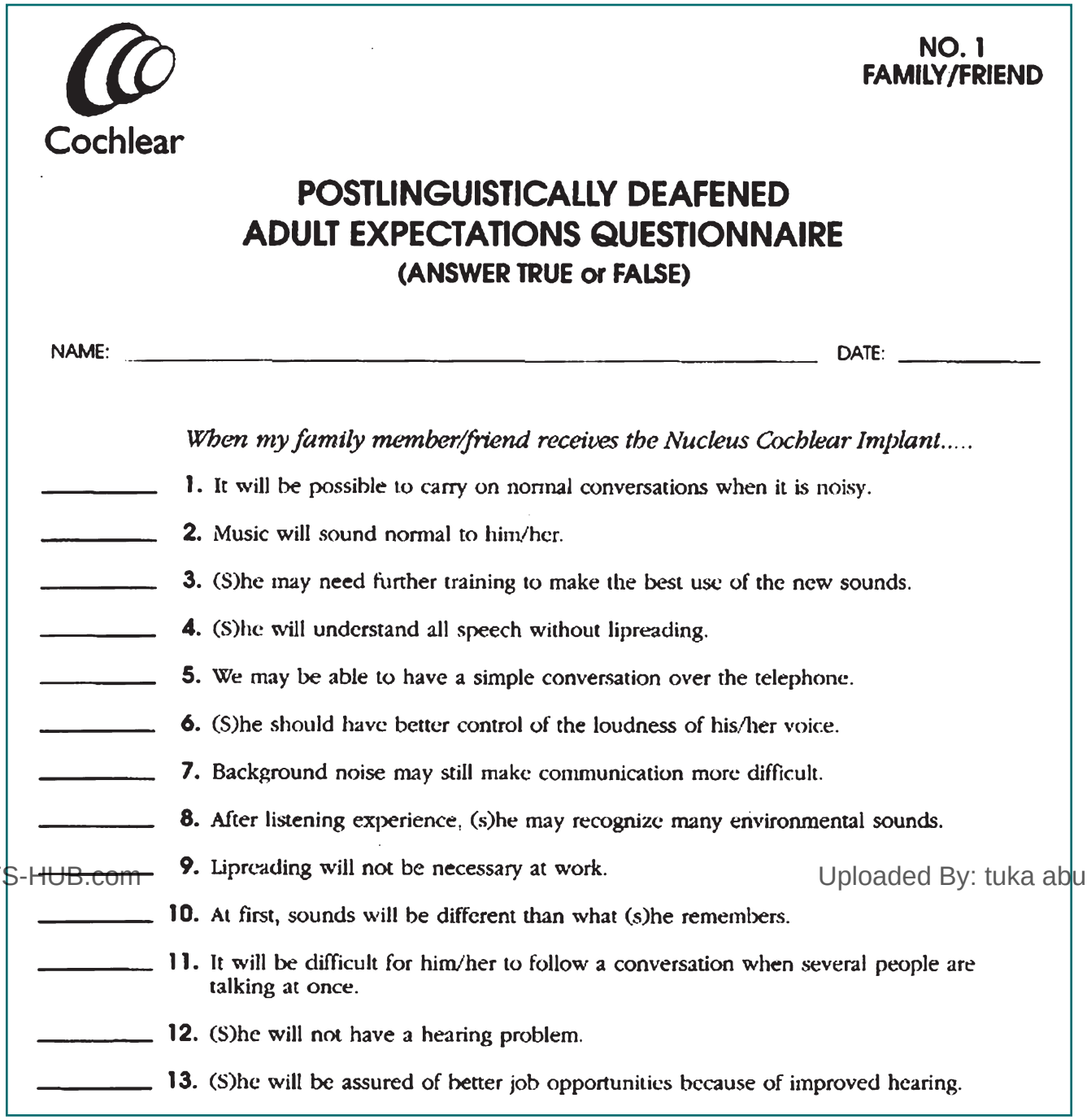
| | |
|---|---|
| **GENERAL INFORMATION:**<br>-General surgical considerations/risks<br>-Audiologic and rehabilitation/habilitation schedule<br>-Financial aspects: cost for surgery, audiologic and rehabilitative services, equipment, batteries, etc.<br>-Contraindications following surgery: MRI precautions, contact sports, etc.<br>-Warranty policies; reliability of internal/ external equipment<br>-Likelihood residual hearing eliminated by implant, potential increased risk for meningitis | **ADULT: POST-LINGUAL DEAFNESS:**<br>-If CI team recommended implant, highly likely communication ability with CI better than with HA<br>-Substantial variability in outcomes, outcome cannot be guaranteed<br>-Most score ≥ 80% sentence recognition<br>-May not be pleased with sound quality during the first few days of use; takes some time to adjust<br>-Performance often improves in first 6 mo or so<br>-Most are able to converse over the phone<br>-Likely more trouble than those with normal hearing in difficult environments (e.g., noise)<br>-Self-motivation is key! |
| **ELDERLY > 85 YRS; DURATION OF DEAFNESS:**<br>-Outcomes with CI typically better than with HA<br>-Variable outcomes, so may not be determined easily<br>-Understanding on telephone and for TV will often improve, but may still be difficult<br>-May need longer adjustment period; reach full potential at 6 mo to ≥ 1 yr<br>-Full-time use and frequent exposure to spoken language necessary to build auditory skill<br>-Post-implant aural rehabilitation is recommended for optimization of performance<br>-Self-motivation and positive attitude are strong keys to success | **CHILD IMPLANTED < 4 YEARS:**<br>-Outcomes vary depending on child characteristics<br>-Many children develop good speech & language skills with intensive audition-based therapy<br>-At activation, child will have auditory skill of infant unless substantial residual hearing for some time<br>-Post-implant, receptive language increase 3 mo-2 yrs; expressive language observed 18-30 mo<br>-Post-activation, sounds have little meaning and auditory responses may be limited; time is needed for auditory skills to develop<br>-Academic Considerations: FM/RF in the classroom |
| **CHILD PRE-LINGUAL DEAFNESS; CI 4-7 YRS:**<br>-If CI team recommended implant, highly likely communication ability better with CI than with HA<br>-Outcomes variable and dependent on many factors<br>-Typically poorer outcomes than those implanted earlier; should improve sound detection<br>-Some develop speech and language skills, but likely delayed; others develop limited skills | **OLDER CHILD > 7 YRS & LIMITED SPEECH; ADULTS WITH PRELINGUAL DEAFNESS:**<br>-Spoken language will always be limited compared to other populations of CI users<br>-Likely that telephone use will always be difficult<br>-Sound detection should improve<br>-Initial auditory reaction may be unfavorable |

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**FIGURE 3–1.** Presurgery checklist for clinician to discuss with recipient about realistic expectations, typical early cochlear implant outcomes, and experiences over the first year postsurgery.

that the implant manufacturers provide written materials that discuss some of the aforementioned information pertaining to cochlear implantation, and this information may also be found on the manufacturer's websites.

In addition, the audiologist may consider using a questionnaire, such as the measure developed by Cochlear Americas in Figure 3–2, to assess preimplant expectations and aid in counseling. The authors of this book have occasionally



Cochlear

NO. 1
FAMILY/FRIEND

**POSTLINGUISTICALLY DEAFENED ADULT EXPECTATIONS QUESTIONNAIRE**
**(ANSWER TRUE or FALSE)**

NAME: _____ DATE: _____

*When my family member/friend receives the Nucleus Cochlear Implant.....*

_____ **1.** It will be possible to carry on normal conversations when it is noisy.

_____ **2.** Music will sound normal to him/her.

_____ **3.** (S)he may need further training to make the best use of the new sounds.

_____ **4.** (S)he will understand all speech without lipreading.

_____ **5.** We may be able to have a simple conversation over the telephone.

_____ **6.** (S)he should have better control of the loudness of his/her voice.

_____ **7.** Background noise may still make communication more difficult.

_____ **8.** After listening experience, (s)he may recognize many environmental sounds.

_____ **9.** Lipreading will not be necessary at work.

_____ **10.** At first, sounds will be different than what (s)he remembers.

_____ **11.** It will be difficult for him/her to follow a conversation when several people are talking at once.

_____ **12.** (S)he will not have a hearing problem.

_____ **13.** (S)he will be assured of better job opportunities because of improved hearing.

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**FIGURE 3–2.** Sample questions from a Cochlear Americas questionnaire to assess a patient's preimplant expectations. *Note.* Provided courtesy of Cochlear Ltd., © 2003 Cochlear™ Ltd.

counseled a recipient about realistic expectations in great detail and later determined that the recipient was still misled about the outcome most likely to be obtained from cochlear implantation. The questionnaire shown in Figure 3–2 is one of the most effective means to determine exactly where a candidate's expectations reside, which provides the clinician with an opportunity to provide additional counseling as needed.

When counseling families of pediatric recipients, parents and caregivers must understand that experience and maturation are required before substantial gains are seen in a young child's communicative abilities. For example, it may take more than a year for young children without additional disabilities to develop spoken language vocabulary of even a few words. Unfortunately, approximately 40% of children with hearing loss have additional disabilities that may hinder speech and language development. Furthermore, children who are implanted at an early age (i.e., ≤1 year) may have cognitive or neurological impairments that have not been identified yet but will influence spoken language outcomes.

To prepare families of pediatric recipients and put them at ease, audiologists need to discuss all of the potential responses following activation including sound awareness, anxiety, happiness, fear, and indifference. Most children do not display a strong aversion to their initial experience of hearing with their cochlear implant, particularly when the clinician introduces stimulation appropriately. In fact, some are very happy with the new sound and will smile and point to their ear when they hear programming stimuli or external sounds. However, in some cases, children do become upset, cry, and seek comfort from caregivers. To prevent this negative reaction, the audiologist should make every attempt to avoid overstimulation. Even with careful programming, however, a few children, especially those with minimal experience with sound, still become upset when they first hear through the implant. This is most likely due to the child being startled by the new signal he or she receives.

Finally, some children initially may be ambivalent to the sound they receive. This response will often be concerning to the child's family because they question whether the device is working. Typically, the clinician can conduct measures, such as the electrically evoked compound action potential, that indicate that the auditory system is responding to the stimulation from the implant. Of course, this information can be conveyed to the child's caregivers to alleviate their concerns. All of the aforementioned responses are possible for a young child receiving a cochlear implant, even when he or she eventually develops considerable conversational and auditory skills. It is important that the family is prepared for each of these responses, so they are not overly alarmed or disappointed on activation day.

Prior to implantation, it is beneficial to provide materials that help preschool and elementary age children to understand the cochlear implant process. Manufacturers offer books for young children that describe the process from surgery to activation in kid-friendly language (Figure 3–3). Also, some families create photo albums depicting their own experiences and share them with other families beginning the journey of cochlear implantation (Figure 3–4).

Additionally, it is imperative that the cochlear implant team counsel a child's family about the importance of full-time cochlear implant use and intensive therapy focused on the development of auditory and spoken language abilities. Numerous studies have shown that optimal cochlear implant outcomes are dependent on the family's commitment to full-time implant use and audition-based therapy (Ching & Dillon, 2013; Ching et al., 2013; Moog & Geers, 2003). Of course, the mode of communication a family uses after cochlear implantation is entirely the choice of the family. The clinician should assist the family in ensuring that sufficient habilitative therapy is in place to support the child's development after implantation. However, it is also the clinician's responsibility to adequately educate the family about the different factors that influence the outcomes children achieve with their cochlear implants. Research has established that better auditory and spoken language outcomes are achieved when families commit to Auditory Verbal/auditory oral based therapy as compared to a Total Communication or manual-based approach (Dettman, Wall, Constantinescu, & Dowell, 2013; Jiménez, Pino, & Herruzo, 2008; Moog & Geers,

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**FIGURE 3–3.** Example of one children's book that describes the process from surgery to activation. *Note.* Provided courtesy of MED-EL.

2003). Families should be aware of this research when determining the type of intervention that is best for their family and child.

## Reviewing Logistics of Cochlear Implantation

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The audiologist should also discuss the schedule of audiological, medical, and rehabilitative appointments that will be necessary for the patient before and following cochlear implantation. This discussion will prepare the family for the time and financial commitments associated with pre- and postoperative appointments. The majority of medical and audiological appointments are during the first year (i.e., diagnostic testing, programming). After the first year, medical and audiology appointments are fairly infrequent. As previously discussed, auditory-based habilitative/rehabilitative appointments are essential (e.g., 1 hour per week) throughout the first few years of a child's implant use and are critically important for infants and young children. Because of limitations in reimbursement, trained personnel, and time, most adult recipients do not receive formal aural rehabilitation, but in reality, the vast majority of adults are likely to benefit substantially from postactivation aural rehabilitation. This rehabilitation is usually instrumental in building the confidence of the newly implanted adult. The focus of this rehabilitation should include orientation to the new recipient's cochlear implant equipment including the appropriate use of multiple programs and HAT to successfully address difficult situations. The authors of this book have found




**FIGURE 3–4.** Example of a page from one family's photo album.

that many adult recipients are reluctant to resume telephone use because it was previously very difficult with hearing aids. A primary goal of aural rehabilitation should be to facilitate the recipient's return to telephone use. Additionally, many long-term deafened cochlear implant recipients have developed maladaptive strategies to compensate for their hearing loss. Aural rehabilitation should assist the new recipient in identifying these strategies and eliminating them. Also, the therapist should identify consistent error patterns in speech recognition and assist the recipient in ameliorating these errors. If the errors persist, then the therapist should report the details of the error(s) to the programming clinician so that programming attempts may be made to improve performance. A full discussion of the activities that comprise adult aural rehabilitation is beyond the scope of this book. However, the therapist or audiologist should be prepared to recommend and describe a variety of exercises that may be done at home to facilitate listening development with the cochlear implant. Table 3–1 provides a brief description of some activities that may be suggested for adult recipients. It should be noted that in most cases, postlingually adult recipients quickly adapt to the signal provided by their cochlear implant. As a result, the aural rehabilitation they receive will typically span over the course of a few weeks rather than years as is the case for newly implanted young children.

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**Table 3–1.** Activities to Promote Auditory-Skill Development in Adult Recipients

| *Activity* | *Description* |
|---|---|
| 1. Promote sound awareness | Recipient will respond to loud sounds (e.g., hand clap; drum; "ba, ba, ba") without visual cues |
| 2. Promote sound identification | Respond to name 95% of the time with no visual cues |
| 3. Discriminate basic tenants of sound | Discriminate sound differences, such as loud vs. soft; high/low pitch; continuous/abrupt sounds; slow/fast stimuli; cheerful/angry speech; male/female voice |
| 4. Detect all 6 Ling sounds | Recipient should be aware of Ling sounds: ah, ee, oo, m, sh, s |
| 5. Imitate all 6 Ling sounds | Ask recipient to repeat Ling sounds: ah, ee, oo, m, sh, s |
| 6. Discrimination of words with differing phonemes | Discriminate between words, such as hat, shoe, bed |
| 7. Discrimination of words with medial vowel differences | Discriminate between words such as, bat, bed, boat |
| 8. Auditory tracking | Ask recipient to follow along with words in a book or offer suggestions of books on tape |
| 9. Identify days of the week | Ask the recipient to tell you what day the clinician stops on as the clinician repeats the days (e.g., Monday, Tuesday, Wednesday . . . where did I stop?) |
| 10. Identify months of the year | Ask the recipient to tell you what month the clinician stops on as the clinician repeats the days (e.g., January, February, March, April . . . where did I stop?) |
| 11. Discriminate familiar phrase cards using phrases that differ in length | Ask the recipient to listen and select between two cards (e.g., "hi" vs. "How are you?") |
| 12. Discriminate familiar phrase cards using phrases with similar lengths | Ask the recipient to listen and select between two cards (e.g., "How old are you?" vs. "What is your name?") |
| 13. Discriminate words with same vowels, but consonants differ in place and voicing | Discriminate between words, such as bet, bed, beg |
| 14. Discriminate, identify, and comprehend open set conversation with known topic | Clinician selects topic familiar to recipient |
| 15. Discriminate, identify, and comprehend open-set conversation with unknown topic | Clinician selects topic |
| 16. Telephone practice | Recipient should practice talking on the phone to clinician, friends, and family |
| 17. Practice hearing order from a menu | Clinician should give the recipient his or her order from a menu |

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## Familiarity with Cochlear Implant Hardware

One consideration in cochlear implant management is to provide the patient with the external cochlear implant equipment prior to activation. This will give the adult, family, or child the opportunity to read the user manual and become acquainted with the equipment and accessories, which saves the clinician a considerable amount of time required to explain the basics of the external equipment. Also, it provides the recipient/family with an opportunity to identify questions pertaining to the equipment that they might like to address with the clinician.

This also will give children the chance to wear the processor without the provision of stimulation, which may put them at ease during and after the activation appointment. However, the incision site will need to be free of infection and irritation before placing the processor on the child's head. Also, some clinics recommend the use of a relatively weak magnet a day or two before the activation appointment rather than a highly magnetized transmitting coil to prevent irritation near the surgical site.

After establishing the aforementioned expectations of implantation (i.e., under-promising), the clinician should set forth to maximize the recipient's outcome (i.e., over-deliver). In reality, many adults do achieve some open-set speech understanding on the first day of cochlear implant use, and as a result, they are thrilled with the early success. Additionally, most children immediately exhibit more favorable responses with their cochlear implant than they possessed with their hearing aids. The following sections of this chapter describe the components the clinician should include in the cochlear implant programming process in order to maximize the potential and outcome of each cochlear implant recipient.

## Setting the Stage

Clinicians must set the stage for success on activation day and during all subsequent programming sessions. The first consideration is the use of appropriate décor and furniture: this step is especially important for young children. This action ensures optimal placement and seating for behavioral testing and successful reinforcement with toys, distracters, snacks, and other items needed to facilitate cooperation during the session. It is critical that the chair in which the recipient sits for programming is comfortable and provides adequate support. For infants, it is critical that the chair provides ample support for the child's torso and buttocks, so that the child is stabilized and able to execute a head turn in response to the programming stimulus (in order to facilitate visual reinforcement audiometry). For children of all ages, it is important to ensure that the child's legs and/or feet are well supported and stable. If this does not occur, then he or she may be inclined to kick or shuffle his or her feet throughout the session, which will diminish attention to the programming stimuli. Figure 3–5 provides examples of pediatric chairs and tables used during programming. Babies and young toddlers may be seated in a highchair during programming (Figure 3–6); however, the clinician should be aware that high-backed highchairs may interfere with the transmitting coil and knock it off the head. An effective solution is to place a back support between the baby and the highchair back to provide support and keep the head away from the back. Highchairs with straps are helpful because they prevent the child from moving around during the potentially lengthy programming session. Figure 3–7 shows examples of seating that are conducive for older toddlers and preschool-aged children. A child-sized picnic table is also an option for children who are able to sit still and is suitable for conditioning games, coloring books, play dough, and so forth, used during pediatric programming sessions. Adults should be provided with a comfortable chair with armrests so the recipient is able to attend throughout the programming session.

For adults and children, the audiologist should be careful to avoid giving cues about the programming signals that are presented to the patient. Children, in particular, are very adept at watching the audiologist's computer monitor, hands, facial expressions, body language, and so forth, in an attempt to determine when they should

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**FIGURE 3–5.** Examples of pediatric chair and table used during programming with fish bowl game used for programming.

respond to programming stimuli. Older children and adults are often seated at a 45° angle from the audiologist and computer to prevent them from seeing the monitor of the programming computer and to allow the clinician to observe responses. Also, the clinician should ensure that the recipient's family members are not providing cues to the user. It is not uncommon for the parent of a child to unintentionally change facial expressions or body language during the presentation of stimuli, so the programming clinician must ensure the child is not receiving augmentative cues about the programming stimuli presented during sessions.

In most cases, family members and friends should be welcome to attend programming sessions. However, if the clinician believes that the extended family and friends are preventing the recipient from focusing on the task at hand, he or she may need to request that the family members abstain from attending future appointments. Instead, a video of the session can be given to families to share with loved ones.

## PROGRAMMING AFTER IMPLANTATION

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### Physical Evaluation

On the day of activation, the clinician should inspect the incision site for signs of irritation or infection and perform otoscopy on both ears. If anything remarkable (e.g., otitis media) is observed, then the clinician should proceed with caution and alert the cochlear implant surgeon. Also, prior to every programming session, the




**FIGURE 3–6.** Highchair for use with babies and toddlers during programming.





**FIGURE 3–7.** Examples of seating for older toddlers and early elementary children during programming.

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clinician needs to visualize the site on which the transmitting coil rests to ensure that the magnet strength is not excessive. If a pronounced indentation and/or discoloration are present, the magnet strength should be reduced.

Determining the appropriate magnet strength is a subjective skill that the clinician may need to perfect over time. It is important to select a magnet strength that is sufficient to prevent the transmitting coil from repeatedly falling off of the recipient's head. However, it is likely even more important that the magnet strength is not too great because excessive adherence of the coil may compromise circulation to the skin underneath the coil. If excessive adherence secondary to excessive magnet strength persisted for a prolonged period of time (e.g., several weeks), it is possible that the skin under the coil may necrotize, which could result in a skin flap breakdown that would necessitate reimplantation. Such an event is very unlikely, but nonetheless, the clinician should exercise caution against providing a magnet strength that is too strong. In the case of older children and adults, the recipient will typically complain of a headache or a dull discomfort if the coil magnet strength is too strong. A greater concern exists for young children who cannot verbalize any discomfort that they may experience from excessive magnet strength. As a general rule of thumb, young children (6 months to 2 years old) and elderly women are often more likely to have very thin skin flaps and require a weaker magnet strength. In contrast, middle-aged males and obese recipients are more inclined to have thicker skin flaps and require a stronger magnet strength. Also as a general rule, when the clinician places the coil on the head, there should be some attraction, but there should not be a strong pull that forcefully draws the coil from the clinician's hand to the recipient's head. Further, the coil should not easily fall off when the recipient turns his or her head back and forth. However, it should most likely dislodge when the recipient brushes his or her hand against the coil.

It should be noted that the parents of newly implanted young children are often inclined to request excessive magnet strengths, because the coil tends to frequently fall off of the head of active toddlers. Again, the clinician should exercise caution when considering an increase in magnet strength and counsel the parents accordingly if a stronger magnet seems unwarranted. If the magnet never falls off of the head of an active toddler, then it is likely that the magnet is too strong. It should also be mentioned that higher magnet strength may be necessary at activation, because the swelling from the implant surgery may have yet to completely resolve. When swelling and discomfort are alleviated around the incision site, the clinician should physically inspect the area above the internal device to ensure the device is stabilized on the desired location. Once the swelling has subsided, the clinician should determine whether the magnet strength needs to be reduced.

It is standard practice for the surgeon to request a postoperative x-ray to confirm that the electrode array is inserted properly. Following this imaging and prior to activation, the programming clinician needs to confer with the cochlear implant surgeon about any intracochlear electrode contacts that are not fully inserted into the cochlea. In addition, it is important to know whether any electrodes are positioned unfavorably (e.g., insertion was shallow, deep, or typical). This information is essential because it is the only way that the programming clinician will know with complete certainty whether electrodes should be disabled. Finally, the surgeon should inform the programming audiologist of any other pertinent details that may influence programming.

## Selecting a Signal Coding Strategy

After the recipient is prepared for initial activation, the audiologist should begin by selecting the signal coding strategy that will be used in the recipient's sound processor. As described in Chapter 2, each manufacturer offers several signal coding strategies that vary in a relatively substantial manner. Chapters 4, 5, and 6 contain the authors' manufacturer-specific recommendations for which signal coding strategy and additional parameters the programming clinician should select for newly implanted recipients. Research has shown that some recipients perform better with one signal coding strategy over another (Pasanisi et al., 2002). However, the authors of this

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book tend to begin with the signal coding strategy that is recommended by the manufacturer. Most recipients perform very well with the manufacturer's recommended signal coding strategy. As a result, during the first month of implant use, the authors typically will only provide the recommended signal coding strategy to the newly implanted recipients.

At this time, it is prudent to note that one of the most important programming factors that influence a recipient's outcome and performance is the stimulation levels that are programmed by the clinician and used to stimulate the recipient's auditory nerve. As a result, the authors of this book contend that the clinician should initially focus on establishing the optimal stimulation levels for the recipient rather than adjusting secondary programming parameters or trialing various signal coding strategies.

If the recipient is struggling substantially after one month of use (i.e., no open-set speech recognition for adults who are postlingually deafened, complaints of poor sound quality, outcome much poorer than expected by the clinician) and the clinician is certain that stimulation levels are set appropriately, then the clinician should attempt to trial different stimulation rates within the recommended signal coding strategy. If difficulties persist, then the clinician should evaluate whether alternative signal coding strategies provide improved performance. If the recipient is uncertain of which signal coding strategy (or stimulation rate) provides the best performance, then the clinician may provide the recipient with at least two programs with different signal coding strategies (or stimulation rates) to allow the recipient the opportunity to experience each in real-world settings.

Young children, in particular, are typically unable to provide reliable feedback about perceptual differences between signal coding strategies. Therefore, as a first step, the manufacturer-recommended signal coding strategy should be used (see Chapters 4–6 for the authors' manufacturer-specific recommendations on signal coding strategies and programming parameters for use with children). If audiologic assessment, and more importantly, speech and language development are unfavorable, then a reasonable approach is to try an alternative signal coding strategy or stimulation rate. However, as previously discussed in the aforementioned paragraphs, a child's progress and performance with his or her cochlear implant is largely influenced by the stimulation levels that are programmed by the clinician. As a result, the authors of this book suggest that the clinician should primarily focus on establishing the most appropriate stimulation levels for a child rather than switching signal coding strategies or adjusting other MAPping parameters. The authors will attempt to change signal coding strategies and adjust other programming parameters (e.g., stimulation rate, IDR, maxima) only after it has been determined that the stimulation levels have been optimized and the child is continuing to struggle.

## Streamlined Versus Comprehensive Programming

In recent years, manufacturers have created signal coding strategies and programming platforms that expedite cochlear implant programming for two reasons. First, the number of recipients and potential candidates has increased substantially in the past decade, making it difficult for clinics to manage their expanding patient populations. Second, reimbursement for cochlear implant mapping is typically poor. Therefore, it is not financially feasible to spend a great deal of time with each recipient.

Historically, programming involved a comprehensive battery of measures to determine stimulation levels for each electrode contact in the array. In contrast, streamlined approaches involve measurements for only a few electrodes in the array. In the following paragraphs, basic programming procedures will be discussed for streamlined and comprehensive approaches.

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## Measuring Stimulation Levels

Researchers and clinical audiologists often give a great deal of attention to programming parameters including stimulation rate, input dynamic range, and frequency allocation. Although these parameters clearly can impact performance, the

most important aspect of programming is the optimization of the recipient's stimulus levels. The following section discusses the various procedures used to determine appropriate stimulation levels for adults and children.

### Setting Threshold Levels for Adults

Setting the threshold of electrical stimulation (i.e., T level) for an adult recipient is important because it determines the amount of stimulation necessary to restore audibility and normalize loudness for low-level sounds. In most cases, procedures for determining T levels are similar to those used to obtain routine audiometric thresholds and are relatively straightforward. However, as previously mentioned, research has suggested that many adults with contemporary Advanced Bionics and MED-EL cochlear implants experience good outcomes and have sufficient audibility for low-level sounds when T levels are set to zero or estimated on the basis of the upper-stimulation level (Spahr & Dorman, 2005). When T levels are set to a minimum amount of stimulation or at a certain percentage of the upper-stimulation level, the signal coding strategy uses a logarithmic function to map acoustic inputs toward the upper end of the electrical dynamic range in an attempt to ensure adequate audibility for low-level sounds. Also, these mapping functions are based on the typical electrical dynamic range of an implant user. As a result, they yield satisfactory results for the typical user, but the results may be less than ideal for recipients who have an electrical dynamic range that is markedly different from typical recipients.

T-level assessment often begins on a low-frequency channel because listeners with hearing loss are more likely to recognize a stimulus that was familiar prior to surgery (i.e., most individuals had some low-frequency hearing prior to surgery). The low-frequency stimuli should be presented at a level that is clearly audible, but not uncomfortable, to give recipients a reference for the stimulus they are trying to detect. After audibility is established, traditional threshold testing begins using an adaptive-bracketing procedure, such as the modified Hughson-Westlake (Carhart & Jerger, 1959). Recipients tend to respond to lower stimulation levels for descending presentations as compared to ascending presentations (i.e., a hysteresis effect). Therefore, to ensure audibility of soft acoustic inputs, T levels should be based on responses acquired in the ascending direction.

The step size for ascending and descending presentations varies. Assessment of precise T levels at the initial activation can be difficult for several reasons including (1) recipients are not accustomed to the signal from the implant, (2) it may have been a long time since a recipient has heard any kind of auditory signal at a particular frequency (or they may have never heard at a certain frequency), and (3) the recipient may be experiencing tinnitus or internal noise within the auditory system, which is difficult to discern from the programming stimulus. Therefore, at initial activation, it is likely that the recipient may respond only at suprathreshold levels. Because the user's T-level responses are likely to change over the first few weeks of cochlear implant use, it is not critically important to determine T levels with an exact precision during the first few days of implant use. As a result, the clinician should use larger step sizes for the first few appointments. After a week or more of implant use (and occasionally after just a day or two), the "internal noise" or tinnitus in the auditory system typically diminishes, and the adult recipient is much more adept at responding to signals at true T level. As a result, step size typically is decreased to provide a more precise measure of the electrical threshold.

Using the computer/processor interface, the audiologist presents a train of biphasic pulses to the external sound processor at a stimulation rate equal to what is used in the recipient's program. The audiologist asks the recipient to indicate whether he or she detects the stimulus by saying "yes" or raising a hand. The exact definition of T level varies across manufacturers, but as a general rule, T level is set to 50% threshold level (or just below) for CIS-type strategies and to the minimum amount of stimulation necessary for 100% detection for *n*-of-*m* strategies. As previously mentioned, T levels should be determined according to two responses obtained in an ascending presentation mode.

After establishing the T level on a given channel, the clinician should proceed to measure T levels for other electrodes across the electrode array.

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For young children, the authors recommend that the clinician "hop" across the electrode array in a staggered manner in an attempt to obtain T-level (i.e., minimum response level) measures at each segment of the electrode array. In other words, after measuring T level on an apical electrode, the clinician should measure T level on a basal electrode next and then measure T level on an electrode in the middle of the array. From that point, another apically located electrode may be measured following by another basally located electrode. Young children often have short attention spans and may abruptly decide that they no longer wish to participate in the T-level measurement process. If this occurs, the aforementioned staggered approach at T-level measurement will allow for a survey of T-level measurements across the frequency range. If the recipient fatigues during the measurement of T levels, the clinician may estimate T levels across the rest of the electrode array on the basis of measurements made at different points throughout the array (i.e., interpolation).

At follow-up programming appointments, the clinician should be fairly confident with the recipient's T levels based on previous programming sessions. Therefore, the clinician may begin testing below the expected threshold of audibility to improve efficiency and to avoid fatigue or loss of attention related to repeated presentation of the signal at suprathreshold levels. To prevent any confusion, however, an audible reference stimulus may be given prior to testing if the clinician feels that the recipient may benefit from a reference signal. Doing this may assist the recipient in being confident in the signal he or she is detecting. Additionally, at any point at which the clinician becomes concerned in the reliability of the recipient's responses (e.g., abundant false positives, uncertainty of responses, loss of attention), the authors recommend presenting the programming stimulus at a level that is clearly audible. This enhances the reliability of the recipient's response as well as the clinician's confidence in the recipient's responses.

T levels should be measured several times during the first year of cochlear implant use. Research has shown that T levels of adults increase slightly over the first two months of use and typically are stable from that point forward (Hughes et al., 2001). After an adult recipient has a stable cochlear implant program and is performing well, evaluation and programming should occur one to two times per year. The authors strongly recommend that children are seen for programming at least every three months until they are 7 years old. A suggested programming and evaluation schedule for adult and pediatric implant recipients is provided in Chapter 7.

Negative consequences may arise when T levels are not programmed appropriately. When set too low, the recipient will not have adequate audibility of low-level or soft sounds. When set too high (i.e., above the real threshold), the subject may hear too much ambient noise and the recipient's electrical dynamic range will be reduced unnecessarily. Reduction of an already limited electrical dynamic range will make it even harder for the audiologist to accommodate the wide range of typical acoustic inputs and desired loudness experiences. Also, when T levels are set too high in CIS-type strategies, the user may report a continuous noise that sounds like "bacon frying," "buzzing," "hissing," "humming," or "static."

In addition to traditional threshold measurement techniques (i.e., modified Hughson-Westlake), three other approaches can be taken for determining T levels: (1) the count-the-beep method, (2) psychophysical loudness scaling, and (3) threshold estimation. The count-the-beep method uses a gated stimulus and often is selected for recipients who have difficulty distinguishing between their tinnitus and a continuous programming stimulus. When using this approach, the patient is asked to count the number of programming stimuli that were arbitrarily selected and presented by the clinician (i.e., the clinician presents one, two, three, four, or five "beeps," and the recipient must determine the correct number). With this method, the T level typically corresponds to the stimulus level that results in 100% accuracy. Skinner, Holden, Holden, and Demorest (1995) showed that use of the count-the-beep method for setting T levels results in better access to low-level sounds when compared to programs created via the audiometric-style approach (i.e., Hughson-Westlake). The authors have successfully used this technique with recipients who pro-

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vide excessive false positive responses. It has also been used successfully with recipients who have less than ideal sound-field detection thresholds when T levels are based on conventional audiometric measurement.

The second alternative procedure for establishing T levels, psychophysical loudness scaling, determines the stimulation level corresponding to a soft-loudness percept. Using a chart for assessing loudness growth, the clinician gradually increases the presentation level starting at a point that is inaudible and eventually stops when the stimulus level is perceived as "loud." T level is set to the stimulus level that corresponds to a psychophysical rating of "very soft" or "soft" (Figure 3–8). When comparing traditional and loudness scaling methods, Skinner and colleagues (1995) concluded that programs established using

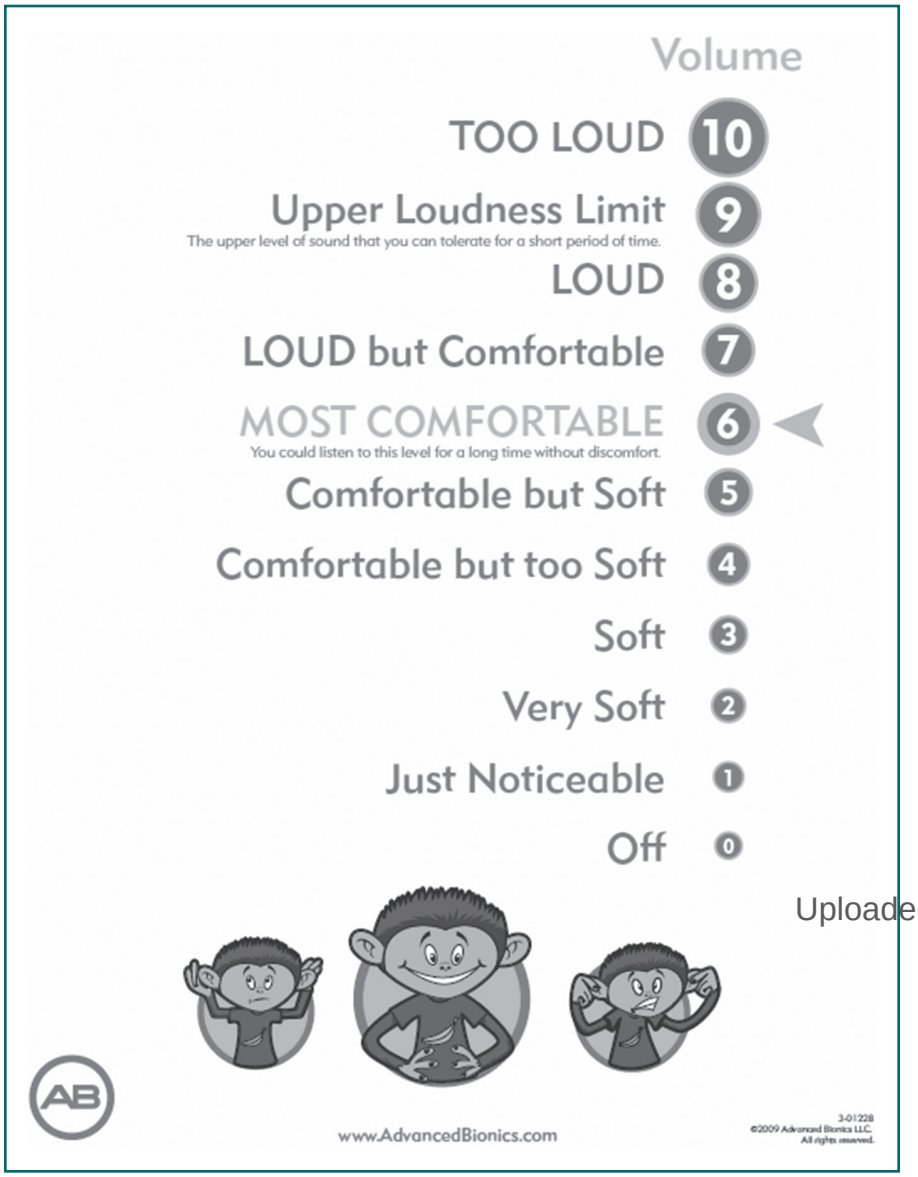



**FIGURE 3–8.** Example of an Advanced Bionics loudness scale for determining electrical threshold with adults. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

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loudness scaling resulted in better access to low-level sounds for recipients of the Nucleus 22 cochlear implant and Spectra body-worn sound processor. Because input-processing characteristics and mapping functions have improved relative to the technology that was available in the Spectra processor, audibility of low-level inputs is often sufficient with T levels set using an audiometric-style approach. However, the psychophysical loudness scaling approach may be beneficial for users with abnormally slow loudness growth over the lower portion of their electrical dynamic range (EDR) (e.g., all stimuli from 125–155 clinical units are perceived as soft) and typical or fast growth for the upper portion of the EDR (e.g., 155–164 clinical units equals the percept ranging from soft to loud, but comfortable). Limited or slow loudness growth over a substantial portion of the lower electrical dynamic range sometimes is described as a "T-tail," which refers to the limited change in loudness percept across a wide range of stimulation levels in the lower part of the EDR. When a T-tail occurs, it often is helpful to increase T levels to the point at which loudness begins to grow with increases in stimulus level. Detection of T-tails and the point at which loudness begins to grow above threshold can be identified using only the psychophysical scaling method. The authors have certainly found this psychophysical approach (i.e., setting T level to a "soft" percept) to be successful in improving sound-field detection thresholds (audibility of low-level sounds) for recipients who have elevated sound-field thresholds (e.g., 35 dB HL and higher) and programs based on conventional measurements (e.g., Hughson-Westlake). However, it should be noted that T-tails are relatively rare, and the psychophysical scaling approach is typically not required to facilitate good access to low-level sounds.

The final alternative method for determining T levels is through the use of manufacturer developed estimation strategies. T levels are estimated on the basis of the user's upper-stimulation levels, typical electrical dynamic ranges with those upper-stimulation levels, and logarithmic mapping functions that aim to place the majority of acoustic inputs into the upper portion of the electrical dynamic range. Research has shown that most recipients achieve adequate audibility for low-level sounds when T levels are estimated from upper-stimulation levels (Spahr & Dorman, 2005). However, some studies have shown that estimated T levels result in inadequate audibility of low-level sounds for some recipients (Holden et al., 2011). This is probably more likely for patients who have atypically narrow electrical dynamic ranges; the authors of this book contend that T levels should be measured manually to ensure audibility for low-level sounds.

### Special Considerations for Setting Threshold Levels for Children

Setting electrical-threshold levels in children may be a challenge; however, T levels can be established accurately for most children by using traditional, age-appropriate methods to determine audiometric thresholds. These techniques include behavioral observation audiometry (BOA), visual reinforcement audiometry (VRA), conditioned play audiometry (CPA), and the standard audiometric approach used for adults. The success of a particular technique is dependent on the chronological and developmental age, mood and behavior, and listening experience of the child as well as the conditioning, reinforcement strategies, and skill level of the programming audiologist and programming assistant.

Regardless of the technique, an animated and enthusiastic programming test assistant is essential for obtaining valid stimulation levels with children. It is almost impossible for a single person to effectively program the cochlear implant sound processor of a young child, manage the child's behavior, and facilitate cooperation during a lengthy programming session. Many clinics designate an audiologist as the programmer and a second audiologist or a speech-language therapist (e.g., Listening and Spoken Language Specialist [LSLS]) as the test assistant. Children will be particularly comfortable with the speech-language pathologist or LSLS whom they see on a weekly basis. Also, the LSLS can practice the CPA response (i.e., drop the toy in the bucket when stimulus heard) in weekly therapy sessions to encourage cooperation during audiology appointments. Other clinics may use trained support personnel as the programming assistant. All of

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these options are viable as long as the assistant is familiar with the objective of the task and adept at managing the behavior of the child. The job of the test assistant is to facilitate a conditioned response from the child, keep the child engaged, ensure the transmitting coil remains on the head, manage the child's behavior, assist in objective measures (i.e., electrically evoked stapedial reflex threshold), and help in observing the child's responses.

In pediatric audiology, BOA has long been considered an unreliable method of determining the threshold of audibility of a young child. Recently, however, Jane Madell (2008) described a unique variation of BOA for infants (0 to 6 months) that provides a more reliable and valid measure of behavioral threshold. Briefly, the Madell approach involves the observation of a change in the child's sucking pattern when listening to test stimuli. To aid in the success of the measure, Madell recommends that the child is hungry upon arrival to an appointment. A pacifier then is inserted into the child's mouth, and test stimuli are presented. Detection of a stimulus is noted when a time-locked sucking pattern changes (i.e., cessation or initiation of sucking). The authors of this book consider the sucking response to be one of the most effective means to obtain minimal response levels for very young recipients (6 to 9 months). However, considerable caution should be exercised when using any type of BOA approach to evaluate minimal response levels. Numerous studies have clearly shown that infants' BOA responses can span a wide range and are occasionally well above true threshold. As a result, the clinician must be very careful not to set T levels too high when using any type of BOA technique.

Traditional VRA testing techniques are optimal (and desired over BOA) for determining T levels in pediatric implant recipients with a developmental age ranging from 8 to 20 months. As a result, it is imperative that pediatric cochlear implant clinics have VRA equipment in sound-field test booths and programming rooms. The authors of this book are compelled to stress that accurate VRA assessment of T levels is entirely predicated upon the programming clinician's and programming assistant's ability to effectively administer the VRA procedure. It is imperative that both the programming clinician and test assistant are familiar with a gold standard VRA protocol. A complete description of VRA technique is beyond the scope of this book. However, a brief description of VRA for programming cochlear implants in children is provided below.

1. Preparation
   a. The child should be seated comfortably (torso and feet supported) with reinforcing toys at ± 90°
   b. Reinforcing toys should be invisible prior to signal presentation.
   c. The programming assistant should ensure that the child's attention is at midline. This may be achieved by displaying a mundane toy (e.g., peg, block). The toy should be interesting enough to establish midline attention but not so distracting that it causes the child to ignore the programming stimulus.
2. Establish the Conditioned Response
   a. Present a signal below a level at which the clinician believes a response will be audible and increase in relatively large step sizes (e.g., 5–10 CL for Nucleus recipients) until the child makes a full (90 degree) head turn. It is important to note that in classical VRA, the procedure is initiated by a period of response shaping (i.e., conditioning). However, in clinical VRA, this period of response shaping may waste valuable attention and responses with young children. As a result, most experts suggest that VRA should be initiated by immediately searching for threshold. Once the clinician and test assistant observe a time-locked, volitional head turn to the test signal, the clinician should commence with a threshold search (see point "3" ["Threshold Search"] in this outline.
   b. When child makes a time-locked head turn, the clinician should activate the VRA toy after child makes 90-degree head turn.
   c. Test signal: Begin with apical channel. After threshold (MRL) is obtained, proceed in a staggered approach (basal electrode followed by middle electrode, etc.)
      i. Measure as many electrodes as the child's attention will allow.

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ii. Response shaping is necessary if volitional head turn does not occur.

d. Response Shaping
   i. Present signal at level presumed to be audible and AVT orients child to VRA toy. Determining a level that is likely to be audible may be achieved through the use of electrically evoked compound action potential (ECAP) assessment. Presentation at levels exceeding the ECAP threshold should be audible.
   ii. Pair signal presentation with animated toy reinforcement for 1 to 2 seconds
   iii. After shaping, observe for volitional head turn at same level used for shaping
   iv. Use captivating toy (light, puppet, etc.) if necessary to bring back to midline
   v. Attempt response shaping a second time if volitional head turn does not occur
   vi. Abandon VRA if volitional head turn does not occur after final shaping attempt.
   vii. Attempt to establish conditioned response with bone conduction stimulation (vibrotacticle stimulation at equipment limits of audiometer at 250/500 Hz). Use of vibrotactile stimulation to establish VRA conditioned response will allow clinician to determine whether child is capable of a head turn and may also facilitate the child's understanding of VRA procedure.

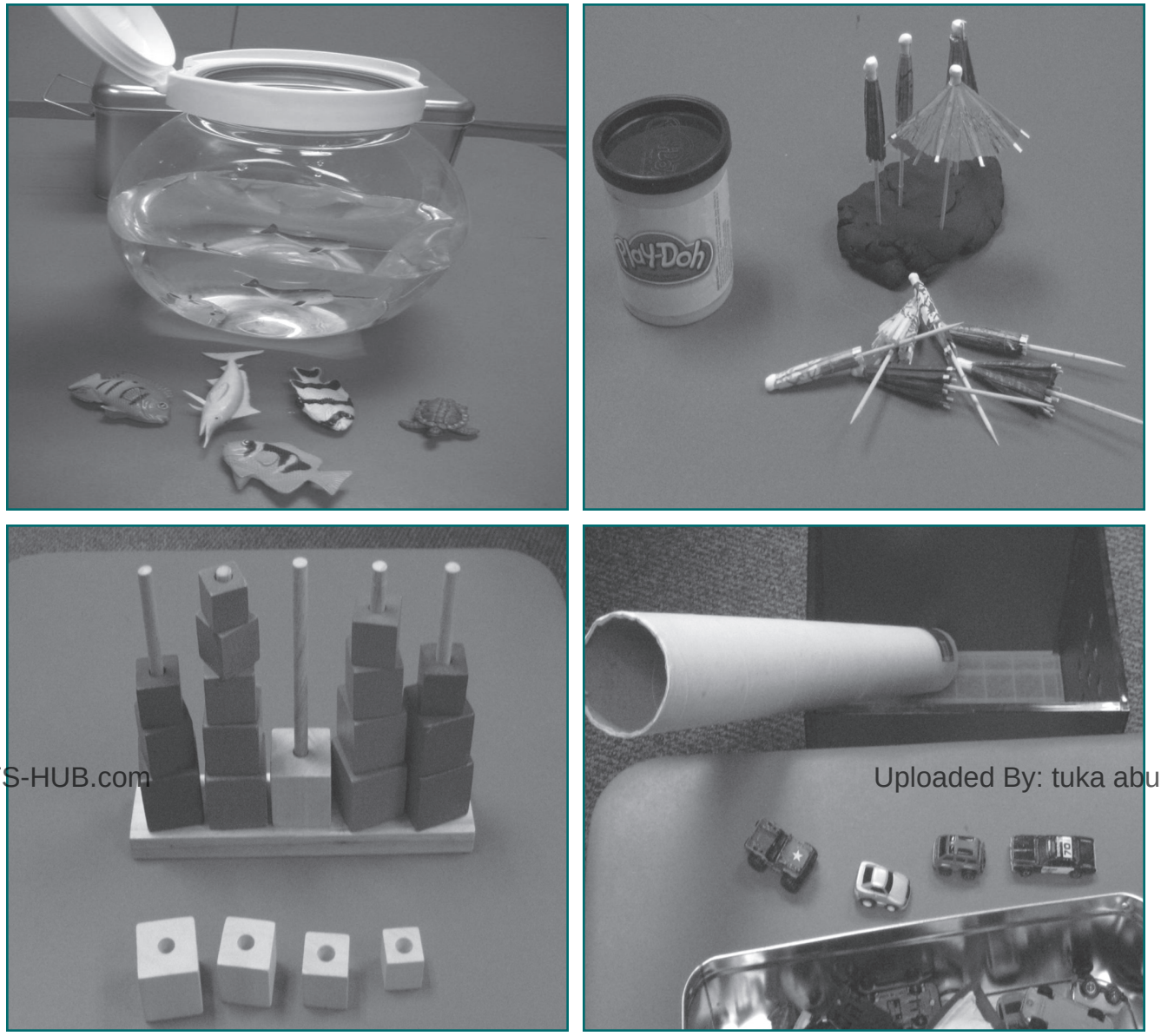
3. Threshold Search
   a. After conditioned response is confirmed, decrease signal level (e.g., by 10 CL) and present test signal again
   b. Use up/down 10 CL bracketing steps to determine threshold (up 5/down 10 dB and eventually up 2/down 4 CL steps may be used for later tests when the clinician can pursue threshold with a finer resolution, because she or he has a better idea of where threshold resides)
   c. Threshold is obtained when child responds twice (or three times if doubts persist) to ascending presentation → Alternate signals as necessary (NBN/WT/Speech/Ear)
   d. Test Order
      i. Apical, Middle, Basal, Mid-Apical, Mid-Basal, Most Basal
4. Pearls
   a. Use Control Trials
      i. Less than 30% suggest reliable test; Greater than 50% → Abandon VRA
      ii. It's easy to teach a head turn but the hard part is to prevent head turn when it is unneeded.
   b. One accurate threshold is better than several inaccurate thresholds
   c. VRA is a test of detection, not a test of orientation
   d. Once a conditioned response is established, prompting/cueing should be used with extreme caution.
      i. LSLS (or whoever is assuming the role of the test assistant) should communicate with AuD that signal appears to be audible but child is not responding
      ii. AuD should communicate with LSLS (i.e., test assistant) when signal should be audible but child is not responding.
   e. Providing credit for responses other than a full head turn or a CPA response is playing with fire.
   f. Alternate frequency (i.e., channel) of stimulation to maintain attention
   g. When in doubt, do not reinforce!
   h. Communicate!

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Traditional CPA testing techniques are suitable for measuring T levels in children with developmental ages ranging from 18 months to 8 years. Although 18 months is traditionally younger than expected for this task, the authors have worked with children who develop a conditioned play response prior to 18 months of age. Some older children (≥6 years) will prefer a conventional audiometric approach, whereas others will not be able to maintain attention over long periods of time, and as a result, CPA serves as a

good tool to facilitate their attention to and cooperation in the threshold-seeking task. Therefore, clinicians should be flexible and use the approach that yields the most reliable and valid threshold responses for a particular child.

One factor that influences the success of CPA is a child's interest in the variety of activities, toys, and games used during testing. Because children with cochlear implants must attend multiple programming sessions and audiometric evaluations per year, it is critical to provide a variety of exciting and entertaining CPA tasks. Figure 3–9 provides several examples of CPA games proven successful for establishing T levels. For young children or those who are inattentive and hyperactive, it may be necessary to use multiple CPA toys during the session. In fact, the authors of this text will occasionally arrange for 10 to 15 different CPA tasks to be available in order to obtain T levels across the array for a child with a short attention span.




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**FIGURE 3–9.** Examples of games used for conditioned play audiometry and programming sessions.

Although conventional toys and snacks often serve as excellent manipulatives to facilitate CPA assessment, the authors of this book also have achieved excellent CPA success through the use of electronic media. For instance, we have connected a monitor to a button the child may push when he or she detects the programming signal. When the child pushes the button, the slide is advanced within a PowerPoint slide show possessing child-friendly pictures (images of Disney figures, Pixar figures, comic book heroes, princesses, animals, etc.). Additionally, there are several tablet apps that may be used to obtain T levels via a CPA approach. For instance, the child can "launch an angry bird" each time he or she hears the test signal. The clinician is encouraged to use his or her imagination to determine ways in which tablet apps may be used to measure T levels.

The same ascending/descending approach used to measure thresholds in adults also is applicable to young children. However, for young children with shorter attention spans, larger step sizes may be used to obtain T levels. Also, similar to adult procedures, measurement of T levels should begin with a low-frequency (i.e., apical) electrode contact. When a valid threshold is obtained for a low-frequency contact, the clinician should promptly move on to an electrode contact in the middle of the array, followed by a basal electrode contact. If the child's attention is maintained, T levels should be obtained on as many electrodes as possible. Interpolation is used to estimate T levels on unmeasured channels.

At initial activation, the child's T levels are likely above the true threshold. Thus, it may be necessary to provide a slight decrease in the T levels for the child's map. T levels in children appear to increase rather significantly through the first few months of implant use and stabilize by 3 months to 1-year post implantation (Henkin et al., 2006; Henkin, Kaplan-Neeman, Muchnik, Kronenberg, & Hildesheimer, 2003; Hughes et al., 2001). After the first 6 months of implant use, the authors recommend that young children return for evaluation and programming on a quarterly basis until the child is 7 years old (or 2 years post-activation for older children) in order to ensure that T levels are appropriate.

By no later than 1 month postactivation, the clinician should measure the child's minimal response levels in the sound field to frequency-specific stimuli. Responses should be no greater than 30 dB HL in the low frequencies (250 and 500 Hz) and should not exceed 25 dB HL in the mid to high frequencies. If the child's responses do not meet these criteria, the clinician should seek to optimize T-level stimulation through manual adjustments or consider adjusting other parameters (e.g., IDR, maplaw) to optimize access to low-level sounds (see Chapters 4 through 7).

Manufacturer-developed, threshold-estimation procedures may be used with children only when T levels cannot be measured easily, and sound-field detection thresholds indicate that the estimated T levels are providing good access to low-level sounds (i.e., sound-field detection thresholds at 25 dB HL or better). In general, the authors recommend manual measurement of threshold with children to ensure adequate stimulation for low-level inputs. Psychophysical loudness scaling procedures for setting T levels also are not recommended for children because the task often is too difficult for them to perform. However, given the sensitivity of loudness scaling for detecting T-tails and the point at which loudness begins to grow above threshold, clinicians need to be sensitive and aware of two red flags. First, if a child has an unusually large electrical dynamic range, and the clinician is absolutely certain that upper-stimulation levels are set appropriately, then it is probable that loudness growth is very shallow in the lower part of the electrical dynamic range. As a result, it may be helpful to provide a global increase in T levels across the electrode array, particularly if sound-field thresholds are elevated. A global change in stimulation refers to increasing or decreasing stimulation across all channels by the same amount. The clinician also should be attentive for signs of insufficient access to low-level sounds. If the child's caregivers or LSLS reports that the child does not consistently respond to soft sounds, it may be necessary to increase T levels. Finally, if sound-field detection thresholds are elevated, T levels should be increased to provide better access to low-level sounds. Prior to adjustment of T levels, the clinician should verify that

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elevated sound-field thresholds are not related to a faulty microphone. Slightly malfunctioning microphones typically cause greater elevation for high-frequency sounds, whereas more significant faults may cause elevated sound-field thresholds across the frequency range. Additionally, if the sound processor contains microphone covers to protect against moisture and debris, then the clinician should ensure that the covers are clean and free of obstruction and debris. Dirty microphone covers are a common culprit of elevated sound-field thresholds. It should be noted that MED-EL recommends adjustment of the maplaw parameter to improve access to low-level sounds instead of increasing THR levels. This is discussed in detail in Chapter 6.

One streamlined approach to measuring T levels is to provide gradual and global increases to T levels across the electrode array until the child responds to speech and environmental sounds. However, this approach should be used as a last resort because T levels are likely to vary across the electrode array. Therefore, based on crude responses to environmental stimuli, global changes to T levels likely will result in insufficient settings for some channels.

### Additional Measure to Ensure Adequate T Levels and Confirm Audibility

For children and adults, many clinicians ensure adequate audibility for sounds across the speech frequency range by confirming that the recipient can detect all six Ling sounds (/ah/, /oo/, /ee/, /s/, /sh/, and /m/; Ling, 1976, 1989). These six phonemes encompass the typical range of speech frequencies, so audibility for the Ling sounds should ensure audibility for most all of the sounds within the speech frequency range. During the first few days of implant use, all six Ling sounds should be audible when spoken at an average conversational level, 6 feet away from the user. By 1 month of implant use, all six Ling sounds should be audible when spoken at an average conversational level from a distance of at least 15 feet from the user. Furthermore, a goal of accurate discrimination of all six Ling sounds should be rapidly pursued, and on a daily basis, the parents of young children should ensure that the child can hear all six Ling sounds.

### Setting Upper-Stimulation Levels for Adults

The provision of optimal upper-stimulation levels (i.e., M, most comfortable level [MCL], or C levels depending on manufacturer) is one of the most challenging aspects of the programming process. Even with adults, it can be difficult to determine exactly where upper-stimulation levels should be set across the electrode array in order to provide optimal sound quality and speech recognition. Underestimating the upper-stimulation levels may sacrifice speech recognition, hamper sound quality, and limit self-monitoring abilities of the recipient's voice (i.e., intensity, vocal quality). Overestimating stimulation levels may cause discomfort, hinder speech recognition, adversely influence sound quality, and foster an aversive reaction to the implant.

Similar to T levels, the maximum level of stimulation that a recipient will tolerate is likely to change over the first few months of cochlear implant use. Many users are able to reach a desired level of stimulation for optimal performance after several weeks of cochlear implant use. However, some recipients will require gradual changes over several months to reach an optimal upper-stimulation level. In only rare exceptions are adults comfortable with appropriate upper-stimulation levels during the first couple of days or week of implant use. Most importantly, recipients differ in what they initially will tolerate. The clinician must balance the importance of an optimal cochlear implant program with the prevention of overstimulation in the early stages of implant use.

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One popular method to measure upper-stimulation levels is via psychophysical loudness scaling. To begin this procedure, the programming stimulus is presented to an electrode contact or to a band of channels, and the recipient is asked to indicate the loudness percept of the stimulus by pointing to a loudness scaling chart (see Figure 3–8). The clinician gradually increases the level of the programming stimulus until the recipient reports that it is comfortably loud. This

process is repeated for many, if not all, of the electrode contacts in the array. If upper-stimulation levels are not measured for each individual electrode contact, interpolation is used to set the levels of the unmeasured electrode contacts. It should be noted that the authors of this book consider this method to overly time consuming, and more importantly, it may not be entirely indicative the of upper-stimulation levels that are optimal for real-world use because electrode-specific measures do not account for summation that occurs when listening to broadband, every day sounds.

Another approach to setting upper-stimulation levels is to provide a global increase in upper-stimulation levels while the user listens to live speech. The upper-stimulation levels will be set where speech and environmental sounds are most comfortable. Some clinicians also will choose to then tilt or provide tapered decreases or increases in the low- or high-frequency portion of the array to address patient complaints or comments regarding sound quality. For instance, if the recipient reports that sounds have too much bass or are too sharp, the clinician can taper off the upper-stimulation levels in the low or high frequencies, respectively. On the other hand, if a recipient cannot hear his or her voice very well or if speech lacks sufficient clarity, the clinician may increase upper-stimulation levels in the low or high frequencies, respectively.

The authors of this book frequently use the method of globally increasing upper-stimulation levels to the recipient's most comfortable listening level while he or she listens to live speech and environmental sounds. When using this approach, the upper-stimulation levels should be relatively flat across the array (i.e., a flat upper-stimulation-level profile; upper-stimulation levels are similar across the array with only gradual changes across the array). Then, the clinician should gradually increase the upper-stimulation levels from a very low level (i.e., near or below the T levels) while the recipient listens to speech and environmental sounds. The recipient's task is to inform the clinician when upper-stimulation levels reach a point that provides optimal sound quality, comfort, and clarity for speech. Oftentimes, recipients will inform the clinician that the optimal level has been reached, when in reality, an additional increase may be beneficial. As a result, when a recipient initially reports that upper-stimulation levels are satisfactory, the authors of this book typically inform the recipient that we are going to provide a small increase in stimulation in order to determine whether sound quality and speech recognition improve. We note to the recipient that we will decrease the upper-stimulation levels if we reach a point that the recipient considers to be too loud or associated with poorer sound quality. Typically, the additional increase in upper-stimulation level does improve the recipient's percept. We continue to increase upper-stimulation levels in small increments until the recipient reports that sound quality, speech recognition, and/or comfort begins to deteriorate. At that point, upper-stimulation levels are decreased to a point that provides optimal sound quality, speech recognition, and comfort for the recipient. This approach allows the recipient to fully explore his or her electrical dynamic range to ensure that it is optimized.

After upper-stimulation levels are globally set to a comfortable level, it is imperative that the clinician completes another step in order to optimize the recipient's outcome. Specifically, the clinician should conduct loudness balancing at upper-stimulation level across the electrode array. Ensuring equal loudness at upper-stimulation level is likely one of the most important components in creating a program that provides optimal speech recognition and sound quality. Indeed, many evidence-based hearing aid prescriptive methods have unequivocally demonstrated that the provision of equal loudness across the frequency range results in optimal speech recognition.

Establishing equal loudness across the electrode array assists in maintaining the typical loudness/intensity relationship that exists for different phonemes. For example, low-frequency phonemes (e.g., vowels) are typically louder and more intense than high-frequency phonemes (e.g., fricatives). If high-frequency upper-stimulation levels were considerably louder than the low-frequency upper-stimulation levels, then the recipient may perceive the fricatives to be much more intense than vowels, which would likely impair speech recognition. Additionally, if one electrode possesses a C level that is much louder than the remaining electrodes, then stimulation on that

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electrode may dominate the recipient's impression of overall loudness. As a result, the recipient will adjust the volume control to ensure that electrode provides comfortable stimulation, and the remaining electrodes may possess suboptimal upper-stimulation levels.

The authors of this book typically conduct loudness balancing at upper-stimulation level (or 80% of the upper-stimulation level for MED-EL recipients) for two channels at a time. Loudness balancing typically begins with the most apical channel and progresses toward the more basal end of the array. However, for some recipients who have a difficult time tolerating high-frequency stimuli, we will conduct loudness balancing from the basal end toward the apical end. We typically use a chart to facilitate the loudness balancing procedure (Figure 3–10). It is important to stress to the recipient that he or she should attend to the loudness of the signal rather than the pitch of the signal. Many users perceive the more basally located electrode to be sharper because it possesses a higher pitch. It is imperative that the clinician instructs the recipient to focus on the difference in loudness between the two channels rather than the difference in pitch.

When conducting loudness balancing, the adjustment to the stimulation level always should be to the second electrode in the pair because the first electrode is balanced to the preceding electrodes in the array. On occasion, clinicians will adjust T levels to achieve loudness balance at the lower end of the electrical dynamic range (e.g., 25 or 50%). However, in the authors' experience, loudness balancing should be confined to upper-stimulation levels because the T levels already are set appropriately using reliable and valid measures or estimation; therefore, it is an uncommon situation when they would need to be adjusted.

It should be noted that the upper-stimulation-level profile is typically flatter or more gradual across the electrode array after loudness balancing has been completed. In general, the upper-stimulation level is typically flatter than the T-level profile (Botros, Banna, & Maruthurkkara, 2013). The reader is referred to Chapter 2 for more information on loudness balancing.

A third and more objective approach to setting upper-stimulation levels is the use of electrically evoked stapedial reflex thresholds (ESRT). Many clinicians use ESRT as a standard component of their programming test battery because it is a straightforward and simple measure to administer. First, the clinician places an acoustic immittance probe into the external ear canal contralateral to the cochlear implant. Second, he or she continuously records the acoustic admittance, set to a 226 Hz probe tone, while presenting the programming stimulus for obtaining upper-stimulation level to the cochlear implant in an ascending manner. When the level of the stimulus from the implant is sufficiently intense to elicit the stapedial reflex, the clinician observes a decrease in admittance, which is time-locked with stimulus presentations (Figure 3–11). This decrease in

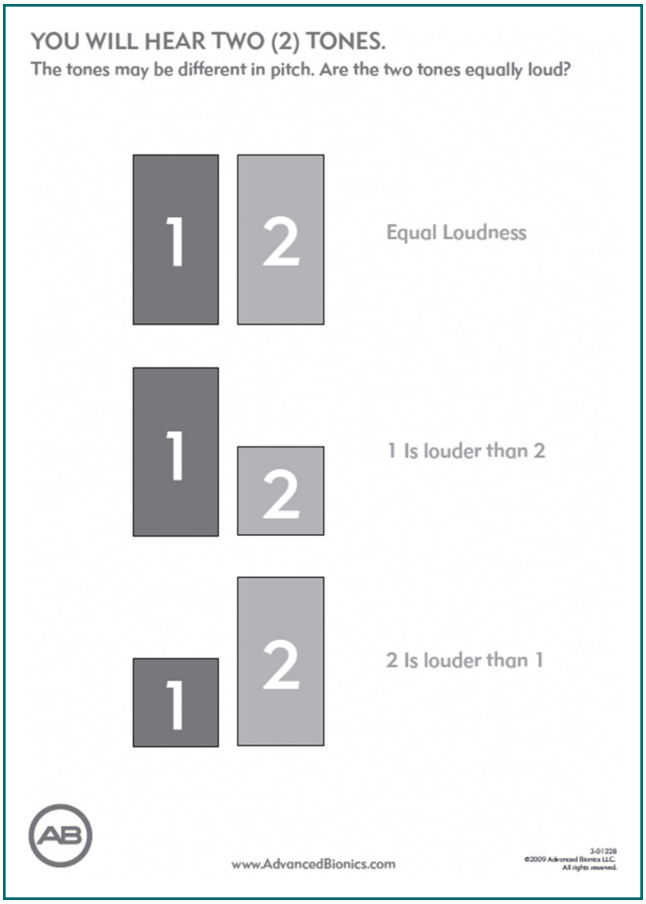



**FIGURE 3–10.** Advanced Bionics chart that may be used during loudness-balancing procedure. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

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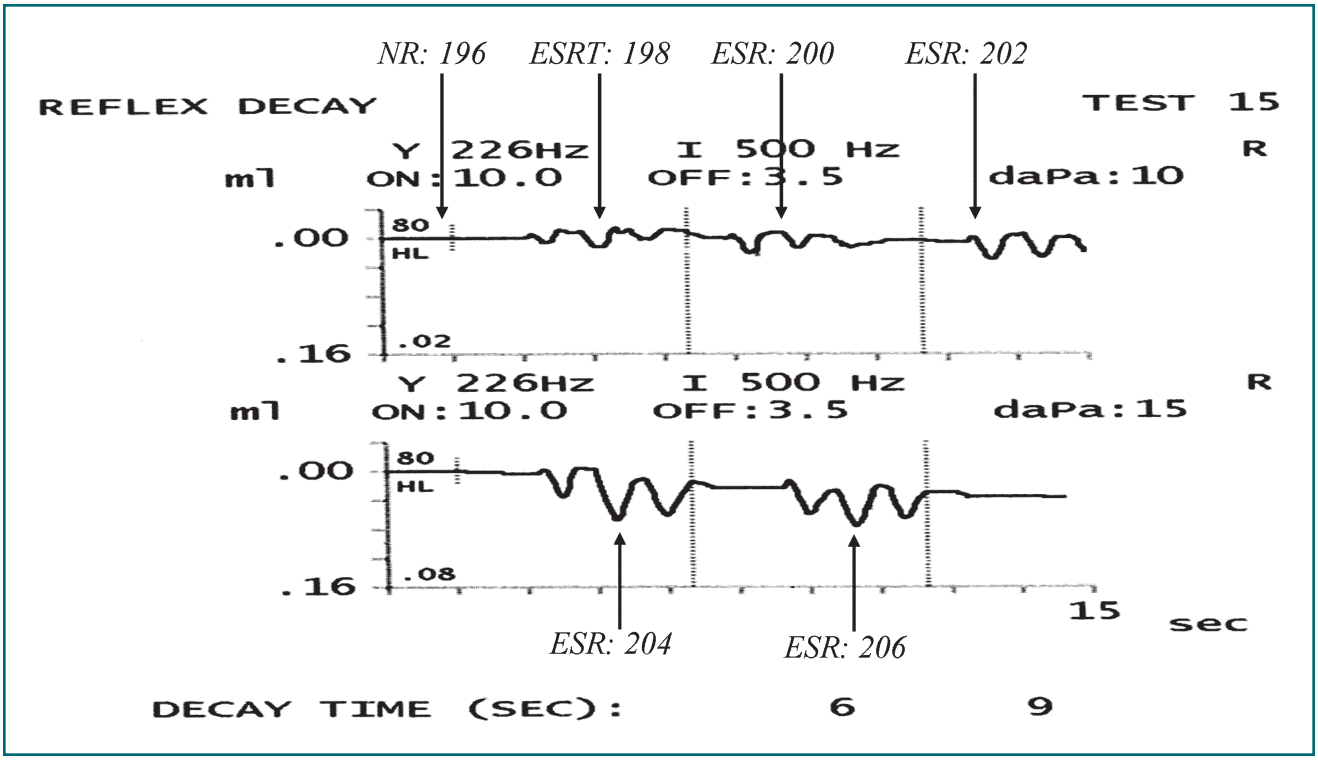



**FIGURE 3–11.** Visual display of the ESRT obtained with a cochlear implant recipient.

admittance can be observed in the ear contralateral to the cochlear implant because the stapedial reflex is a bilateral response. It actually is recommended that the immittance probe be placed in the contralateral ear as the response is more likely to be measured in an ear that is unaltered by surgery.

The ESRT is measured on as many channels as possible and remaining channels are set via interpolation. Initially, the upper-stimulation-level profile is set according to the ESRT and then upper-stimulation levels are decreased globally to a level that the clinician is certain will be comfortable for the recipient (i.e., near threshold). Next, while the recipient listens to speech and environmental sounds in live speech mode, the upper-stimulation levels are gradually increased globally (in small increments) to a level that provides optimal speech recognition, sound quality, and comfort. It should be noted that the upper-stimulation levels should typically not be set higher than the ESRT. Doing so likely would result in continuous elicitation of the stapedial reflex threshold to moderately intense sounds, such as average conversational speech. This is not typical for persons with normal hearing and should not be a reality for cochlear implant users. Adults who used high-power hearing aids for many years frequently will desire upper-stimulation levels that are excessively high, but setting the upper-stimulation levels at erroneously high levels may hamper performance. In these cases, the ESRT can serve as a good guide to determine an appropriate limit for upper-stimulation levels.

Published reports suggest ESRT is measurable in approximately 65 to 85% of implant users (Battmer, Laszig, & Lehnhardt, 1990; Brickley et al., 2005; Gordon, Papsin, & Harrison, 2004). Subtle middle-ear anomalies are likely responsible for the inability to measure the response in approximately 30% of recipients. In fact, the response is initially measured with the immittance probe in the ear opposite the implant because cochlear implant surgery tends to reduce the likelihood that the ESRT is measured with the probe in the implanted ear. The exact mechanism of this is unknown, but it is thought that the surgery likely

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increases the stiffness of the middle ear (possibly at the stapes), which will reduce the chances of measuring the ESRT via immittance assessment. If the response is not present in the contralateral ear, the measurement may be attempted in the ipsilateral ear. Unfortunately, an ipsilateral response may not be present because, as stated previously, the cochlear implant surgery may result in subtle fixation of the middle ear system. It should be specifically noted that it may be more problematic to measure the ESRT in bilaterally implanted recipients because both ears have undergone surgery.

Several research studies show that the ESRT is an excellent predictor of upper-stimulation levels (Brickley et al., 2005; Jerger, Oliver, & Chmiel, 1988). In fact, correlation coefficients between ESRT and upper-stimulation levels are strong and range from .79 to .85 (Brickley et al.; Hodges et al., 1997; Lorens, Walkowiak, Piotrowska, Skarzynski, & Anderson, 2004). Recipients also will typically perceive equal loudness when upper-stimulation levels are based on ESRT assessments (Hodges et al.). As previously discussed, balancing of loudness across the electrode array should result in optimal sound quality and speech recognition. In fact, programs that are based on ESRT often result in higher sound-quality ratings and similar or better speech-recognition performance when compared to programs based on conventional psychophysical loudness assessments (Hodges et al.; Spivak, Chute, Popp, & Parisier, 1994; Wolfe & Kasulis, 2008).

Once assessment of upper-stimulation level is completed, the clinician may consider sweeping at upper-stimulation level. Sweeping simply involves the sequential presentation of the programming stimulus across the electrode array. It should be noted that the authors of this book do not conduct sweeping on a routine basis. Instead sweeping is conducted when the user is experiencing problems with sound quality or nonauditory side effects (e.g., facial nerve stimulation). Sweeping may also occasionally be conducted during the early stages of implant use to ensure that all electrodes are functioning as desired.

When implementing this procedure, the programming stimulus is presented sequentially or swept across all electrode contacts in the array beginning at the apical end and ending at the base. Three objectives are associated with electrode sweeping: (1) measuring sound quality, (2) determining appropriate pitch transitions, and (3) confirming equal loudness across channels. Sweeping to evaluate sound-quality issues is not performed at every mapping session, but it is a good practice for newly implanted recipients or for those experiencing difficulties that cannot be resolved with conventional mapping techniques or troubleshooting of the external equipment.

A determination of sound quality is performed by asking the recipient to listen to the swept test signal and report whether any of the stimuli have poor sound quality (i.e., noise). Normally, the stimulus should possess a tonal percept, but if the user reports hearing noise, experiences pain or other unwanted sensations, has facial nerve stimulation, or simply reports poor sound quality, the clinician may consider disabling or adjusting the stimulation on the contact.

Second, the clinician may use electrode sweeping to ensure that the recipient perceives an orderly and expected transition in pitch as stimulation is swept from the apical (i.e., low pitch) to the basal (i.e., high pitch) portion of the array. If the expected tonotopic organization is not observed, the clinician may either disable or reallocate the anomalous electrodes to different channels.

Finally, sweeping may be used to confirm equal loudness across the entire array rather than the traditional two to three electrodes at a time (i.e., loudness balancing). When using this procedure, the clinician instructs the recipient to listen to all of the stimuli and raise a hand if any sounds are substantially softer or louder than the rest. This task may be performed at upper-stimulation levels or at lower portions of the electrical dynamic range.

In summary, the authors of this book typically rely on global increases in upper-stimulation levels in live speech mode coupled with channel-by-channel loudness balancing at upper-stimulation level to optimize the determination of upper-stimulation levels for adult recipients. In many cases, the upper-stimulation-level profile is flatter than the T-level profile, but exceptions do exist. The authors also measure the ESRT with adult recipients in order to have an objective measure of loudness to which behaviorally determined upper-stimulation levels may be compared. In some

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cases, the ESRT suggests that upper-stimulation levels should be altered (increased or decreased). The ESRT is occasionally used as the primary method to set upper-stimulation levels for recipients who are unable to provide reliable judgments pertaining to loudness.

The clinician should consult with experienced programming audiologists or manufacturer's representatives to gather an understanding of typical upper-stimulation levels and electrical dynamic ranges for a particular make, model, and electrode array of the implant under study. For example, Nucleus recipients typically possess an electrical dynamic range between 30 to 50 CL, Advanced Bionics recipients typically possess M levels ranging between 150 to 250 charge units with M levels below 100 and above 300 charge units being relatively rare, and MED-EL recipients typically possess MCL levels between 10 to 25 charge units with MCL over 40 charge units being quite exceptional. Questions should arise when a recipient's stimulation levels and/or electrical dynamic range differ markedly from typical values. Of course, appropriate exceptions to typical values do exist, but the clinician should confirm through speech recognition assessment, sound-field detection thresholds, and subjective assessment that the recipient's levels are optimized, particularly when the stimulation levels deviate from what is typical for a particular make and model of implant/electrode array and for a recipient with a given preimplant history.

### Setting Upper-Stimulation Levels for Children

The ESRT is the highly recommended procedure for setting upper-stimulation levels in children, and clinicians always should attempt to complete this measure with pediatric recipients. ESRT is crucial in determining optimal upper-stimulation levels because psychophysical loudness measures typically are not feasible for children until they are at least 4 years old (at 4 years, a child may begin to provide very basic feedback regarding whether a sound is too soft or too loud). Furthermore, precise measures of loudness scaling and loudness balancing are often not achievable until approximately 8 years of age.

Similar to adults, research supports the use of ESRT for predicting upper-stimulation levels for children. Typically, the ESRT is within an average of nine clinical units of upper-stimulation levels for Nucleus recipients (Spivak & Chute, 1994) and are often lower than upper-stimulation-level settings based on behavioral measures (Bresnihan, Norman, Scott, & Viani, 2001). In other words, when using behavioral measures, clinicians often have a tendency to overestimate comfort levels of children. The authors have found that ESRT is typically within 10 to 15 CL of C-levels for Nucleus recipients, 5 to 10% of M level for Advanced Bionics recipients, and near MCL for MED-EL recipients).

Because of the improved comfort and performance associated with ESRT-based programs, children often will use their cochlear implants for longer periods of time and experience fewer episodes of discomfort to speech and environmental sounds (Bresnihan et al., 2001). Some programming clinicians may be compelled to slowly increase upper-stimulation levels for children over time (Zwolan, O'Sullivan, Fink, Niparko, & CDACI Investigative Team, 2008), and eventually, upper-stimulation levels are likely to be inappropriately high. It is assumed that this occurs, because children attend several programming sessions per year over the course of their first several years of implant use, and at each session, the clinician tends to provide a small increase in upper-stimulation levels. Over a large number of sessions, the upper-stimulation levels increase slowly but markedly and are often much higher than what are observed in the programs of well-performing adult recipients. The ESRT serves as a more objective approach for setting upper-stimulation levels and may assist the clinician in avoiding overstimulating a child.

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In the event that the ESRT cannot be recorded, the clinician must rely on behavioral measures to determine upper-stimulation levels. For children who are preschool-aged and younger, this almost invariably means using behavioral observation of reactions following stimulation. There are numerous strategies available to estimate upper-stimulation levels from behavioral observations. First, the authors recommend that the clinician globally increase upper-stimulation lev-

els while the child listens to speech and environmental sounds. The clinician should begin with a relatively flat upper-stimulation-level profile and upper-stimulation levels that should be inaudible to the child. Initially activating at audible levels may startle and upset the child making it difficult to complete the remainder of the programming session.

The clinician should activate the sound processor in live speech mode and gradually increase the upper-stimulation levels (small increments) while the parent or programming assistant continuously vocalizes to the child (e.g., "ba, ba, ba"; "hi, Johnny!"). The programming clinician should note the level at which the child initially responds to speech. The upper-stimulation levels should be set to a level that provides a typical electrical dynamic range (relative to the T-level measurements—i.e., 40 CL in Nucleus devices) or to upper-stimulation levels that are typical for the implant being used. Of course, the clinician should not attempt to reach "expected" upper-stimulation levels if the child exhibits signs that the stimulation at that level is aversive. Considerable caution should be exercised to ensure that upper-stimulation levels do not cause discomfort or other undesirable effects.

A child's responses to the stimulation he or she receives may be somewhat subtle, so the programming clinician and programming assistant must be very adept at observing even the most subtle signs of anxiety on the part of the child. It is also helpful to have the child's caregivers watch the child's responses and provide feedback if they feel that the child is becoming anxious. Typically, it is ideal to have the child engaged in a mildly entertaining task while this occurs (e.g., looking at toys, watching television). When the child begins to exhibit signs that the stimulus is too loud, the clinician should immediately stop the programming stimulus. Subtle signs that the stimulation is reaching levels that may approach the child's upper level of loudness comfort include holding his or her breath; exhibiting facial expressions of mild concern; looking to a caregiver for reassurance; tensing or stiffening of the body or shoulders; playing more actively or aggressively; wringing hands, clothes, or toys; and producing blinks in response to the stimulus.

For example, it is common that a child will be happily playing with a toy (e.g., stuffed animal, ball, block, doll, toy car) while stimulation is being gradually increased, but once the signal begins to approach a louder level, then the child will toss it off of the highchair tray. Also, if the child is seated in a highchair, he or she may reach for a parent when the signal begins to approach a point that is too loud. If the child is in his or her parent's lap, then he or she may turn to hug the parent or bury his or her face in the parent's chest. It is critically important to promptly observe these signs as they are a possible indication of mild discomfort. When these signs are observed, the stimulation should be immediately decreased until the child appears to be at ease again. Continuing to increase the stimulation level eventually will possibly lead to overstimulation, an adverse reaction, and a negative association with the cochlear implant. If this occurs, it may be difficult to convince the child to wear the sound processor during the early stages. It is recommended that stimulation be slowly increased again to observe whether the child will once again exhibit a similar behavior. If that occurs, then the clinician should provide stimulation levels that appear to provide audibility for speech and environmental sounds while also facilitating the child's comfort. If the child shows no signs of discomfort, then the clinician should set upper-stimulation levels to a point that provides a typical electrical dynamic range (relative to the T levels that were previously measured) or to an overall level that appears to provide audibility for speech and environmental sounds without exceeding levels that are typical for the implant and electrode array of the child. It should be noted that it is wise to sweep at upper-stimulation level once the levels have been set in live speech mode. This will allow the clinician to observe for any channel-specific undesirable effects, such as loudness discomfort or a nonauditory response, such as facial nerve stimulation.

When limited information is available and upper-stimulation levels are set globally, it is typically most appropriate for upper-stimulation levels to possess a fairly flat profile with only gradual changes across the array. Fine-tuning of upper-stimulation levels may be accomplished over time as the clinician gathers information

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from observations during programming sessions as well as from the child's caregivers and therapists. For example, an increase in high-frequency upper-stimulation levels may be warranted if the AVT reports that the child does not consistently respond to high-frequency phonemes (fricatives such as /s/).

Although it is the exception, some children simply do not exhibit any signs that they are detecting the programming stimulus. The clinician should be very cautious to avoid overstimulation in these cases. T levels and upper-stimulation levels should be gradually increased over time while seeking feedback from the child's AVT and caregivers. For instance, an increase in T levels and upper-stimulation levels would be warranted if the child's caregivers reported limited to no responsiveness to speech and environmental sounds during everyday use. The clinician should consistently administer evidenced-based, subjective questionnaires to evaluate the auditory skill development of the child. Excellent tools to accomplish this objective include the LittlEars and PEACH questionnaires (Ching & Hill, 2005; Tsiakpini et al., 2004). The reader should also refer to the University of Western Ontario PedAMP (Bagatto et al., 2011) battery for the assessment of auditory skill development, which includes normative values that should be achieved with levels of auditory experience.

The electrically evoked compound action potential (ECAP) may serve as a guide to aid in the determination of stimulation levels for children who do not have ESRT responses and who provide limited behavioral responses to programming stimuli and environmental sounds. However, as mentioned several times throughout this book, the correlation between ECAP and behavioral stimulation levels is typically modest at best, and as a result, the ECAP should be used as a last resort to guide behavioral stimulation levels. The ECAP does provide a level of stimulation that should definitely be audible to the recipient. As a result, it may be used to determine levels at which VRA conditioning/response shaping occur as well as program/MAP stimulation levels that should be audible. The provision of audibility is a primary objective of early implant use, so the ECAP provides valuable information in that regard. T levels should never exceed the ECAP threshold. Also, the clinician should be concerned when the upper-stimulation levels are substantially higher than the ECAP thresholds. Additional information pertaining to the use of ECAP for programming purposes may be provided in Chapter 7.

Upper-stimulation levels may also be measured to a single-channel programming stimulus. This approach provides a frequency-specific assessment of loudness judgment, which may yield helpful information regarding the child's stimulation needs across the frequency range. However, the authors of this book prefer to initially begin by measuring the child's responsiveness to broadband signals, such as live speech. Channel-specific measures can be useful when the child is experiencing difficulty with obtaining good access to sounds within a particular frequency range. These measures may also be helpful in the event that a child is expressing discomfort with use of the device. Channel-specific measures may identify a particular channel or group of channels that are responsible for the aversiveness.

If channel-specific measures are completed, the clinician should use a similar approach as previously described for the adjustment of upper-stimulation levels to live speech. Specifically, the programming stimulus should initially be presented at a low level and then be slowly increased in small steps. The clinician, the programming assistant, and the child's caregiver should all closely observe for any signs that the signal may be approaching an uncomfortable level for the child. Again, all parties should look for subtle signs of aversiveness on the part of the child, and every attempt should be made to avoid stimulating at a level that would cause discomfort for the child. It is worth reiterating once again that channel-specific measures may result in upper-stimulation levels that vary considerably across the electrode array. Although some recipients do require a different magnitude of stimulation across the electrode array, it should be mentioned again that upper-stimulation levels for programs using monopolar coupling are often relatively flat when compared to the T-level profile. At the very least, the clinician should be leery of upper-stimulation

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levels that change markedly from one electrode to the next or from one section of the electrode array to the next. The clinician should confirm that a valid reason exists to support the decision to set upper-stimulation levels at substantially different levels across the electrode array.

For older children (≥4 years), upper-stimulation level may be set via simplified psychophysical loudness approaches as described by Serpanos and Gravel (2002). Figure 3–12 shows an example of a basic loudness scaling card, which based on the authors' experience, is used successfully with preschool and elementary-aged children. Children who are 8 to 9 years old and good readers often can participate in the traditional psychophysical loudness scaling procedures. Again, it should be noted that loudness assessment is frequently a difficult task with children who are born with severe to profound hearing loss. This is the primary reason for why the authors of this book prefer to use ESRT when available to determine upper-stimulation levels. Many older children are able to complete loudness balancing assessment (i.e., which of the two sounds is louder?). When ESRT cannot be obtained, the authors of this book recommend the use of loudness balancing with children 7 years of age and older. Figure 3–10 shows an example of a loudness-balancing card that may be used during the procedure.




**FIGURE 3–12.** Advanced Bionics loudness scaling card for young children. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

In summary, setting the upper-stimulation levels for young children is quite possibly one of the most difficult tasks facing the programming audiologist. The clinician should proceed conservatively in order to prevent loudness discomfort and reluctance on behalf of the child to use his or her cochlear implant. However, optimal progression of the child's auditory-skill development is greatly influenced by the provision of sufficient stimulation form the implant across the frequency range. Inappropriately low stimulation levels will deprive the child of the access he or she needs to develop age-appropriate speech, language, and auditory abilities. When possible, the provision of upper-stimulation levels for children should be based on the ESRT measurement. When ESRT cannot be obtained, the clinician must rely on a combination of behavioral observation of the child's responses to speech, environmental sounds, and programming stimuli as well as on the clinician's knowledge of what constitutes appropriate stimulation levels for a typical recipient with a particular type of cochlear implant. The clinician should receive continuous feedback from the child's therapists and caregivers to ensure that the stimulation from the implant is not uncomfortable but that it provides sufficient access to speech sounds across the frequency and intensity ranges common for everyday use. Subjective questionnaires are often helpful to provide a formal assessment of the child's auditory skill development that may be referenced to age-appropriate norms. Although the ECAP response is not an excellent predictor of upper (or lower) stimulation levels, it may be used as a guide to determine stimulation levels that should provide audibility and to avoid excessive stimulation.

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### Additional Considerations in the Measurement of Stimulation Levels

While adjusting upper-stimulation levels in live speech mode, it may be helpful for children to play with noise makers (e.g., drum sticks, xylophones, tambourines). Typically, when upper-stimulation levels are set too low, children will play the noise makers enthusiastically and create sounds that are excessively loud. However, when the upper-stimulation levels approach an uncomfortable level, children will stop playing vigorously with the noise makers or will start blinking to any sharp or loud sounds the noise makers produce. The upper-stimulation level should be set below the level that elicits these behavioral changes and at a level that is obviously tolerable by the child. Additionally, these noisemakers may be used to evaluate the child's responsiveness to sounds occurring at different frequency ranges (low-, mid-, and high-frequency sounds).

## Adjustments to Special Parameters

### Stimulation Rate

As mentioned in Chapter 2, a great deal of interest exists in how stimulation rate affects performance. Numerous researchers have explored the effect of stimulation rate on speech recognition and perceptual sound quality (Balkany et al., 2007; Buechner et al., 2010; Frijns et al., 2003; Nie, Barco, & Zeng, 2006; Plant et al., 2007; Shannon, Cruz, & Galvin, 2011; Verschuur, 2005; Weber et al., 2007). However, to be frank, there are very few studies that are well-controlled and that have exclusively evaluated the effect of stimulation rate on the performance of a large number of implant recipients. Most studies of stimulation rate and speech recognition have included a relatively small number of recipients, which limits their applicability to the implant population as a whole. Furthermore, most studies typically evaluate stimulation rate with one particular cochlear implant make, model, and electrode array. There are numerous differences in the technologies and design philosophies of the three major cochlear implant manufacturers, so it is most likely inappropriate to generalize the outcome of a study examining stimulation rate and performance with a specific implant system to an entirely different implant system from another manufacturer.

For example, the electrode contacts within the Cochlear Contour Advance electrode array are spaced less than 1 mm apart, whereas the electrode contacts within the MED-EL Standard array are spaced almost 2 mm apart. As a result, undesirable channel interaction secondary to temporal summation associated with high stimulation rates may theoretically be more likely with the Cochlear Contour Advance array than it would with the MED-EL Standard array. Because of this, higher stimulation rates may be more likely to be beneficial with the device with greater separation between electrode contacts. Furthermore, faster stimulation rates are more likely to be beneficial when the input signal is sampled at a high rate (e.g., several thousand Hz). Cochlear implant systems that are characterized by possessing lower input sampling rates may be less likely to allow for improvement in speech recognition with faster stimulation rates. Additionally, fast stimulation rates require strong current sources that are able to generate robust biphasic pulses with very short pulse widths and a brief rise time. Differences in the current sources between the three manufacturers may result in differences obtained in the benefit of faster stimulation rates between the three manufacturers.

The largest and most well-controlled study on the effect of stimulation rate on speech recognition and user preference was conducted with 71 Nucleus Freedom recipients (Balkany et al., 2007). These authors evaluated speech recognition in quiet and in noise as well as subjective preference obtained with a wide variety of stimulation rates ranging from 500 to 3,500 pps. There was no difference in speech recognition obtained between the different stimulation rates. Furthermore, recipients tended to prefer the slower rates (500 Hz to 1800 pps) over the faster stimulation rates (2,400 pps to 3,500 pps).

Also, Weber et al. (2007) compared speech recognition at three different stimulation rates (500, 1,200, and 3,500 pps) with 13 recipients who used a prototype of the Nucleus 5 sound processor. Similar to the Balkany (2007) study, Weber and colleagues (2007) reported that no improvement

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in speech recognition was observed with rate increases from 500 to 1,200 pps and from 1,200 to 3,500 pps. Plant et al. (2007) also evaluated speech recognition at different stimulation rates for 15 adult recipients using a prototype of the Nucleus 5 sound processor. These subjects typically preferred slower stimulation rates, but certain exceptions did exist in which a faster stimulation rate (2,400 to approximately 3,000 pps) was preferred. The researchers concluded that faster stimulation rates did not improve the speech recognition of Nucleus users. Again, however, it should be noted that the results of this study may not necessarily apply to Advanced Bionics and MED-EL recipients because of the numerous technical difference between the three systems.

Shannon, Cruz, and Galvin (2011) evaluated speech recognition in quiet and in noise as well as subjective preference for seven subjects using the Advanced Bionics CII cochlear implant. Performance was assessed at stimulation rates ranging from 600 to 4,800 pps. These researchers concluded that there is essentially no improvement in speech recognition when stimulation rate is increased from 600 to almost 5,000 pps.

Buechner and colleagues (2010) also evaluated speech recognition as a function of stimulation rate for 13 users of the Advanced Bionics HiResolution cochlear implant system. Performance was assessed at 1,500, 2,500, 3,000, and 5,000 pps. These authors also reported limited to no mean improvement in speech recognition as stimulation rate was increased from 1,500 to 5,000 pps. Additionally, speech recognition tended decrease when rate was increased from 2,500 to 5,000 pps. Buechner and colleagues did note that considerable intersubject variability existed in regard to the stimulation rate that yielded the best performance, which suggested that the optimal stimulation may vary on a recipient by recipient basis.

Likewise, Frijns et al. (2003) also assessed speech recognition in a group of Advanced Bionics CII recipients and concluded that increases in stimulation rates up to 1,500 pps produced improvements in speech recognition for most subjects, but as a group, further increases typically resulted in no further improvement. Again, there were some subjects who did show modest improvement at higher rates.

Nie, Barco, and Zeng (2006) evaluated speech recognition as a function of stimulation rate (ranging from 1000 to 4000 Hz) for five MED-EL implant recipients. They reported a mean improvement on word recognition and sentence recognition in noise with the faster stimulation rates. The authors also noted that there was variability in the benefit of faster stimulation rates obtained across the five subjects. Verschuur (2005) also measured speech recognition in a group of six adult MED-EL implant recipients. Performance was assessed at stimulation rates ranging from 400 to almost 2,000 pps. Verschuur (2005) noted no significant difference in the mean performance as a function of stimulation rate. However, he did note that some subjects tended to benefit from faster stimulation rates, whereas some subjects performed better with slower rates and showed no improvement with the use of faster stimulation rates.

When viewed collectively, research examining the effects of stimulation rate on speech recognition tends to indicate that most users benefit from increase in rate from a few hundred pps to about 1,000 pps, but little to no benefit occurs when stimulation rates are increased from 1,000 to 1,500 pps to 2,000 pps and beyond. However, it is important to note that individual exceptions do exist, and as a result, it may be helpful to allow recipients to experience a range of stimulation rates in order to select the rate that optimizes individual performance.

Because sound clinical decisions should be based on evidence-based research, an initial stimulation rate of approximately 900 to 1,500 pps is recommended by the authors of this textbook. The exact recommendation for an initial stimulation rate is specific to each cochlear implant system (i.e., make and model) and is provided in Chapters 4 through 6.

Regardless of the initial stimulation rate, recipients may be given the opportunity to try multiple stimulation rates, particularly if they are not satisfied or performing poorly with the initial setting. In most cases, the recipient's unsophisticated auditory-skill level with the cochlear implant at activation may not facilitate valid comparisons between stimulation rates during the initial stages of implant use. However, at some point between the 1-week and 1-month postactivation

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programming session, the user should be experienced enough to indicate which stimulation rate is preferable. During this comparison process, the clinician may allow the user to listen to programs with a range of stimulation rates (i.e., 500, 1,500, 3,000 pps). Contemporary programming software often makes it quite simple for the recipient to change stimulation rate. In many instances, a patient is able to make an unequivocal decision about the preferred stimulation rate. If a recipient is wavering between two different rates, the clinician can create a program for each rate, which will allow the user to experiment during daily situations. The recipient then can select the preferred rate at a follow-up appointment in 1 month.

It should be noted that when a recipient is performing very well with the initial stimulation rate (900 to 1,500 pps), the authors of this book typically do not experiment with other stimulation rates. Furthermore, most of the Nucleus recipients that the authors of this textbook manage are using 900 pps stimulation rates and performing quite well with that rate. Altering the rate in Nucleus recipients requires the clinician to make additional measures of stimulation level because the stimulus used with Nucleus implants is measured with a fixed pulse width. Changing the stimulus rate requires a change in pulse width, which in turn requires a remeasurement of stimulation levels. A "prediction" feature may be used to estimate the stimulation levels across the array based on a measure made on one channel in order to create a program at a new stimulus rate. In contrast, the stimulus used with Advanced Bionics and MED-EL implants is measured in charge units, so a change in stimulus rate results in an automatic change in pulse width. As a result, there is no need to make additional channel-specific measures after changing the stimulation rate. Because of this, the authors of this book are more inclined to experiment with different stimulation rates for Advanced Bionics and MED-EL recipients. Both Advanced Bionics and MED-EL have promoted the use of higher stimulation rates more so than Cochlear, and they have claimed that there are characteristics of their systems that enable the recipient to be more likely to benefit from faster stimulation rates. With that said, most Advanced Bionics and MED-EL recipients managed by the authors of this book are well-served by stimulation rates between 1,000 to 2,000 pps, and once again, most researchers have found that higher stimulation rates often do not typically provide much improvement in recipient performance.

It is often impossible for young children to provide reliable feedback about various stimulation rates. For pediatric programming, the clinician ought to begin with the stimulation rate that is best supported by clinical research or the manufacturer's recommendation. Once again, the authors prefer stimulation rates between 900 to 1,500 pps. Device-specific recommendations are provided in Chapters 4 through 6. Clinicians should experiment with stimulation rate in young children only when expected progress is not achieved.

A few additional items pertaining to stimulation rate are worth noting:

1. Increasing stimulation rate often has the effect of decreasing T levels because of temporal summation. C levels do not change nearly as much with stimulation rate increases. As a result, the recipient's electrical dynamic range will likely increase with increases in stimulation rate (Shannon, Cruz, & Galvin, 2011).
2. Increases in stimulation rate (particularly above 2,000 pps) may result in shallower loudness growth at the lower end of the electrical dynamic range, which may reduce the salience of low-level speech sounds (Kreft et al., 2004).
3. Increases in stimulation rate may enhance the likelihood of undesirable channel interaction (Middlebrooks, 2004).
4. Increases in stimulation may increase the pitch perceived by the recipient.
5. Although rare, high stimulation rates (i.e., greater than 2,000 pps) may elicit tinnitus or complaints of poor sound quality (e.g., harshness, echo).
6. Slower stimulation rates (i.e., below 1,000 pps) may be more appropriate for persons with auditory nerve dysfunction (e.g., ANSD, abnormal adaptation, cochlear nerve deficiency), persons with significant cochlear mal-

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formations (e.g., common cavity), persons with an exceptionally long duration of deafness (e.g., greater than 20 years), and the very elderly.

### Pulse Width

When the desired current level is hindered by voltage constraints of the cochlear implant system, the pulse width should be increased (i.e., widened). However, an inverse relationship exists between pulse width and stimulation rate. Higher stimulation rates require a narrower pulse width. Therefore, if a patient requires high levels of stimulation, the pulse width may have to be increased. As a result, the stimulation rate will have to be decreased. The authors of this textbook have generally found pulse widths between 21 to 40 usecs avoid voltage compliance limits by limiting the current required to reach a desired loudness percept but allow for a suitable stimulation rate (e.g., 900 to 2,000 pps). The exact manner in which stimulation rate and pulse width are managed varies slightly from one cochlear implant system to another and are discussed in detail in Chapters 4, 5, and 6.

### Channel Gain

Channel gain affects the intensity of the signal before it is processed by the external sound processor. After the clinician ensures that stimulation levels are optimized for recipients using CIS-type strategies, adjustments to channel gain may address subtle sound quality complaints or speech recognition difficulties. For instance, if upper-stimulation levels are set as high as a patient can tolerate, but he or she still complains of muffled speech, channel gains may be increased in the high frequencies. Additionally, if a recipient reports that his or her own voice is too soft and low-frequency upper-stimulation levels cannot be increased without exceeding the voltage limits of the device, the clinician may increase low-frequency channel gains to enhance the loudness. Furthermore, to improve speech recognition and comfort in noise, the clinician may create a program that includes decreased gains in the low-frequency channels. In reality, when stimulation levels are set appropriately, there is usually no need to adjust channel gains. Finally, gains should be adjusted as a last resort in *n*-of *m*-type strategies.

### Frequency Allocation

Frequency allocation is particularly relevant when one or more electrodes must be disabled. In the Advanced Bionics system, it is determined automatically and is not a clinician-adjustable parameter. However, in the Cochlear Corporation and MED-EL devices, the clinician can control how frequencies are allocated to electrodes, but significant differences exist in the way this parameter is adjusted between the two systems (see Chapter 6).

In general with contemporary cochlear implant systems, when electrodes are disabled, the bandwidth of each channel becomes wider, but the overall range of input frequencies remains the same. Wider channel bandwidths may result in the processing of two different frequencies within the same channel, which will make them sound similar. As a result, narrower channel bandwidths are preferable. According to research, restricting the upper end of the input frequency range from about 10,000 Hz to approximately 8000 Hz resulted in narrower channel bandwidths across the speech frequencies and improved speech recognition (Skinner, Holden, & Holden, 1995, 1997). Because disabling electrodes may influence frequency allocation and performance, it should be done only when necessary. Disabling may be necessary when an electrode has abnormal electrode impedance values, elicits undesired nonauditory side effects (e.g., facial nerve stimulation, pain), possesses a poor auditory response relative to other electrodes, or requires stimulation levels that are considerably higher than the rest of the electrodes within the array.

The total range of frequencies allocated for adults should extend out to at least 6000 Hz to allow for good speech recognition and sound quality across most environments. For children, it may be desirable to allocate frequencies out to at least 7000 Hz. Children often require access to higher frequency speech inputs (i.e., >6000 Hz) than adults in order to acquire optimal speech recognition and production (Stelmachowicz, Pittman, Hoover, & Lewis, 2001).

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Frequency allocation is particularly important for recipients who use electroacoustic stimulation (EAS). The reader is referred to Chapter 10 for a detailed discussion of the management of recipients who use EAS.

### Input Dynamic Range (IDR)

As previously mentioned, the IDR parameter is implemented differently across the three cochlear implant systems; thus, device-specific management is discussed in Chapters 4, 5, and 6. However, the lower end of the IDR functions similarly across the three manufacturers. The lower end of the IDR should correspond to an input level of approximately 25 dB SPL (i.e., average level for softest speech sounds) in the recipient's program for typical listening situations. The lower end of the IDR should never be increased for children. The pediatric population requires consistent access to soft speech to facilitate incidental learning, hear speech from a distance, and support optimal speech recognition. Adults may be provided with a separate program in which the lower end of the IDR is elevated by 10 to 20 dB in order to improve speech understanding and comfort in noisy environments (i.e., restaurants). Also, when necessary, an increase in the lower end of the IDR of 5 to 10 dB may address complaints of low-level ambient noise in quiet environments. Again, this adjustment is unlikely to ever be appropriate for children, as they require excellent access to low-level sounds. Additionally, adults should be encouraged to acclimate to the audibility of low-level sounds rather than adjusting parameters so that these sounds are inaudible.

### Maxima

The maxima parameter is available only in *n*-of-*m* strategies. It determines how many channels will be stimulated at one time, and the channels selected are those with the greatest amplitudes. Preferred settings vary across individuals; however, research demonstrates that nine or more channels are required for optimal performance in noise (Dorman, Loizou, Spahr, & Maloff, 2002). In practice, the maxima parameter often is set between 10 and 12.

## KEY CONCEPTS

The key information provided in this chapter should provide the reader a basic understanding of the following basic principles for cochlear implant programming:

- Detailed counseling is necessary prior to cochlear implant activation to facilitate realistic expectations with the implant, discuss postimplant appointments, discuss typical performance at activation, and to review progression of typical improvements with the implant over time.
- Following implantation, the recipient will receive a physical exam, imaging, and cochlear implant programming, which involves selection of a coding strategy, setting stimulation levels, and adjusting other parameters.
  - The provision of optimal stimulation levels is likely to be the most important parameter that the programming audiologist determines for the recipient. A variety of well-established techniques exist to determine the most appropriate stimulation levels for adult and pediatric recipients.
- Several parameters that are available and adjustable on some cochlear implant systems include stimulation rate, pulse width, channel gain, frequency allocation, IDR, and maxima.

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# 4

# Programming Advanced Bionics Implants

## ADVANCED BIONICS CPI-3 PROGRAMMING INTERFACE

All three cochlear implant manufacturers provide an interface device to allow for the connection of a recipient's sound processor to the programming computer. The Advanced Bionics Computer Programming Interface-3 (CPI-3) (Figure 4–1) may be used to program most every Advanced Bionics sound processor that is still serviced by the manufacturer. In fact, the only processors that cannot be programmed with the CPI-3 are the S-Series body-worn processors and the Platinum behind-the-ear (BTE) processor. It should be noted that the Platinum Speech Processor (PSP) body-worn processor may be programmed with the CPI-3 interface in the SoundWave 2 platform for CII and 90K recipients but not for C1 users. The CPI-3 is connected to the computer via a USB to a mini-USB cable. The USB port of the programming computer powers the CPI-3 so that it does not need an external power source.

One significant upgrade to the CPI-3 interface compared to its Advanced Bionics predecessors (mentioned later in this section describing Advanced Bionics hardware) is the fact that two sound processors may be connected to the CPI-3 at the same time. This facilitates more efficient programming of persons with bilateral cochlear implants. The CPI-3 possesses three LEDs, a blue light that indicates the device is receiving power and two green lights that indicate the device is connected to the recipient's sound processor(s). The CPI-3 may be clipped to the recipient's cloth-

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**FIGURE 4–1.** Advanced Bionics Computer Programming Interface -3 (CPI-3). *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

ing, it may be worn around the recipient's neck on a lanyard, or it may be placed on the computer desk. Advanced Bionics provides an array of specialized programming cables that may be plugged into the mini-USB ports at the top of the CPI-3 (see Figure 4–1) to allow for a connection to a range of Advanced Bionics sound processors.

## PROGRAMMING ADVANCED BIONICS SOUND PROCESSORS

Creating an effective, base program for the Naída CI Q70 sound processor is simple, but the Naída CI Q70 does possess an impressive array of sophisticated and unique hardware and signal processing options. As a result, the audiologist should be familiar with the processor and the various signal processing features of the Naída CI Q70 in order to assist the user in reaching his or her full potential with the cochlear implant(s).

The Naída CI Q70 sound processor possesses the Phonak Quest digital signal processing (DSP) chip, and as a result, it offers numerous signal processing features that are new to the cochlear implant industry. These features are discussed in the proceeding sections, and directions for programming the Naída CI Q70 to utilize these features also are provided.

### Programming Software

At the time of this writing, most Advanced Bionics sound processors are programmed using the SoundWave 2.2.25 software platform. The exceptions only apply to recipients of C1 internal devices. Specifically, clinicians may use the SoundWave 2.2.25 software platform to program the Harmony sound processor for C1 recipients, but the SCLIN software platform must be used to program all previous generation processors (e.g., S-Series and PSP body-worn processors and the Platinum BTE processor) for C1 recipients. Programming with SCLIN software is discussed later in this chapter, and Figure 4–2 provides information from Advanced Bionics regarding the necessary software for their sound processors.

| Processor Type | Implant Type: C1 | CII | HiRes 90K | HiRes 90K Advantage |
|---|---|---|---|---|
| Naída CI | - | ✓[5] | ✓[5] | ✓[5] |
| Neptune | - | ✓[1] | ✓[1] | ✓[1] |
| Harmony | ✓[2] | ✓[3] | ✓[3] | ✓[1] |
| Auria | - | ✓[3] | ✓[3] | ✓[1] |
| Platinum Sound Processor (PSP) | ✓[4] | ✓ | ✓ | ✓[1] |

[1] Requires SoundWave 2.1 or later
[2] Requires SoundWave 2.0 or later
[3] Requires SoundWave 1.4 or later
[4] Requires SClin2000 and CPI-II
[5] Requires SoundWave 2.2 or later and CPI-3

**FIGURE 4–2.** Software necessary for programming each Advanced Bionics sound processor. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

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## Patient Information Entry

After launching this application, the clinician is prompted to open the file of an existing recipient (Figure 4–3) or to create a new patient (Figure 4–4). Once the patient's personal information (e.g., name, date of birth, gender) is entered into the patient information field, then the software requires the clinician to identify and enter the ear(s) that is/are implanted. Once the implant is connected to SoundWave, the software identifies the implant type and fills in the relevant information.

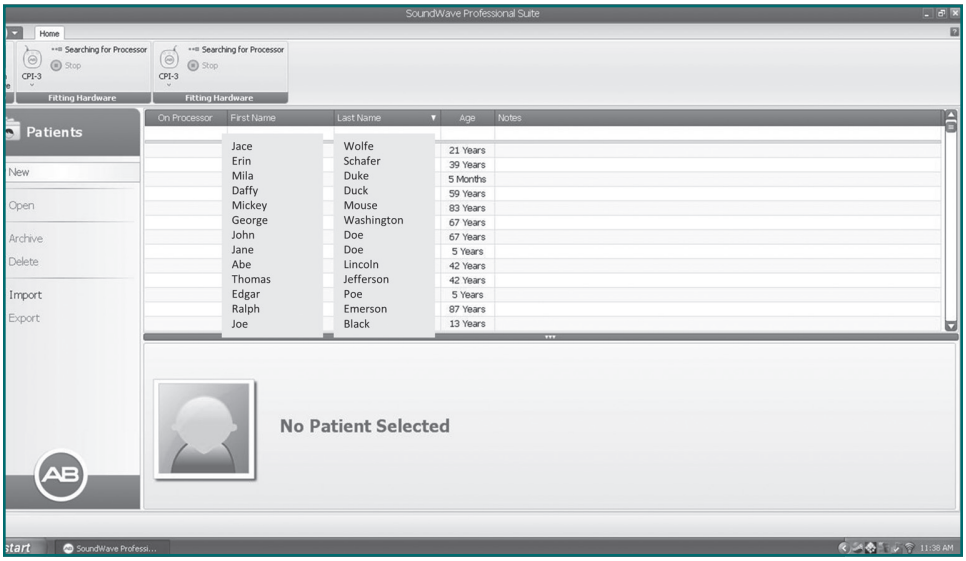
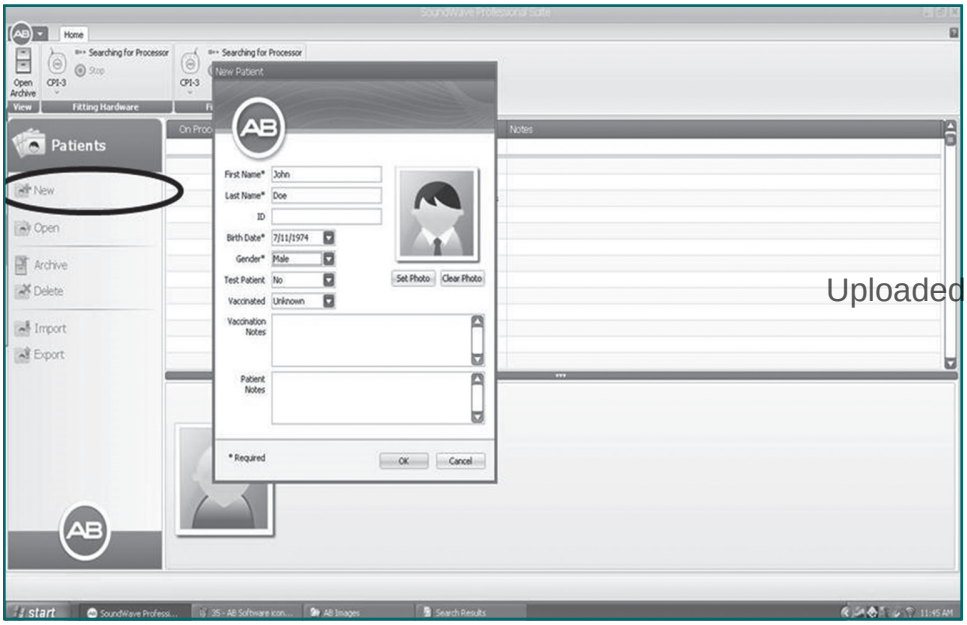



**FIGURE 4–3.** Opening the file of an existing recipient in SoundWave 2.2 software. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com




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**FIGURE 4–4.** Creating a new patient in SoundWave 2.2 software. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

It is important that the audiologist confer with the surgeon to learn the exact model of the implant and the electrode array that the recipient has received. Most new recipients will receive the HiRes 90K Advantage cochlear implant with the HiFocus Mid-Scala electrode array, but exceptions may possibly exist.

## Connecting the Sound Processor and Conducting the Electrode Impedance Measurement

The recipient's sound processor should be connected to the programming computer by using the CPI-3 computer-programming interface, which is coupled to the USB port of the programming computer, and a processor-specific cable provided by Advanced Bionics. When the CPI-3 interface is successfully coupled to the programming computer, a CPI-3 interface icon is displayed at the top of the SoundWave 2.2.25 programming screen (Figure 4–5A). The SoundWave 2.2.25 software will also display a processor icon along with a textual description of the processor when the processor is successfully coupled to the programming computer (Figure 4–5B).

Once the processor is connected, it should be placed on the recipient's ear (or body in the event that the processor is worn on the body rather than on the ear), and the external headpiece should be adhered to the implant site. When communication is established between the external sound processor and the cochlear implant (i.e., when "lock" is obtained), the SoundWave software will indicate that lock has been obtained by displaying an Advanced Bionics implant icon along with a textual description of the implant (Figure 4–5C), and the software will then automatically initiate the electrode impedance measurement (Figure 4–6). If desired, the clinician can disable the automatic impedance measurement, which would then require the clinician to manually initiate the impedance measurement.

The clinician may use the series of aforementioned icons for troubleshooting purposes. For instance, if the CPI-3 icon does not display at the top of the screen, then there is something wrong with either the USB connection (i.e., the computer did not successfully recognize the presence of the CPI-3 at the USB port), the programming computer's USB port, the USB to mini-USB cable, or the CPI-3 device itself. In most cases, this problem can be mitigated by simply removing the USB

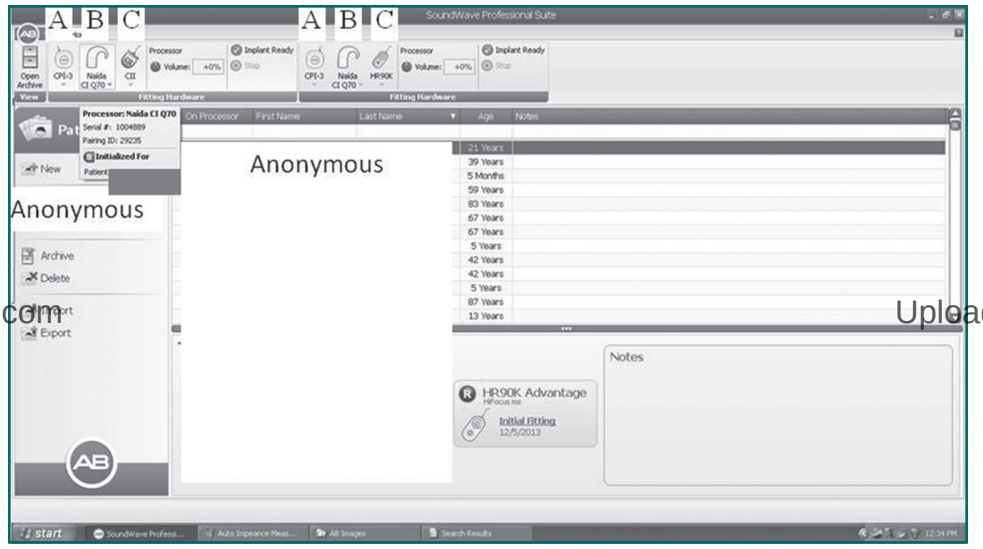



**FIGURE 4–5.** Successful connection of the: CPI-3 interface to the programming computer (**A**), processor to the programming computer (**B**), and external sound processor to the cochlear implant (**C**). *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

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**FIGURE 4–6.** Electrode impedance measurement in SoundWave 2.2 software. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

cable from the programming computer and plugging it into the same or a different USB port. If the problem persists, the clinician should "click" on the down arrow that resides next to the "AB" icon in the upper left-hand corner of the screen. This arrow will open a drop-down menu, which includes the option of "Fitting Hardware." The clinician can open the "Fitting Hardware" feature and select "Auto Configure Ports" (Figure 4–7). This should enable the CPI-3 device(s) that are connected to the programming computer.

When the processor icon does not display in the SoundWave software, the clinician should ensure that the Naída CI Q70 processor is firmly attached to the programming connector. If a Harmony processor does not successfully connect to the computer, the clinician should "wiggle" the programming connector that couples to the processor. Many times, this will facilitate a connection. The clinician may also inspect the contact points at the base of the Harmony processor. If the connection points are obstructed with debris or oxidation, the clinician may attempt to remove the debris (with a brush and/or alcohol swab) and retry connecting to the processor. If the software continues to show that no connection has been established to the processor, the clinician should attempt to connect to the recipient's backup sound processor or to a clinic loaner processor or to use a different programming cable.

Finally, when the software indicates that a connection has been established with the CPI-3 interface and sound processor, but not with the cochlear implant (e.g., internal device), the clinician should replace the cable and headpiece. If the problem persists, then the clinician may need to replace the sound processor, as it is possible that the cable port on the sound processor is faulty. If the problem persists, the clinician should contact the manufacturer for assistance in troubleshooting a possible flaw with the cochlear implant.

The automatic measurement of electrode impedance is a unique feature of the Advanced Bionics SoundWave software, as the other implant manufacturers require the clinician to manually start the impedance measurement. The clinician may also select the "Impedances" tab to review the results of the impedance test or to measure impedances again (Figure 4–8). Afterward, the results of the impedance measurement will be displayed in a table at the base of the programming screen (Figure 4–9), which is available when the clinician launches a new or existing program to begin the programming process.

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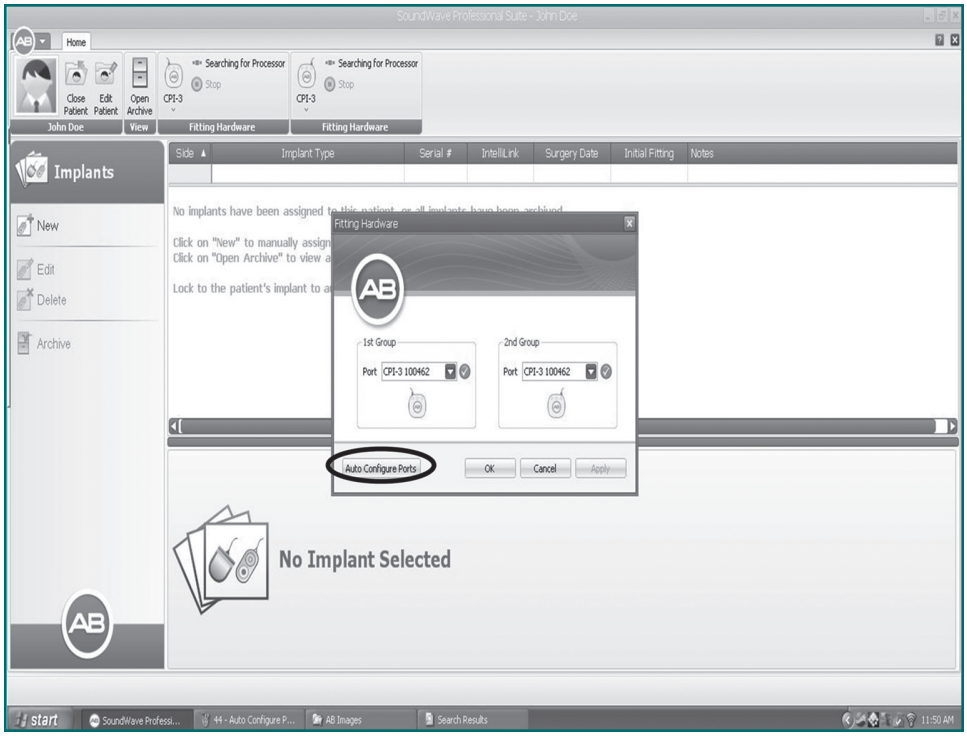
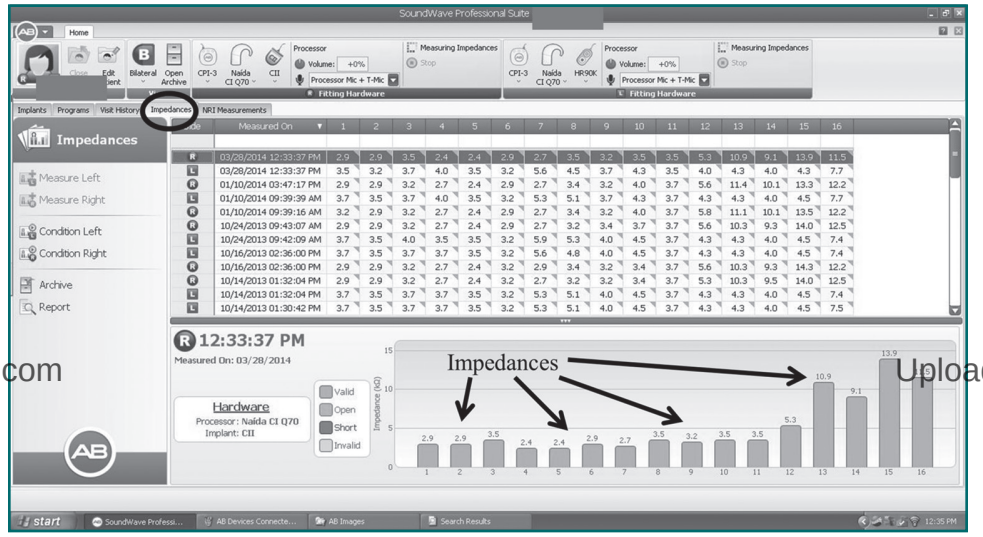



**FIGURE 4–7.** Selection of "Fitting Hardware" feature (*left*) and "Auto Configure Ports" option (*right*). *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com




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**FIGURE 4–8.** Selection of "Impedances" tab to review the results of the impedance test or to remeasure impedances. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

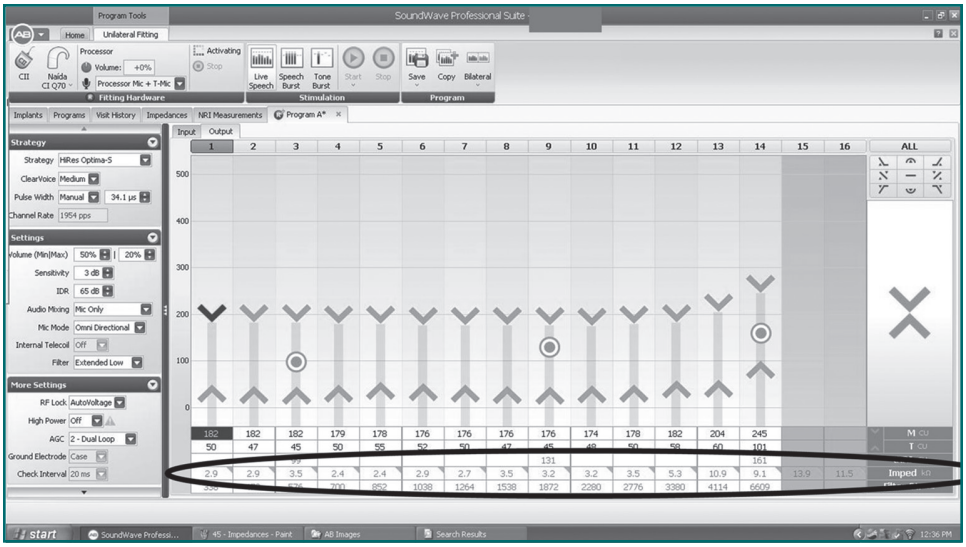



**FIGURE 4–9.** Results of the impedance measurement will be displayed in a table at the base of the programming screen. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

Advanced Bionics impedances are measured in the monopolar electrode coupling mode, with impedance measurements made on all 16 intracochlear electrodes with the case ground electrode serving as the terminal point of the return path. The normal range of electrode impedance values in the Advanced Bionics CII and 90K implants ranges from 1.0 to 30.0 kOhms and in the C1 implants ranges from 1.0 to 150 kOhms. Electrode impedance results are depicted in bar graph form in the electrode impedance tab measurement field. Electrodes with normal impedances are displayed as green bars indicating that they are serving as "valid" functioning electrodes (see Figure 4–8). "Open" electrodes possess impedance values that are higher than the normal range and are displayed in yellow on the bar graph. As mentioned before, when electrode impedances are slightly outside of the normal range (e.g., within 10 kOhm of the upper limit of normal) at implant activation or after a long period of non-use, it is recommended that the clinician provide electrical stimulation to the electrode contacts for a few minutes and then readminister the impedance test. "Short" electrodes are defined either as a single electrode with an impedance value between 0.5 and 1.0 kOhms or as an electrode having a significant electrical interaction with one or more other electrodes. Electrode shorts are displayed in purple in the bar graph display. Finally, electrodes with impedances below 0.5 kOhms are considered to be "invalid" and are displayed in the color gray in the bar graph display. Short electrodes should definitely be disabled and excluded from use in the recipient's program. Open and invalid electrodes should also be disabled if they are persistently outside of the normal impedance range.

Advanced Bionics SoundWave software utilizes a form of electrical field imaging and modeling to identify electrode shorts. With the electrical field imaging technique, a small amount of current is sent to an electrode via monopolar electrode coupling, and the resultant voltage that is present at every other electrode in the array relative to the stimulated electrode/monopolar ground connection is measured and compared. Shorted electrodes possess abnormally low electrode impedance values and will show a measured value that is close to that of the stimulating electrode.

Within the impedance tab, the clinician also may perform electrode conditioning to eliminate any substances present at the electrode contact (i.e., protein deposits, air bubbles) through the application of a low-level current. These unwanted

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substances typically cause elevated electrode impedances at initial activation of the implant; however, impedances almost always decrease over time with consistent stimulation. The clinician should consider conducting electrode conditioning after a prolonged period of non-use of the implant, particularly if the electrode impedances are relatively high (e.g., greater than 10 kOhms).

## Creating a Program in SoundWave 2.2 Software

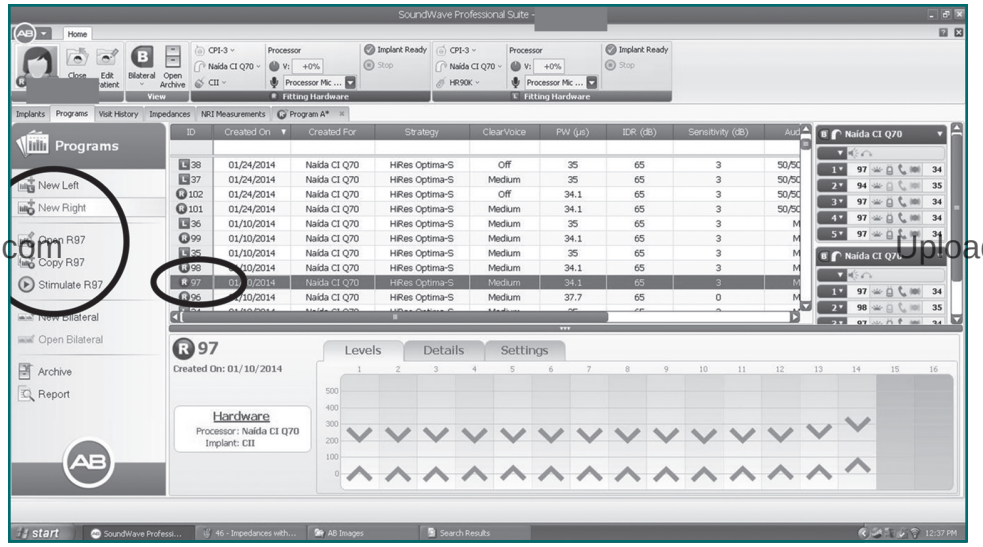
As shown in Figure 4–10, the clinician must select the "Programs" tab in the SoundWave 2.2.25 software to create a program for the recipient. A new program may be created by selecting the "New Right," "New Left," or "New Bilateral" options on the left of the screen while in the "Programs" platform or by selecting a program that has been created earlier in the session or during a previous programming session. The clinician may select an old program by simply "double-clicking" on the program number within the list of saved programs.

Once in the programming screen, the clinician is able to view a "programming ribbon bar" that stretches across the top of the programming screen (Figure 4–11). The CPI-3, processor, and implant icons are located at the left-hand side of the programming ribbon bar. Each has a drop-down menu that accesses special options for each component. For instance, the clinician may identify and install the fitting hardware (e.g., CPI-3) and initialize or backup a sound processor.

## Selecting a Sound Processing Strategy

The first step of creating a new program is to select the signal coding strategy the recipient will use. The clinician may select from a number of signal coding strategies available within SoundWave 2.2.25 including: HiRes-P (HiResolution Paired), HiRes-S (Sequential), HiRes-P with Fidelity 120, HiRes-S Fidelity with Fidelity 120, HiRes Optima-S, HiRes Optima-P, CIS (Continuous Interleaved Sampling), and MPS (Multiple Pulsatile Sampling) (see Chapter 2 for descriptions of coding strategies).

The software default signal processing strategy is HiRes-P. However, for users of the Naída CI Q70, Neptune, and Harmony processors, the authors recommend using with HiRes Optima-S signal coding strategy as the Optima strategy plays an important role in optimizing battery life for Naída CI Q70 users. The HiRes-S signal coding strategy is recommended over the HiRes-P strategy because the stimulation rate per electrode stimulation rate should be more than sufficient




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**FIGURE 4–10.** Creating a new program or opening an existing program while in the "Programs" platform. *Note.* Image provided courtesy of Advanced Bionics.

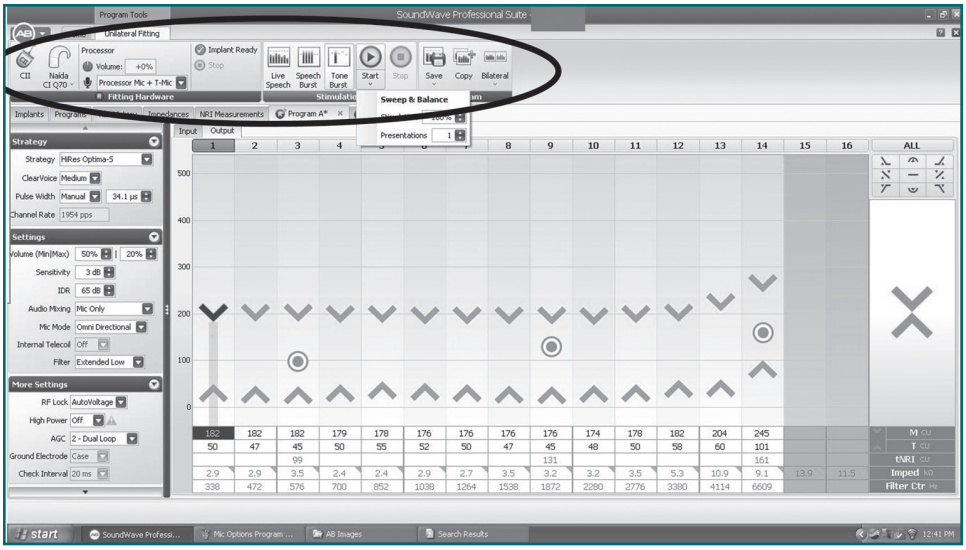



**FIGURE 4–11.** "Programming ribbon bar" that stretches across the top of the programming screen. *Note.* Image provided courtesy of Advanced Bionics.

for most recipients. Furthermore, the sequential nature of electrode stimulation of the HiRes-S variants theoretically reduces the chance for deleterious channel interaction when compared to the partially simultaneous stimulation provided by the HiRes-P strategies.

Clinical trials with Fidelity 120 suggest that most users experience better sound quality for speech and music when compared to conventional HiRes sound processing (Advanced Bionics Corporation, 2009). Finally, the authors recommend manually setting the pulse width for the HiRes-S Fidelity 120 and the HiRes Optima-S strategies to 37.7 microseconds. The wider pulse width of 37.7 microseconds allows for a per-electrode stimulation rate between 1,500 to 2,000 pps, which should be more than sufficient considering that most studies have shown increases in per-electrode stimulation rates up to about 1,500 pps result in improvement in speech recognition for most subjects with performance becoming variable at higher stimulation rates (Wilson, 2006; see Chapter 3 for more detail).

Exceptions to the authors' recommendation include cases in which the Optima signal coding strategy calls for a pulse width greater than 37.7 microseconds. In such a case, the authors recommend that the clinician employ the wider pulse width suggested by the software. It should be noted that in default mode (as described in an upcoming section), the SoundWave software automatically manages the pulse width to avoid voltage compliance issues, and many users receive satisfactory outcomes while using the pulse width and stimulation that is automatically determined by the SoundWave software. Finally, at the time of this writing, the Optima and Fidelity 120 strategies are not approved by the FDA for use with children.

Previous clinical trials show substantial variation in user preferences across the aforementioned signal coding strategies. For this reason, the clinician may want to provide users with the opportunity to listen to each strategy. The authors contend that most recipients achieve satisfactory performance with the signal coding strategies that were recommended in the preceding paragraph. Furthermore, the authors' anecdotal experience suggests that it is preferable to focus on determining the ideal stimulation levels for a recipient during the first few weeks of implant use, as most recipients will do quite well with contemporary signal coding strategies when stimulation levels are optimized. However, for subjects who are struggling with sound quality complaints and/or poor speech recognition after at least one month

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of use, the clinician should attempt to identify whether alternative signal coding strategies and/or stimulation rates may allow for an improvement in recipient performance. As previously discussed, most recipients perform quite well with the Optima and HiRes Fidelity 120 signal coding strategies as these modes are highly sophisticated and represent the most advanced strategies currently available in the Advanced Bionics system (Advanced Bionics, 2012; Büchner et al., 2012; Dunn, Tyler, Witt, & Gantz, 2006). However, the authors do manage some recipients who seem to perform better with the legacy signal coding strategies (e.g., CIS/MPS).

With the emulated CIS and MPS signal coding strategies, every other channel is disabled (typically every even-numbered electrode). It is possible that the reduced number of channels in the older signal coding strategies may limit channel interaction, which may be beneficial for a small subset of recipients. Furthermore, it should be noted that a wider pulse width and slower stimulation rate is employed with these legacy signal coding strategies (e.g., 829 pps for CIS), which also may be beneficial for a small proportion of recipients. Again, it should be noted that use of emulated legacy programs (e.g., CIS and MPS) for CII and 90K recipients is the rare exception rather than the rule. Ultimately, clinical judgment should be used to determine whether alternative signal coding strategies should be introduced, and if so, to determine the appropriate time to introduce the different signal coding strategies.

It should be noted that C1 recipients using the Harmony processor may be programmed in the SoundWave 2.2.25 software platform, but they may only be programmed with the Harmony sound processor. C1 recipients using older processors must be programmed in the SCLIN software platform. C1 Harmony wearers must use one of the original Advanced Bionics signal coding strategies: Simultaneous Analog Stimulation (SAS), CIS, or MPS. Most of these recipients will have used SAS, CIS, or MPS for many years, and as a result, it is not common practice to deviate from their customary signal coding strategy (i.e., it is not typically recommended to switch the CIS if the recipient has always used MPS or SAS).

### *ClearVoice*

ClearVoice is a speech-enhancement strategy that intends to improve speech perception in noise without compromising speech understanding in quiet. It is an algorithm that is integrated into HiRes Fidelity 120 sound processing and makes channel-specific determinations of whether the input signal is mostly noise or contains useful information for understanding speech. ClearVoice attenuates the "noisy" channels while leaving the information-bearing channels relatively unaffected. The result is an improvement in the overall signal-to-noise ratio (SNR).

A clinician selectable ClearVoice gain setting determines the maximum channel-specific attenuation that is applied in a given program—up to 6 dB of attenuation for Low, up to 12 dB of attenuation for Medium, and up to 18 dB of attenuation for High. This selection can be programmed to meet the individual requirements and personal preference of each user.

The ClearVoice algorithm may be used with the HiRes Optima and HiRes Fidelity 120 signal coding strategies and may be selected for the Harmony, Neptune, and Naída CI Q70 sound processors. For implementation of ClearVoice, the input signal is analyzed using a fast Fourier transformation (FFT), and the spectrum is divided into 15 bands across the 16 intracochlear electrode contacts. Then, a Hilbert transform is used to extract the temporal information from the speech signal. An SNR estimator is used to provide an ongoing estimate of the background noise level in each of the 15 spectral bands. Additionally, an estimate is determined of the speech and noise present in each of the 15 bands. Next, the instantaneous energy is measured in each of the 15 channels and compared to the ongoing noise level within each channel. For channels in which the instantaneous energy is similar to the ongoing noise level, a determination is made that the channel primarily contains noise, and the level (i.e., gain) of that channel is attenuated by the amount indicated by the ClearVoice setting (e.g., Low, Medium, or High). If the instantaneous energy present within a channel is higher than the ongoing noise, then the input is considered to be speech, and little to

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no attenuation is provided. For channels that are deemed to primarily consist of noise, the amount of attenuation is determined by the SNR within a channel as well as the ClearVoice setting selected by the clinician. ClearVoice settings of Low, Medium, and High result in a maximum attenuation of 6, 12, and 18 dB, respectively. This selection can be programmed to meet the individual requirements and personal preference of each user. Furthermore, recipients may be equipped with multiple programs, each possessing different strength of ClearVoice. From the point in which the sound processor is activated with a ClearVoice program, approximately 1.6 seconds of signal analysis is required before ClearVoice begins to attenuate channels with an unfavorable SNR.

An FDA clinical trial showed that ClearVoice significantly improved speech understanding in speech-spectrum noise and multitalker babble, did not compromise listening in quiet, and was preferred for everyday listening (Koch, Quick, Osberger, Saoji, & Litvak, 2014). Of historical note is that this trial represented the first time a regulatory superiority claim had been met for a new cochlear-implant sound-processing algorithm compared to existing strategies (ClinicalTrials.gov identifier: NCT01066780).

## Setting Stimulation Levels and Stimulation Rate

In the SoundWave platform, stimulation levels are measured in charge units. Charge units are the product of the amplitude of the stimulus current, the pulse width, and an arbitrary constant (k = .013). This unit of stimulation represents a constant charge, and its value is always the same, regardless of the pulse width. In other words, if the pulse width is increased, the current is decreased to maintain the same charge value. Specifically, when voltage compliance is approached or reached, and increases in stimulus current amplitude can no longer be provided, the pulse width is automatically increased by the software. This frees the clinician from having to monitor the pulse width to ensure stimulation remains within voltage compliance.

Specifically, the SoundWave software provides three options for adjusting the pulse width including *APWI* (Automatic Pulse Width I), *APWII* (Automatic Pulse Width II), and *Manual*. *APWI* is the default selection for the HiRes and HiRes with Fidelity 120 signal coding strategies. In the *APWI* mode, the software automatically determines the current amplitude and pulse width required to yield a charge unit level that is within voltage compliance for the implant. One objective of the *APWI* algorithm is to use the shortest pulse width that satisfies voltage compliance requirements, so that the fastest possible stimulation rate may be provided. The exact function of the *APWI* algorithm is proprietary, but does allow for stimulation on a small number of electrodes to be outside of voltage compliance. Typically, these electrodes will possess an electrode impedance or M level (e.g., upper-stimulation level) that is markedly higher than the rest of the electrodes in the array.

*APWII* is the default mode for the HiRes Optima signal coding strategy but may also be used with HiRes and HiRes Fidelity 120 strategies. *APWII* also automatically manages pulse width and current amplitude, but a stricter algorithm is employed so that stimulation on every electrode (even those with outlier impedance or upper-stimulation levels) is within voltage compliance. The software determines the shortest pulse width that provides stimulation within voltage compliance limits while also maximizing stimulation rate. The *APWII* algorithm plays an important role in optimizing battery life for the Naída CI Q70 sound processor, and as a result, the *APWI* mode is unavailable for use with the HiRes Optima signal coding strategy. However, the clinician may select the *Manual* pulse width mode for HiRes Optima users. When the *Manual* mode is selected, the pulse width typically must be equal to or greater than the pulse width selected by the *APWII* mode.

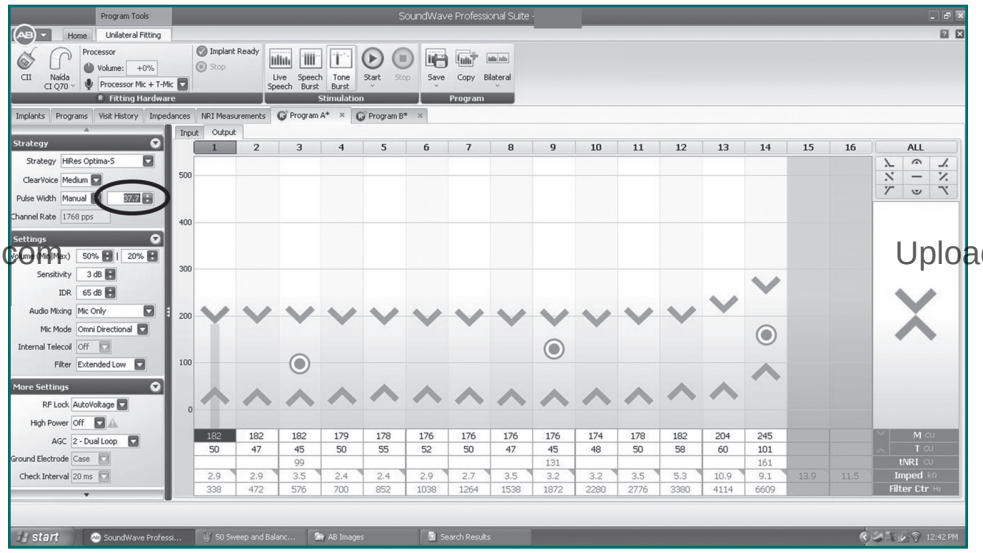
It should be noted that the *APWII* mode is implemented somewhat differently for the Optima strategies versus other Advanced Bionics signal coding strategies. For the Optima strategy, the *APWII* mode employs a 4-volt target to manage the pulse width required to remain within voltage compliance. In contrast, for the HiRes and HiRes Fidelity 120 strategies, the *APWII* mode

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employs an 8-volt target to determine the pulse width required to remain within voltage compliance. The lower voltage target used with the Optima strategy is designed to improve power efficiency, which should allow for longer battery life. This is especially important for the Naída CI Q70 sound processor, which possesses a small battery designed to enhance comfort, retention, and cosmetics. It should also be noted that the lower voltage target used for Optima strategies is one reason why the Optima may require a longer pulse width than the HiRes and HiRes Fidelity 120 strategies.

The *Manual* pulse width mode may also be selected for the conventional HiRes and HiRes with Fidelity 120 signal coding strategies. In fact, as mentioned earlier, for all HiRes signal coding strategies (HiRes, HiRes with Fidelity 120, and Optima), the authors recommend manually setting the pulse width to 37.7 microseconds (or higher if the software indicates that a 37.7 microsecond pulse width is insufficient to maintain voltage compliance). The processor stimulation rate (Figure 4–12) is tied to the pulse width, and the wider pulse width of 37.7 microseconds allows for a per-electrode stimulation rate between 1,500 to 2,000 pps, which should be more than sufficiently fast for most recipients. Again, a review of published research studies examining cochlear implant stimulation rate suggests that increases in per-electrode stimulation rates up to about 1,500 pps result in improvement in speech recognition for most subjects with performance becoming variable at higher stimulation rates (Wilson, 2006; see Chapter 3 for more detail). Furthermore, the wider pulse width (37.7 microseconds) allows for lower current amplitude, which may limit channel interaction and nonauditory side effects (e.g., facial nerve stimulation). The authors have found this combination of pulse width (37.7 microseconds) and stimulation rate (1,500–2,000 pps) to provide satisfactory performance for most recipients. However, if recipients are struggling with poor sound quality and/or speech recognition after more than 4 weeks of implant use, the authors will manually change the pulse width to trial slower and faster stimulation rates. It should be noted that the software will display a warning icon (Figure 4–13) when the selected pulse width results in stimulation levels on some electrodes being outside of voltage compliance limits. In this case, the clinician should increase the pulse width.

Finally, the default pulse width for CIS and MPS signal coding strategies is 75.4 microseconds. This wider pulse width was necessary when CIS and MPS were used with first generation implants




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**FIGURE 4–12.** Setting pulse width. *Note.* Image provided courtesy of Advanced Bionics.

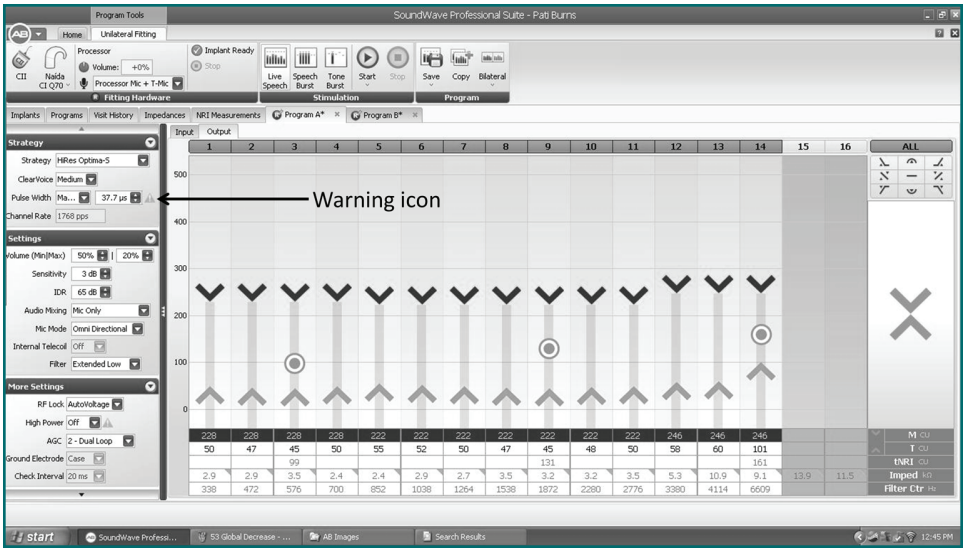



**FIGURE 4–13.** Warning icon to indicate that pulse width is resulting in electrode stimulation levels outside voltage compliance limits. *Note.* Image provided courtesy of Advanced Bionics.

because the current sources of these implants were not sufficient to provide a biphasic rectangular pulse with a large enough of a current amplitude to yield satisfactory loudness growth. The clinician may manually switch the pulse width from 75.4 microseconds to either 150.9 or 226.3 microseconds. It should be noted that 75.4 microseconds is more than sufficient for most CIS/MPS users. Exceptions may arise for users who have high stimulation levels secondary to auditory abnormalities, such as a cochlear deformity (e.g., Mondini's or cochlear duct ossification associated with bacterial meningitis) or an aplastic (deficient) auditory nerve.

### Stimulus Units

The range of stimulation levels provided by the software is 0 to 6,000 charge units, but M levels (i.e., upper-stimulation levels) for most recipients do not exceed 300 charge units (in fact, most recipients have stimulation levels between 100 and 200 charge units). Stimulus level grows linearly in the SoundWave software, so the recipient's loudness percept will typically grow at a slower rate than what is observed in the Cochlear CustomSound programming platform, in which the stimulus increases logarithmically. For this same reason, the electrical dynamic range (e.g., the range between electrical threshold and upper level of stimulation) in Advanced Bionics charge units is typically wider relative to what is typically observed in Cochlear recipients. For more information for typical stimulation levels for children and adults in Advanced Bionics devices, the reader should refer to Zwolan and Overstreet (2005). However, the clinician should carefully review the Zwolan and Overstreet (2005) data, as it was collected with different electrode arrays (1j electrode) and signal coding strategies (conventional HiRes) compared to what are typically used today (at the time of this writing, the most contemporary options are the mid-scala electrode array and the HiRes Optima signal coding strategy). As a result, contemporary users may require different stimulation levels for optimal performance.

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### Measuring T Levels

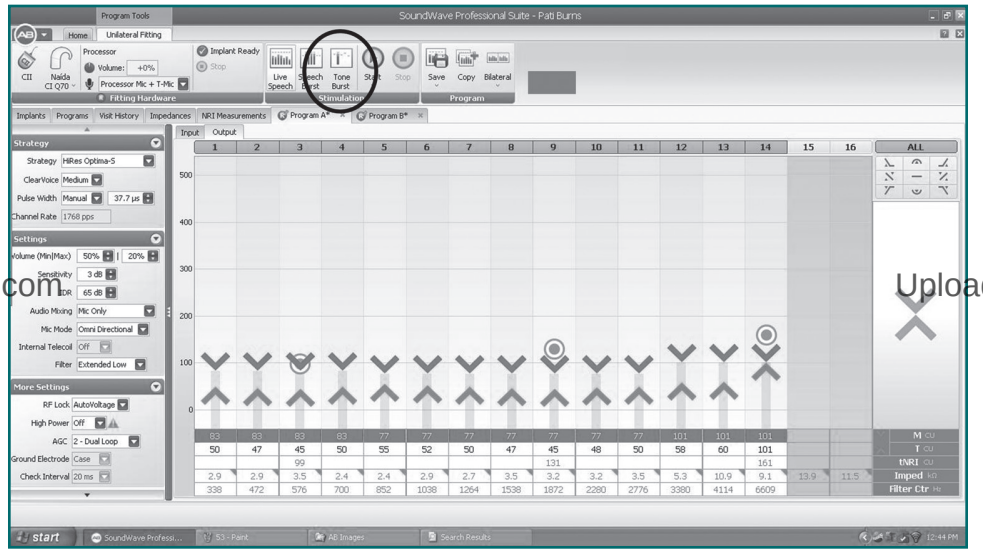
In the default programming mode, the clinician is not required to measure T levels to create programs. Instead, the T level is automatically set to 10% of the established M-level value. In most cases, this estimation of T level allows recipients

to access low-level sounds, and performance is similar between estimated and measured T levels. The clinician does, however, have the option to set T levels manually on a single electrode basis or by adjusting T levels globally. T levels may be measured on individual electrodes by selecting the "Tone Burst" stimulus in the "Stimulation" pane at the top of the programming screen (Figure 4–14). Additionally, the clinician may "unlink" M and T levels and then adjust them as previously described, or one may simply click and drag any individual T-level arrow icon and it will automatically "unlink" to allow T levels to be manually set.

Then, the clinician simply has to use the mouse of the programming computer to "click" on the T-level icon of the channel to be measured. Stimulation levels may be adjusted by using the up/down arrow keys of the keyboard of the programming computer or by selecting the T-level or M-level field and typing in a desired charge unit number and hitting "enter" on the keyboard.

The clinician may also choose to make global adjustments of T levels. This may be done by "clicking" and "dragging" the Global T level adjuster or by clicking on the T level icon and then selecting the "All" option in the upper right corner of the programming screen. Subsequent adjustments will alter all T levels across the electrode array. A global increase in T-level stimulation provides more stimulation for low-level sounds, which may be helpful if the recipient has elevated sound-field detection thresholds or if recognition of soft speech is inadequate. A global decrease in T levels is occasionally provided to eliminate the recipient's percept of a constant low-level noise. All HiRes signal coding strategies are advanced iterations of the CIS signal coding strategy. For all CIS-type signal coding strategies, recipients may complain of a low-level audible noise (often described as frying, hissing, or buzzing) when T levels are too high. In this case, the clinician should confirm that the recipient is not simply hearing ambient noise in the room (i.e., the building's heating or air conditioning system). This may be determined by asking the recipient to step into an audiometric sound booth. If the low-level noise ceases, the recipient is hearing ambient room noise and should be encouraged to acclimate. If the sound persists, then it is likely that T levels are too high.

T levels should be measured when the implant equipment (i.e., sound processor microphone) is functioning within normal limits and sound-field detection thresholds are elevated (i.e., child ≥25 dB HL; adult ≥30 dB HL). The clinician




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**FIGURE 4–14.** Measure T levels on individual electrodes by selecting the "Tone Burst" stimulus in the "Stimulation" pane. *Note.* Image provided courtesy of Advanced Bionics.

may particularly want to consider measuring T levels for children because previous research emphasizes the importance of establishing appropriate T levels in order to provide consistent audibility of low-level sounds for young children (Holden, Reeder, Firszt, & Finley, 2011). For measuring T levels in children, the authors of this book suggest a 5-charge unit-down and 10-charge unit-up approach with tone-burst stimuli. Young children are likely to provide a minimal response level instead of a true threshold; therefore, after measuring T levels the clinician may consider a small global reduction (the authors typically use a 10 charge unit reduction) to lessen the possibility of child perceiving constant low-level electrical noise.

### Measuring M-Level Stimulation

The default method of measuring M levels is unique in the Advanced Bionics programming platform. Instead of presenting biphasic pulses to individual electrode contacts, a Speech Burst stimulus is used. The Speech Burst consists of white noise filtered through three or four channels and is presented at the user 's typical stimulation rate. There are three theoretical advantages to using Speech Burst stimuli over conventional single-channel programming stimuli. First, presentation of the stimulus across multiple channels simultaneously provides a more valid representation of the spectral characteristics of complex speech and environmental sounds when compared to presenting on only one channel. Second, simultaneous presentation on multiple channels more closely resembles expected summation across channels in live speech mode and accounts for loudness summation (i.e., M levels may be more accurate). Third, simultaneous presentation across several channels is potentially more time efficient than single-channel measurements. Furthermore, Advanced Bionics' research studies show similar performance when using programs created using single-channel versus Speech Burst measures (Advanced Bionics, 2003). It should be noted, however, that use of Speech Burst stimuli may over- or underestimate stimulus levels for a given channel if the requisite level for that channel is substantially higher or lower than the other channels that make up the Speech Burst.

In the default mode, clinicians measure M levels for each of the four Speech Burst bands (i.e., Speech Bursts at channels 1–4, 5–8, 9–12, and 13–16) in order to evaluate across the electrode array. After M levels are measured for Speech Bursts, the clinician should decrease M levels globally and activate the program in "Live Speech" mode, which allows the clinician to make channel-specific or global (across all channels simultaneously) adjustments to stimulation levels. In live speech mode, M levels should be increased globally until the recipient reports that speech and environmental sounds are perceived at the most comfortable listening level (Figure 4–15). After these adjustments, the clinician should consider balancing the loudness at M levels on a channel-by-channel basis across the electrode array. Loudness balancing is typically conducted from the apical end of the array toward the basal end. Again, the act of loudness balancing at M levels typically results in a program that provides satisfactory speech recognition and sound quality for the user. Balancing at M level will typically flatten or smooth the M-level profile and account for any channel-specific changes that are not captured by the use of Speech Burst stimuli. If the recipient reports unsatisfactory sound quality or speech perception after the aforementioned M-level adjustments, then the clinician may fine-tune the M levels per the recipient's report (i.e., decrease high-frequency M levels if the recipient reports that speech is too "tinny"). Fine-tuning stimulation levels is discussed in detail in Chapter 7.

It should be noted that additional adjustments may be made while in live speech mode. First, as shown in Figure 4–16, the clinician may select different microphone configurations: T-Mic 2, processor microphone only, a combination of the two, headpiece microphone, omnidirectional, UltraZoom, and so forth. Additionally, the clinician may select different volume control settings, the telecoil input, ComPilot streaming, ComPilot mixing, and so forth. These adjustments are easily accessible to the clinician, so a variety of program settings and sound input options may be trialed during a programming session to determine which options best meet a recipient's needs and/or to demonstrate to the recipient the proper function of various features.

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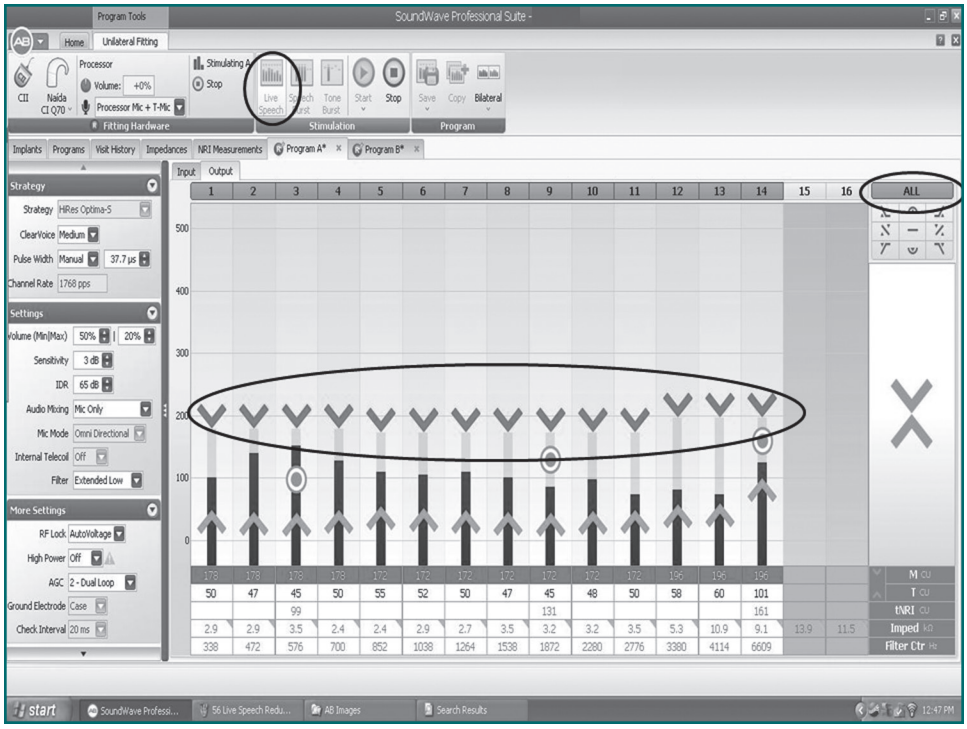
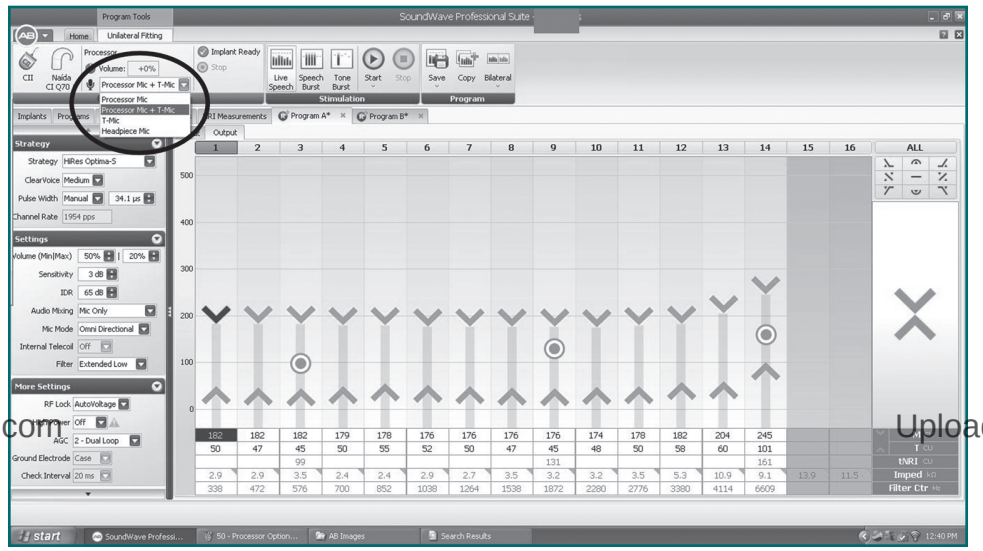



**FIGURE 4–15.** Global adjustments to M levels. *Note.* Image provided courtesy of Advanced Bionics.




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**FIGURE 4–16.** Selection of different microphone configurations. *Note.* Image provided courtesy of Advanced Bionics.

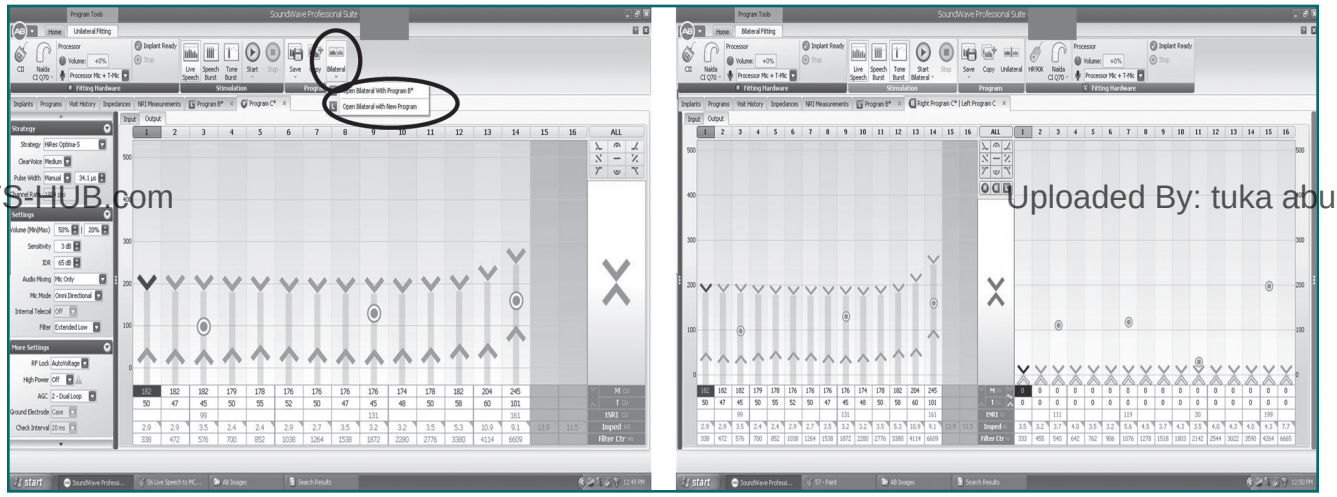
Furthermore, it should be noted that M levels also may be measured with single-channel tone burst stimuli. The clinician simply selects the tone burst option and obtains M level on a single channel using the same keyboard/software function previously described for the measurement of

single-channel T levels. Additionally, some clinicians choose to increase M levels globally (set at a flat profile) in live speech mode to the recipient's most comfortable listening level. Then, channel-specific (or band-specific) adjustments are made to address sound quality complaints of the recipient or to improve speech clarity. The reader is directed to Chapter 7 for additional discussion on setting stimulation levels for cochlear implant recipients.

As mentioned in Chapter 3, upper-stimulation levels are closely related to the electrically evoked stapedial reflex threshold (ESRT); therefore, this may be the ideal and most objective approach for setting M levels in some recipients (i.e., young children). The ESRT also is highly correlated to measured M levels when both were conducted with Speech Burst stimuli (Buckler & Overstreet, 2003). In addition, M levels based on Speech Burst ESRT may allow for better speech recognition than programs created via psychophysical loudness scaling (Wolfe & Kasulis, 2008). In many cases, recipients will prefer M levels that approach but do not reach their ESRT.

To establish M levels based on ESRT, the clinician will (1) measure the ESRT across the electrode array (either with Speech Bursts or tone bursts), (2) decrease the M levels globally to a level that is soft to the recipient, (3) activate the sound processor in live speech mode, and (4) gradually increase M levels until the recipient indicates that the speech signal is most comfortable (see Chapter 3 for additional information). For children who are unable to provide verbal feedback about the signal, M levels should typically be set within 5 to 10% of their ESRT unless the recipient shows signs of discomfort. Of course, in that case, M levels should be decreased to a level that optimizes comfort, sound quality, and performance. For all recipients, it is unlikely that the M level will exceed the ESRT. In most cases, Speech Bursts are excellent stimuli for eliciting the ESRT; however, if the ESRT is variable and inconsistent in charge units across the electrode array (i.e., ESRT to Speech Burst for electrodes 9–12 is substantially higher than 13–16), the clinician should consider measuring it with tone-burst stimuli.

Finally, it should be noted that stimulation levels may also be adjusted for both ears simultaneously when programming bilateral cochlear implant recipients. The clinician simply opens a program for one ear and then "clicks" on the "Bilateral" icon in the ribbon bar at the top of the programming screen (Figure 4–17). This allows the clinician to open an existing program for the other ear or a new program. Using the selected programs, the clinician may provide live speech stimulation for each ear simultaneously and make channel or global adjustments to T and M levels while in bilateral live speech mode.



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**FIGURE 4–17.** Simultaneously adjust bilateral sound processors by clicking on the "Bilateral" icon in the ribbon bar at the top of the programming screen. *Note.* Image provided courtesy of Advanced Bionics.

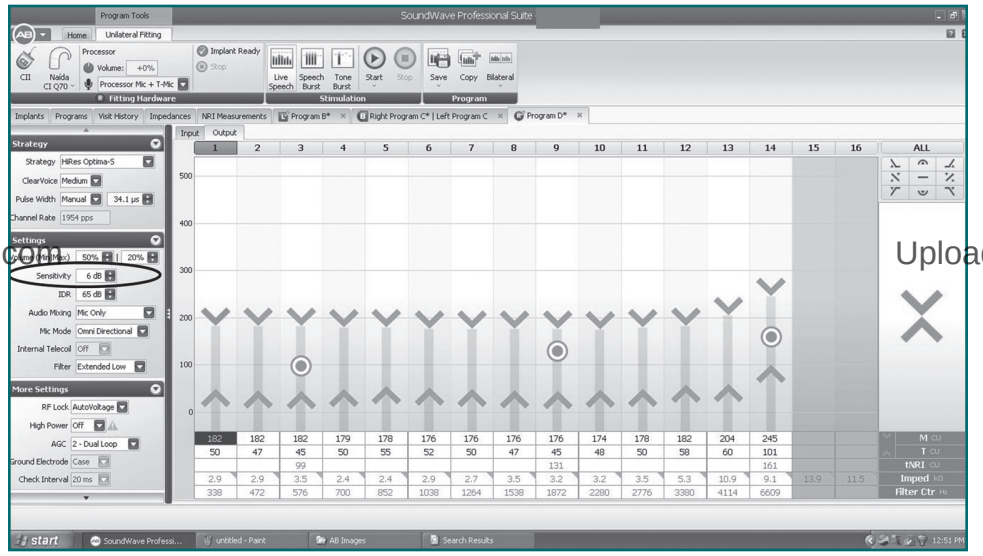
## Additional Programming Parameters

### *Microphone Sensitivity and Volume Control*

The sensitivity of the sound processor can be adjusted in the SoundWave 2.2.25 software, but typically, it should remain at the default setting. Essentially, the sensitivity controls the gain (±10 dB) applied to the signal as it is relayed from the microphone to the bandpass filters. Adjustments in sensitivity affect all frequencies equally. An increase in sensitivity will improve access to low-level sounds, but may degrade hearing in noisy environments. A decrease in microphone sensitivity may improve comfort in noise, but also will diminish audibility for low-level sounds (e.g., 20 to 25 dB SPL). An example of an instance in which the clinician may choose to alter the sensitivity setting would be for a recipient who does not hear low-level sounds well but for whom an increase in T levels produces an audible noise (e.g., frying or buzzing noise). In such a case, a small increase in the sensitivity setting (e.g., +3 dB) may improve access to low-level sounds without the addition of extraneous noise. The sensitivity setting within a program may be adjusted by clicking on the up/down arrow buttons that reside to the right of the window displaying the sensitivity setting (Figure 4–18) or by inserting any whole number between –10 and +10 dB into the window next to the "Sensitivity" control and hitting the "enter" button on the keyboard. Sensitivity of the Naída CI may also be adjusted manually via the AB myPilot remote.

Neptune processor users may also manually adjust the sensitivity on the processor. However, the automatic gain control featured in all Advanced Bionics processors is quite effective at adjusting to a wide range of input levels to capture the signal of interest for the user. As a result, the authors typically disable the sensitivity control. Exceptions to this practice include adult recipients who have unique listening needs across diverse acoustical environments, possess a good understanding of the function of the sensitivity control, and are able to adjust it appropriately to meet their needs across challenging listening settings. Again, most users' needs are addressed satisfactorily without the need for manual sensitivity control adjustments.

Adjustments to the volume control setting on any Advanced Bionics sound processor essentially has the effect of adjusting the M-level stimulation delivered to the recipient. The volume control parameter in SoundWave determines maximum




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**FIGURE 4–18.** Adjust sensitivity setting by clicking on the up/down arrow in the window displaying the sensitivity setting. *Note.* Image provided courtesy of Advanced Bionics.

and minimum M levels that may be obtained with changes to the volume control from the default setting (i.e., the default volume control setting of the Naída CI Q70 sound processor resides at the middle of the volume control range as indicated by the AB myPilot). Additionally, internal audible tone indicators provide a cue to the patient that the default volume control setting is reached. When the Naída CI Q70 sound processor is powered on, the volume setting is at the default control setting. However, if the recipient exits standby mode, the processor resumes function at the volume control setting that was previously used. Also, when the recipient changes programs, the volume setting is maintained. Therefore, if he or she increased the volume control in program 2 and then changes to program 3, the volume will remain at the same increased setting that was selected in program 2.

The default volume setting of other Advanced Bionics processors (e.g., Neptune, Harmony, Auria) is at the "12:00 position" on the volume control wheel). For all Advanced Bionics processors, the extent to which the M level can be reduced ranges from 0 to ±100% and is related to a percentage of the dynamic range. For instance, at a setting of −50%, reducing the volume control from the default position to the minimum position decreases the M level by 50% of the electrical dynamic range. For this example, if T level is 10 charge units and M level is 100 charge units, then adjusting the volume control from the default setting to the minimum position will decrease the M level by 45 charge units (i.e., [100 − 10] * 0.5 = 45). If the minimum setting is programmed to 0%, then a reduction in the volume control below the default position will not decrease the M level at all. This may be appealing for children who have no tolerance problems (i.e., tolerates M levels during all waking hours) and for whom the clinician may be concerned that the child may inadvertently decrease the volume control setting. Finally, if the lower end of the volume control is adjusted to −100%, reduction of the volume control to the minimum setting will result in M levels corresponding to T levels. When using the sound processor at this setting, the stimulation from the processor may be very soft or inaudible. This may be appealing for new recipients who are acclimating to their implant or for children who prefer a low-level setting in the morning. In this case, adjustment from the minimum volume control setting to the preferred setting should typically take place over a few seconds.

The default values within the software for volume control are −50% to +20%. The clinician should exercise caution when setting the upper range of the volume control parameter (i.e., +50%). In the aforementioned example, an upper-range value of 100% would allow the user to increase the M-level setting from 100 charge units at the default position to 190 charge units at the maximum position (i.e., M level of 100 charge units − T level of 10 charge units = an electrical dynamic range of 90 charge units; 100% of EDR = 90 charge units; M level of 100 charge unit + 90 charge units with VC range at 100% = 190 charge units when VC is set to maximum position). If M levels are set accurately at the recommended user setting (i.e., default setting established while programming), then a 100-percent increase in stimulation relative to the electrical dynamic range (or even 50%) is rarely, if ever, appropriate. When the volume control provides too much latitude for increasing M levels, the user may encounter uncomfortable signal levels. As a result, for adults, the upper end of the volume control range should be set between 10 to 25% to allow a small increase in volume as needed. Additionally, when the range of the volume control is large (e.g., ±100%), then small changes of the volume control dial result in relatively large changes in stimulation and presumably the loudness the recipient experiences. With that in mind, the authors typically set the lower end of the volume control range to be no greater than −50%. Given the fact that M levels are set appropriately for the recipient, a smaller volume control range allows the user to be more precise in selecting alterations as needed. For children who have optionally programmed M levels (i.e., based on ESRT), the upper end of the volume control range should be set to 0 to prevent a further increase in M-level stimulation and potential discomfort. It should be noted that the authors do, however, set the lower end of the volume control range to −100% for many new recipients. This allows new recipients to reduce the stimulation significantly as needed.

In fact, for new pediatric recipients, the family is counseled to place the transmitting coil/headpiece on the child's head with the volume control set to −100% and then gradually increase

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the volume control to the desired setting over the course of a few seconds or several minutes. This practice is typically recommended for children during the first few days of cochlear implant use. The parents are encouraged to gradually increase the starting volume level used when the device is initially placed on the head in the morning or after a nap until they are certain that the device may be placed on the head at the recommended volume control setting without anxiety or discomfort.

### Channel Gains

As discussed in Chapter 3, the channel-gain parameter allows for frequency-specific adjustments to the gain of the signal from the microphone. As shown in Figure 4–19, channel gains may be accessed by selecting the input tab that resides right above the channel one marker in the programming screen. When a patient has appropriate stimulation levels, adjustment to channel gains is most likely unnecessary. However, it may be helpful for recipients who cannot tolerate changes in M level and have complaints about sound quality. A program with reduced high-frequency channel gain could address issues with sound quality (e.g., too sharp/tinny) or discomfort, whereas reductions to low-frequency channel gains may improve listening comfort and performance in noisy environments. Increases to low-frequency gain channels will likely change overall loudness, whereas increases to high-frequency gain channels may enhance sound quality.

### Input Dynamic Range (IDR)

In Advanced Bionics implants, the default IDR is 60 dB; therefore, inputs ranging from 25 to 85 dB SPL are mapped into the recipient's electrical dynamic range. Inputs below 25 dB SPL are discarded, and inputs exceeding 85 dB SPL are subjected to significant compression. Most recipients do very well with an IDR of 60 to 65 dB. Decreasing the IDR results in an increase in the "floor" of the range of sounds available to the user. For example, decreasing the IDR from 60 to 40 dB adjusts the acoustic inputs that are mapped to the lower end of the dynamic range from 25 to 45 dB (i.e., minimal influence to the upper end). At this modified setting, soft sounds will not be as

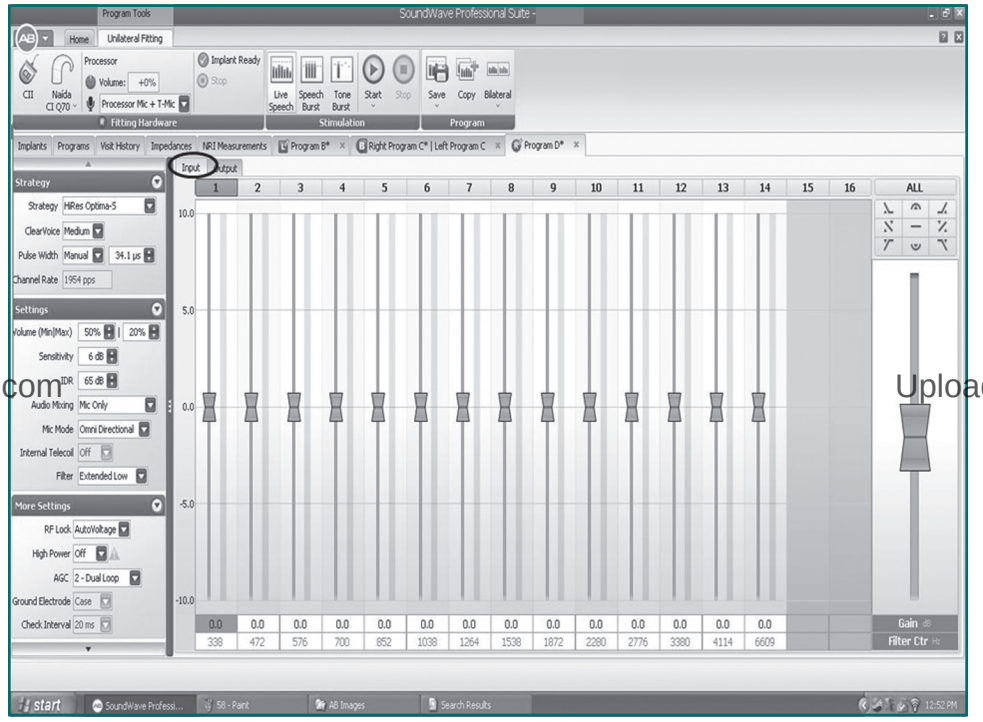



**FIGURE 4–19.** Adjust channel gains with the input tab above the channel 1 marker in the programming screen. *Note.* Image provided courtesy of Advanced Bionics.

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audible, but listening comfort may be improved in noise. Increasing the IDR also influences inputs mapped at the lower end of the dynamic range by positioning low-level sounds higher into the electrical dynamic range. In addition, the IDR increase will improve sound quality for high-level sounds and music appreciation by placing moderate-to-high level sounds higher in the recipient's electrical dynamic range; thus, reducing the chances of high-level compression (Spahr, Dorman, & Loiselle, 2007). In fact, Advanced Bionics recommends creating a program with an increased IDR to enhance sound quality for music. The IDR can be adjusted from 20 to 80 dB and can be set on a program by program basis.

### *Audio-Mixing Ratio*

The function of the audio-mixing ratio is different for the Naída CI Q70 processor as compared to all other Advanced Bionics processors. For the Naída sound processor, the audio-mixing ratio adjusts the relative strength of the signal arriving from the microphone to the signal arriving from the telecoil or from a Phonak Roger receiver connected to the base of the 170 Power Cel (i.e., the telecoil and the Roger signal are the only inputs considered as an "auxiliary source"). For all other Advanced Bionics sound processors, the audio-mixing ratio adjusts the relative strength of signals arriving from the microphone of the speech processor to the signals arriving from any other auxiliary source (e.g., FM system, iPod or other battery operated devices, or T-Mic coupled to the direct auditory input of the processors as well as the telecoil input). In other words, for all sound processors aside from the Naída CI Q70, all input sources other than the sound processor microphone are considered as "auxiliary sources."

Several audio-mixing ratio (Mic/Aux) options are available (Figure 4–20), some of which include: (1) Mic only—the microphone is the only input enabled for the processor, (2) 50/50—the input from the microphone and auxiliary source receive equal emphasis, but there is 6 dB of attenuation to each input, (3) 30/70—the input from the microphone(s) used with the sound processor is attenuated by 12 dB along with 3 dB of attenuation provided to the signal from the auxiliary source to provide a greater emphasis on the auxiliary input, (4) Aux Only—the sound processor microphone is disabled so that the only input the receives is through the auxiliary input, and (5) Aux Only (Atten.)—the sound processor microphone is disabled and the auxiliary input is attenuated by approximately 20 dB. It should be noted that the aforementioned attenuations are applied even if the auxiliary signal is not present.

The difference in the audio-mixing ratio parameter in the Naída CI Q70 sound proces-

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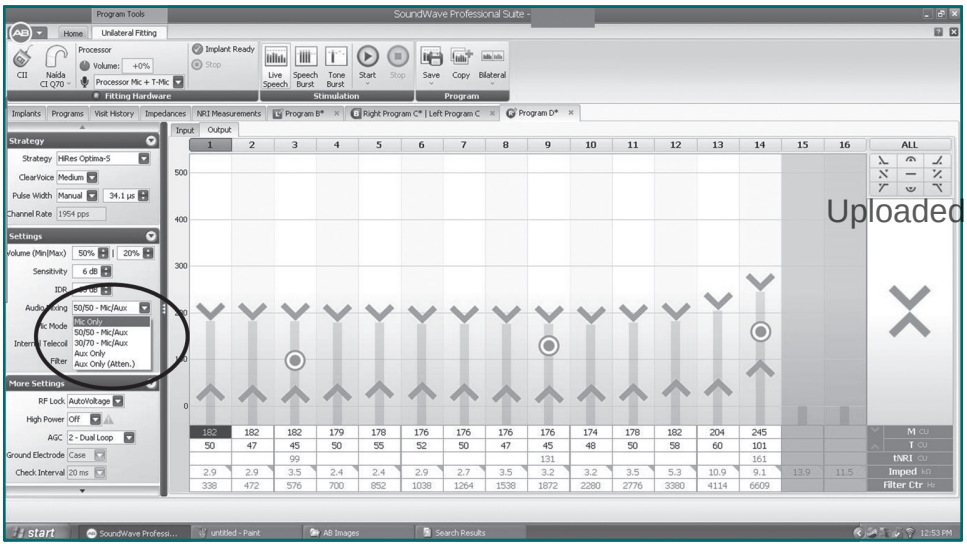
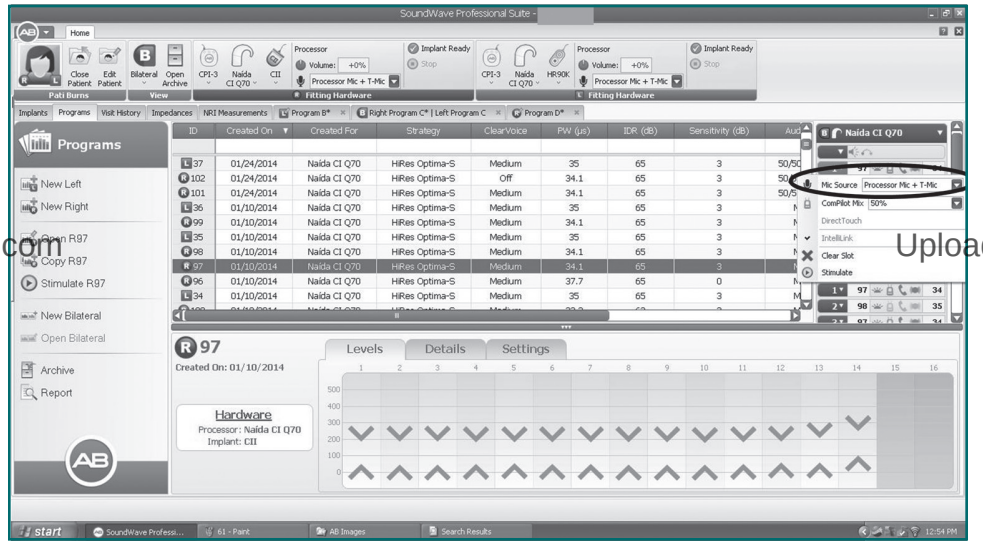



**FIGURE 4–20.** Audio-mixing ratio options. *Note.* Image provided courtesy of Advanced Bionics.

sor as compared to all other Advanced Bionics sounds processors is noteworthy. For the Naída CI Q70 sound processor, the clinician should only create a program with "Aux" input for situations in which the recipient is planning on using the telecoil or Phonak Roger 17 receiver. In this case, the Aux-Only option would allow the recipient to have a "telecoil-only" or "Roger digital RF only "program, which may be helpful when using a telephone or Roger remote microphone system in a noisy environment. Similarly, the 30/70 audio-mixing ratio may be selected for a telecoil (or Roger digital RF) program so that the recipient may understand speech on the telephone better in noisy environments (because of the 12 dB attenuation at the sound processor microphone), but at the same time, the recipient could still monitor his or her own voice (albeit at a slightly lower level because of the microphone attenuation). It is important to note that the recipient will not hear any microphone input when the "Aux Only" program is selected. As a result, this option is an inappropriate choice for young children because they would be deprived of access to sound if the sound processor was inadvertently switched to a program with "Aux Only" mixing. The default audio-mixing ratio for all processors is "50/50" option. The Naída microphone input (T-Mic 2, processor microphone, Processor Mic + T-Mic 2, headpiece microphone) is then selected from the microphone source menu (Figure 4–21). If the clinician is certain that the recipient will never use the auxiliary input, then the authors recommend the "Mic Only" mixing option because no attenuation will be provided to the Mic input(s).

It should also be noted that mixing may occur at the Naída microphone when two microphone inputs are selected from the microphone source menu. As described later in the section pertaining to loading of programs onto the Naída processor (i.e., processor configuration), the clinician may control microphone mixing separately by choosing to enable the recipient's T-Mic 2 and sound processor microphone for simultaneous use. When this option is selected, the signal from each output is attenuated by 6 dB to account for the summation that exists when the microphone signals are combined with one another. Because of this summation and the 6 dB of attenuation at each microphone, the total output of the joint microphone signals will be similar to what was obtained if either microphone was used independently.

It is also worth noting that some Naída CI Q70 users may perceive incidental sounds to be softer when using an auxiliary source, such as a Phonak Roger remote microphone personal RF system. This perception is likely attributed to the 6 dB of attenuation that occurs at the sound pro-




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**FIGURE 4–21.** Microphone mode drop-down menu. *Note.* Image provided courtesy of Advanced Bionics.

cessor microphone and the subsequent reduction in audibility for sounds arriving at the sound processor microphone rather than the Roger remote microphone. In these cases, it may be possible to enhance the salience of the signal of the sound processor microphone by increasing the sensitivity parameter. If this adjustment improves the user's listening experience, then the clinician may wish to create an alternative program for Roger use with a sensitivity setting that is 3- to 6-dB higher than the sensitivity setting used in the default program.

For the Harmony, Auria, CII, and Platinum BTE processors, microphone mixing (between the sound processor microphone and T-Mic) is controlled by adjusting the audio-mixing ratio. In other words, there is not an independent menu to adjust microphone mixing separate from mixing of the telecoil, FM, and so forth. When, the T-Mic is connected to the direct audio input post of the Harmony, CII, or Platinum BTE, a 50/50 audio mixing ratio results in equal emphasis being placed on the signals from the sound processor microphone and the T-Mic (but once again, the output of each microphone is attenuated by 6 dB to account for summation that occurs when microphones are active). The Aux Only program is commonly used with the T-Mic so that all of the sound input is captured by the T-Mic rather than the sound processor microphone. Gifford and Revit (2010) reported significantly better speech recognition in noise with a T-Mic only program relative to a program in which the sound processor microphone is enabled. This improvement is presumably due to placement of the T-Mic in the concha, allowing for preservation of the natural positive directivity of the auricle. For this reason, the authors recommend a default program of Aux Only along with the T-Mic for all Harmony, Auria, CII, and Platinum BTE users. The authors also recommend the T-Mic 2 as the only input source for the default program for Naída CI Q70 users. To achieve this, the clinician must select a "Mic Only" audio-mixing ratio and then select the T-Mic only option in the microphone source ("Mic Source") menu. The microphone mode menu is accessed by selecting the arrow that resides next to the number of each program slot (while in the programming mode) (see Figure 4–21). The clinician then uses the drop-down menu to access the "Mic Source" options and selects "T-Mic." For any Advanced Bionics processor, it is rare to use the "Mic only" mode when the omnidirectional microphone of the sound processor is the only active microphone and input source. "Mic only" is used with the Naída and UltraZoom technology for a directional program for noisy situations. Exceptions when a "Mic only" program may be used would be for a recipient who is wearing the processor off the ear (i.e., body-worn for a baby or an active adult playing sports) or for a recipient who does not or cannot use the T-Mic (i.e., the T-Mic is broken or the recipient does not like to use the T-Mic). An additional discussion of audio-mixing settings and adjustments is provided in Chapter 7.'

### *Mic Mode*

The "Mic Mode" parameter is an option for the Naída CI Q70 sound processor and determines whether the sound processor microphone will operate in omnidirectional or fully adaptive directional mode (see "Mic Mode" on the left side of Figure 4–18). The two options in the drop-down "Mic Mode" menu are "Omni Directional" and "UltraZoom." The "Omni Directional" option is the default mode and results in the implementation of an omnidirectional polar plot pattern for the microphone of the Naída sound processor. The "UltraZoom" option enables the fully adaptive directional microphones (on board front and back microphones), which enables the directional microphone with a polar plot pattern that adaptively adjusts to changes in the competing noise in order to position the null of the polar plot pattern in the rear hemisphere and toward the direction from which the most intense competing noise signal is arriving. As mentioned previously, the Naída CI Q70 possesses two omnidirectional microphones, and the electrical output from these two microphones may be analyzed to achieve the desired directional response. Research has shown that the UltraZoom preprocessing feature can provide very significant improvements in speech recognition in noise, particularly when paired with ClearVoice input processing (Hehrmann, Fredelake, Hammacher, Dyballa, & Büchner, 2012). It should be noted that in the SoundWave

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2.2.25 software, the ComPilot streaming must be enabled for programs with UltraZoom, because the UltraZoom feature is operated on the section of the digital signal processing chip that also operates ComPilot streaming. However, later versions of the SoundWave software will not require ComPilot streaming to be enabled in order to use UltraZoom.

### Internal Telecoil

The "Internal Telecoil" parameter simply enables the telecoil that is built into the Advanced Bionics ear-level sound processor (e.g., Harmony and Naída). The Neptune sound processor accesses the telecoil via the T-Com accessory. For Naída CI Q70 and Harmony processors, the "Audio Mixing" may be used to determine the relationship between the output from the sound processor microphone and the telecoil output. Selection of "Aux Only" provides a telecoil only program, while a 50/50 mixing ratio provides equal strength for the signals arriving from the sound processor microphone and telecoil. For a 50/50 mixing ratio with the Naída, the microphone input (T-Mic 2 or sound processor microphone) is determined by the clinician when loading a program (Figure 4–21). Again, it is important to note that although a 50/50 mixing ratio results in equal emphasis being placed on the inputs from the sound processor microphone and the auxiliary source (e.g., in this case, the telecoil signal), attenuation is provided to each input to account for summation that occurs when a signal is received simultaneously from two different sources. Specifically, 6 dB of attenuation is provided to the sound processor microphone relative to the "Mic Only" condition, and 6 dB of attenuation is provided to the signal from the auxiliary source (e.g., telecoil) relative to the "Aux Only" condition.

For the Harmony sound processor, once again, the audio mixing is used to control the relationship between the strength of the signals from the microphone and an auxiliary input. For the Harmony, the auxiliary input may include the telecoil or any input is received from the auxiliary post, including the T-Mic or an audio signal from the Direct Connect hook. When a T-Mic is used with a 50/50 mixing ratio for Harmony users, both the sound processor microphone and the T-Mic will be active, but the output of each will be reduced by 6 dB to account for the summation that exists a signal is provided simultaneously by two sources. The 6 dB attenuation is provided to both the sound processor microphone and the auxiliary input when a 50/50 mixing ratio is selected for use with the telecoil or Direct Connect hook. It should be noted that the T-Mic and the telecoil cannot be enabled at the same time because the Harmony considers each one as an auxiliary input.

### Extended Filtering

Extended filtering will determine the lowest frequency represented in the low channel. In the default mode, Extended Low Filtering, the high-pass cutoff of channel 1 is set to 250 Hz with a center frequency of 333 Hz. In the Standard Filtering mode, the high-pass cutoff of channel 1 is increased to 350 Hz with a center frequency of 383 Hz. The vast majority of recipients will benefit from the extended low-frequency filtering. If an adult recipient persistently complains of annoyance of low-frequency weighted noise (e.g., traffic noise), the clinician may choose to provide a program with Standard filtering. Additionally, if a recipient persistently complains of poor sound quality including a prominent echo, intrusive ambient noise, or his or her own voice being too loud, then it may be beneficial to try standard filtering.

### Lock

Advanced Bionics optimizes RF power levels to establish efficient yet reliable bidirectional telemetry (i.e., "lock") between the sound processor and the cochlear implant internal device. The Naída CI Q70 sound processor manages requisite RF power settings using AutoVoltage. For the HiResolution strategies, the compliance voltage is 8 volts; therefore, AutoVoltage manages the pulse width and current so that the voltage does not exceed 8 volts. For the Optima strategy, the compliance voltage is reduced to 4 volts to increase power efficiency. In Optima, AutoVoltage manages the pulse width and maximum current so that the voltage does not exceed 4 volts.

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The clinician may also select manual RF power management for Naída CI Q70 users. This option should only be selected in rare cases, as the AutoVoltage mode is designed to optimize battery life for Naída CI Q70 users. For instance, in the rare occurrence that consistent lock could not be obtained with the AutoVoltage setting, the clinician may select the manual mode and determine a manual power setting required to maintain lock. The manual RF power range spans from 1 to 15, and the clinician may manually vary the number to determine the lowest setting required to maintain lock during "live speech mode." When manual power mode is used, the authors typically select a manual setting that is one step higher than the minimum setting that is required to maintain lock during the programming to reduce the likelihood of intermittencies during daily use.

For recipients who use the HiRes and HiRes with Fidelity 120 signal coding strategies, the HiRes 90K system possesses a unique feature called PoEM (Power Estimator), which is designed to optimize battery life as well as monitor and adjust the radio frequency (RF) power needed to maintain a consistent signal across a variety of listening environments. The system uses high RF levels only when absolutely necessary (i.e., high-level noise environments), and it automatically accounts for changes in skin-flap thickness that may occur over time (e.g., edema resides at the skin flap during the weeks after surgery, a haircut). The PoEM also allows for use of the same program across multiple sound processors because the RF needs of each processor are determined during use, not during programming. The PoEM system allows for a continuous estimation of power needs, and this is possible because the Advanced Bionics 90K and CII implants use two separate RF frequencies, one for forward telemetry (49 M Hz) and another (10.7 M Hz) for backward telemetry. This allows for a simultaneous exchange of information between the external and internal devices.

It should be noted that when programming within the conventional HiRes and HiRes with Fidelity 120 signal coding strategies, the clinician may adjust the magnitude of the PoEM as well as the frequency in which the PoEM table is updated. The PoEM magnitude options are "low," "medium," and "high." The "high" PoEM setting is the default option and provides acceptable battery life in most instances. It is rare that the clinician would ever want to stray from the "high" setting. Unlike HiRes Optima, the clinician may also adjust the frequency in which the PoEM algorithm assesses the requisite power level needed to operate the cochlear implant. For conventional HiRes and HiRes with Fidelity 120 programs, the clinician may select a "Check Interval" range from 10 to 50 msec in 10 msec intervals. The default setting is 20 msec, and once again, it is rare that the clinician would need to deviate from the default setting. Experimentation with a different "Check Interval" setting may be attempted if a recipient reports signal intermittencies and other troubleshooting options have been exhausted (i.e., replacement of transmitting cable/coil, processor, etc.).

Similar to the AutoVoltage feature, the PoEM mode may also be disabled, and the clinician manually can adjust the RF level to a fixed setting. As before, the need for a manual RF adjustment is a rare occurrence and is necessary only when the recipient reports signal intermittencies across various environments. Manual adjustments are made in a similar manner as previously described for the HiRes Optima signal coding strategy.

For Harmony or Neptune sound processors, the clinician may enable "High Power Mode." High Power Mode is a program-specific option that increases the power output of the processor for recipients with high power requirements. The "High Power Mode" setting will decrease battery life considerably, so the clinician should only activate it as a last resort. It is disabled by default. The Naída sound processor manages power differently, and therefore, it does not utilize this parameter.

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### Automatic Gain Control (AGC)

The AGC feature allows the clinician to select between the default dual-action AGC input processing ("2 – Dual Loop") and linear ("0 – Linear") processing with peak clipping. The dual-action is used to compress a wide range of incoming sound within the IDR into the recipient's narrow electrical dynamic range. The microphone of the sound

processor has a range of approximately 85 dB SPL (from the noise floor of the microphone to the upper limit prior to saturation). The default IDR of the Advanced Bionics HiRes signal coding strategies is 60 dB with a range from approximately 25 to 85 dB SPL. The clinician may increase the IDR to 80 dB. Remember that the typical electrical dynamic range of the user (from electrical threshold to upper level of comfort) is typically no greater than 30 dB, and as a result, the processor must compress the wide range of acoustic inputs into the narrow electrical dynamic range of the recipient.

The AGC-2 algorithm controls an automatic gain control amplifier that possesses two kneepoints, which are associated with two different time constants. The slow-acting control possesses a compression threshold of 57 dB SPL with attack and release times of 325 and 1,000 msec, respectively. The fast-acting control possesses a compression threshold of 65 dB SPL with attack and release times of 0.6 and 8 msec, respectively. The slow-acting component possesses a low-compression ratio and essentially controls the amplifier gain for most everyday inputs that change slowly (e.g., the wide range of talker levels or noise levels encountered on a daily basis), whereas the fast component possesses a high compression ratio and prevents intense, transient sounds (e.g., door slamming, dog barking) from being uncomfortable.

Linear input processing provides linear gain as a function of changes in input level with peak clipping once the input level exceeds 65 dB SPL. This option will limit a recipient's access to a wide range of acoustic inputs. However, it may provide a louder signal for intense acoustic inputs, which may be appreciated by recipients who desire the input sound to be loud or as a special program for recipients who complain that intense, transient sounds are too soft. For the most part, however, the AGC-2 ("2–Dual Loop") should not be disabled because it allows recipients to perform well across a wide variety of environments (Spahr, Dorman, & Loiselle, 2007).

### Indifferent (Reference) Electrode Ground

As noted in Chapter 1, Advanced Bionics implants possess two reference electrodes, one on the case and another "ring" electrode which resides on the electrode lead. The case electrode is the default reference electrode and is used as the reference electrode for the vast majority of recipients. In rare cases, however, it may be necessary to switch to the ring reference electrode. Such instances include situations in which the case band reference electrode is determined to be faulty or when a nonauditory side effect (e.g., facial nerve stimulation) is present and cannot be resolved with other programming adjustments (i.e., increasing pulse width, channel clipping, or disabling one or more channels).

### Channel Clipping

Channel clipping prevents stimulation from ever exceeding a predefined, fixed level. This is desirable when the user frequently adjusts the volume control, which results in adjustments to M levels. Clipping allows for an increase to stimulation on all channels except for the one in which clipping is implemented (clipping may be adjusted in a channel-by-channel basis). This is beneficial if stimulation at a certain level causes nonauditory side effects, such as facial nerve stimulation. In this case, clipping would be set at the highest stimulation level that does not provoke the undesired response. Clipping may be accessed by right-clicking on an individual channel to expose a drop-down menu, in which "Clipping" may be selected (Figure 4–22). The programming clinician may then use the up/down arrows to change the charge unit level at which clipping is applied or a desired charge unit level may be entered into the field in the table residing below the programming level window (Figure 4–23).

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### HiBAN Technology

HiBAN technology is an acronym for Hearing instrument Body Area Network, a system designed to transmit audio and control data over a short distance (typically 1 meter or less). HiBAN technology is used for several purposes including: (1) the wireless delivery of an audio signal from personal electronic devices (e.g., mobile telephone, MP3 player, television) to the sound processor via a neck-worn interface known as the

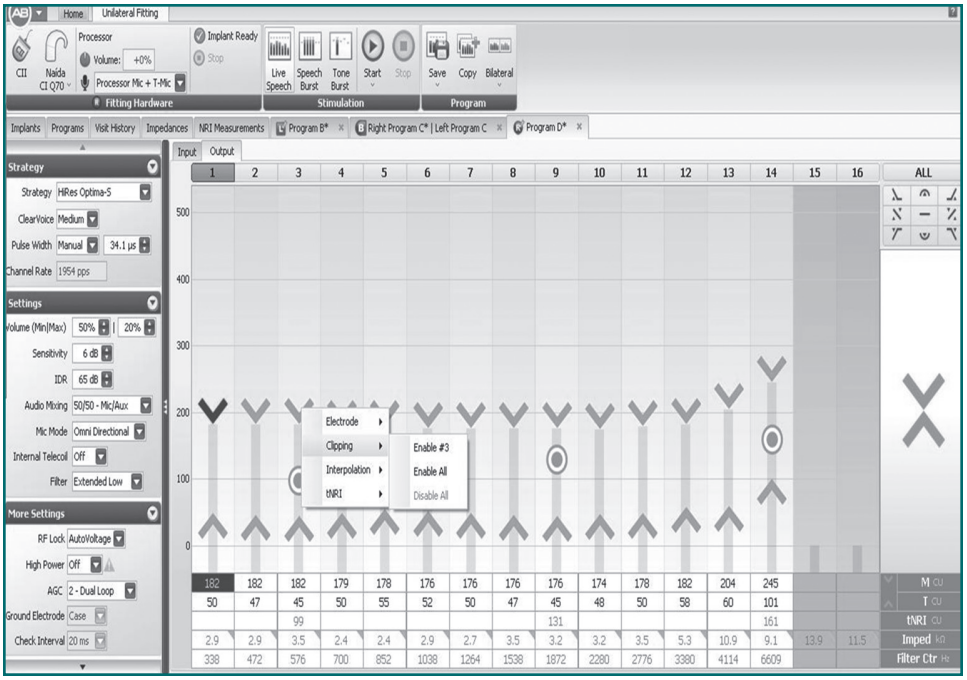
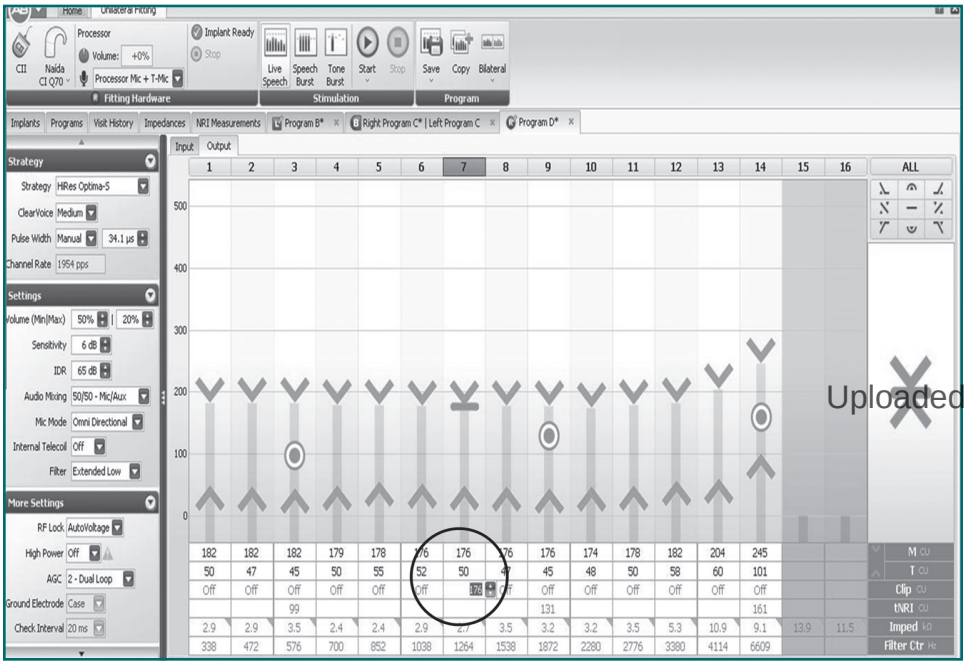



**FIGURE 4–22.** Access clipping by right-clicking on an individual channel and selecting “clipping” in the drop-down menu. *Note.* Image provided courtesy of Advanced Bionics.




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**FIGURE 4–23.** Use the up/down arrows to change the charge unit level at which clipping is applied or a desired charge unit level may be entered into the field. *Note.* Image provided courtesy of Advanced Bionics.

ComPilot, and (2) for bilateral Naída CI Q70 users, HiBAN allows for the delivery of control data from one sound processor to another, and (3) for bilateral Naída CI Q70 users, HiBAN allows for the delivery of an audio signal from one sound processor to another.

The HiBAN network allows for the wireless streaming of control and audio signals by transmitting digital data at a carrier frequency of 10.6 MHz via a digital magnetic induction link from the ComPilot neckloop to a receiving antennae located within the sound processor. The induction link and the 10.6 MHz transmission frequency substantially reduces power consumption relative to long-range RF transmission at a 2.4 GHz carrier frequency (e.g., Bluetooth). Additionally, the 10.6 MHz carrier frequency allows for audio streaming between ears without distortion or absorption by the head (i.e., auricle, noise, head). At the time of this writing, interaural audio streaming cannot be easily completed at a 2.4 GHz carrier frequency because the short wavelength of the 2.4 GHz signal is distorted and/or absorbed by the head. However, it is likely that future advances in technology will overcome this challenge.

The HiBAN system possesses a codec (i.e., digital coder and decoder) program that allows for the compression of the audio signal prior to transmission. This process removes unnecessary components (i.e., acoustic components that do not contribute to the intelligibility or sound quality of the signal) from the audio signal to reduce the size of the signal being transferred and is similar to the process used with MP3 files. The codec program allows for audio signals with a bandwidth up to 8000 Hz to be streamed between ears with minimal transmission delay (2 msec or less).

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**Binaural VoiceStream Technology.** The HiBAN system may also be used to transmit audio signals from one processor to the opposite side for bilateral cochlear implant users via VoiceStream technology. This feature allows for the provision of a number of beneficial bilateral signal processing capabilities. It is important to mention that true binaural processing is contingent upon the auditory system's ability to process short interaural time differences, and studies have indicated that bilateral cochlear implant users typically experience difficulty taking advantage binaural interaural time differences (Arnoff et al., 2010; Salloum et al., 2010). However, bilateral cochlear implantation typically results in an improvement in speech understanding and localization when compared to monaural cochlear implant use (Lammers et al., 2014; Potts & Litovsky, 2014; Schafer et al., 2011). Bilateral benefit is most likely dependent on interaural level differences that exist above 1500 Hz as well as the provision of additional information available from the redundancy of bilateral hearing when compared to monaural acquisition (i.e., bilateral hearing offers additional independent/novel channels of auditory information relative to unilateral hearing) (Salloum et al., 2010). It should be noted that at the time of this writing, Binaural VoiceStreaming Technology is only available for recipients who use Naída CI Q70 sound processors for both ears. However, plans are in place to provide Binaural VoiceStreaming Technology for bimodal users with a Naída CI Q70 sound processor and a compatible Phonak hearing aid on the opposite ear. The proceeding paragraphs highlight individual technologies available within the Binaural VoiceStream Technology portfolio.

**DuoPhone.** The DuoPhone feature utilizes the HiBAN technology to stream a telephone signal from one Naída CI Q70 sound processor to the contralateral Naída CI Q70 sound processor allowing for bilateral telephone listening. The DuoPhone feature is enabled when the telephone signal is captured by the T-Mic 2, the sound processor microphone, or the telecoil. At the time of this writing, there are no published studies describing the benefit of the DuoPhone. However, previous research with hearing aids users has indicated that use of the DuoPhone offers substantial improvement in speech understanding on the telephone compared to monaural telephone use (Stuermann, 2009; Wolfe et al., 2014). The DuoPhone feature must be activated by the programming clinician for use in one of the recipient's programs (Figure 4–24).

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**ZoomControl.** ZoomControl utilizes HiBAN technology to focus on the audio signal arriving

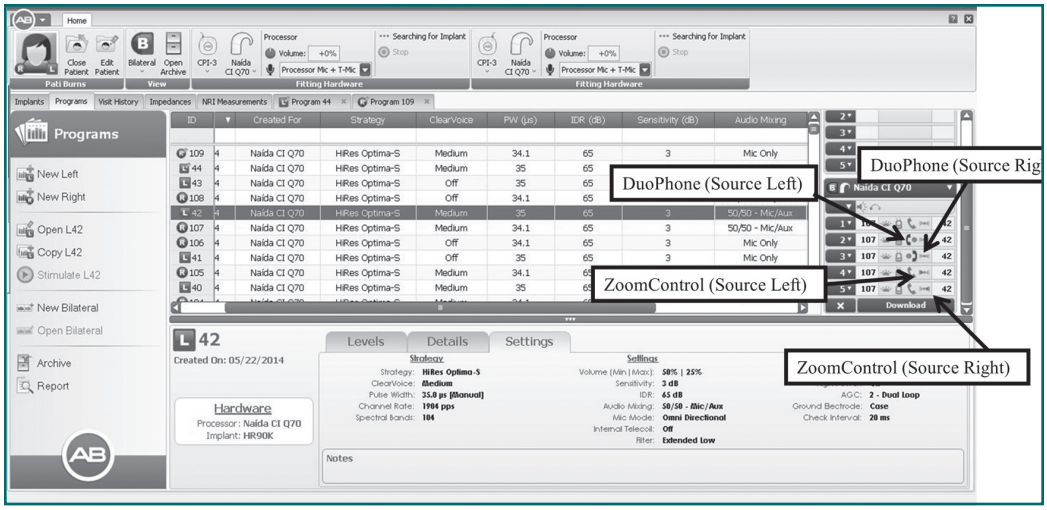



**FIGURE 4–24.** DuoPhone and ZoomControl settings. *Note.* Image provided courtesy of Advanced Bionics.

from either the right or left of the Naída CI Q70 wearer and to stream it to the contralateral ear (see Figure 4–24). For instance, if the Naída CI Q70 user was riding as a passenger in the front right seat of an automobile, ZoomControl could be used to focus on the signal arriving from directly to the left of the wearer and streaming it to the right ear to allow for both ears to take advantage of the more favorable SNR present at the left ear. Research conducted with hearing aid users has found that use of ZoomControl resulted in improved speech recognition when the signal of interest arrived from the side of the user or from behind (Stangl, Wu, Stanziola, & Bentler, 2012).

ZoomControl must be set as a special feature in a program dedicated for special use cases in which the signal of interest arrives from the side of the listener. Although the ZoomControl feature of the Phonak Quest series hearing aids allows the bilateral hearing aid wearer to focus on sounds arriving from behind him or her, the ZoomControl feature of the Naída CI Q70 only allows the recipient to focus only on sound originating directly from the left or right of the wearer. At the time of this writing, the manufacturer had plans to offer ZoomControl for signals arriving from the side of or from behind the user.

The clinician must enable ZoomControl technology in one of the recipient's programs in order to allow for use of this feature. A microphone-enabled program is loaded into a program slot of the Naída CI Q70 sound processor. Then, the clinician must select the ZoomControl symbol (see Figure 4–24), which resides adjacent to the program number loaded into each program slot. Once activated, the side of the ZoomControl icon that is highlighted indicates the side at which sound is captured to stream over to the opposite ear.

The user may access ZoomControl in one of the five programs of the Naída CI Q70 processor by using the program button on the processor or the AB myPilot to switch to a ZoomControl-enabled program. The myPilot allows the user to determine which ear is capturing the signal of interest and streaming it to the contralateral processor. ZoomControl may not be used in a program in which DuoPhone or UltraZoom has been enabled.

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**QuickSync.** QuickSync is a new feature that allows the user to change volume control or program settings for both ears simultaneously be pressing the control on just one processor. Although it is likely that QuickSync will not result in an improvement in speech understanding or hearing performance, it will facilitate a more convenient and simpler tactic to quickly adjust sound processor settings when needed.

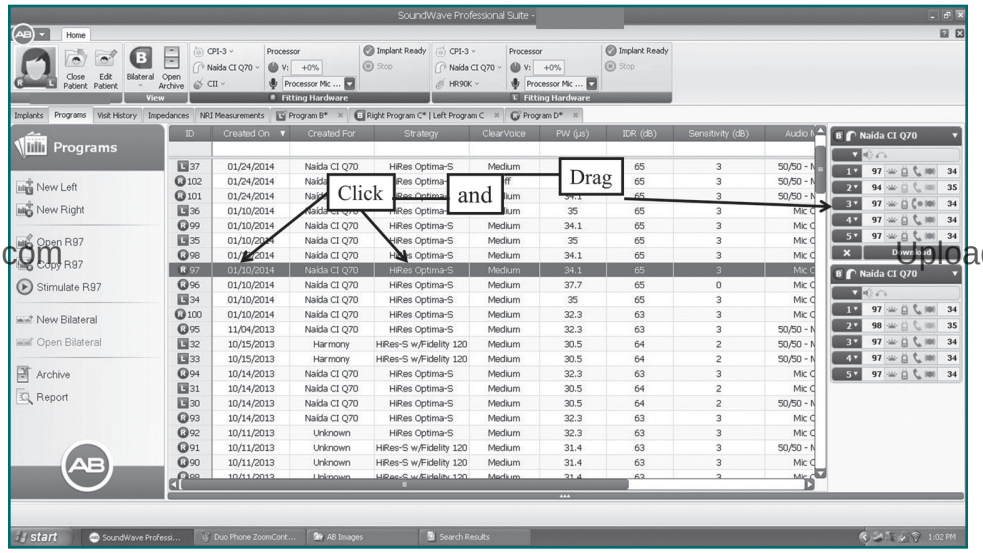
## Loading Programs into Advanced Bionics Sound Processors

To load programs into Advanced Bionics sound processors, the clinician must select the "Programs" tab. On the far right-hand side of the screen, there will be a rectangular box (the Processor Pane) with a processor icon and a textual description of the type of sound processor that is connected to the programming computer (Figure 4–25). Because different programming options exist for the various Advanced Bionics sound processors, the following section is divided into two categories, one for the Naída CI Q70 and another for all other Advanced Bionics sound processors.

When programming a unilateral Naída CI Q70 sound processor, the clinician must first select whether the processor is to be loaded for the right ear, the left ear, or as a bilateral processor. This may be achieved by "clicking" on the arrow in the upper right-hand corner of the processor box to open a drop-down menu which allows the clinician to initialize the processor for the desired ear. For the Naída CI Q70 sound processor, there will be five "slots" in which programs may be loaded. Programs are placed in each program slot by simply "clicking and dragging" the desired program from the menu of programs to the desired program slot (see Figure 4–25). When the desired number of programs is placed in the various program slots, the clinician then "clicks" on the "Download" button to load the processor with the programs.

If the processor is initialized as a bilateral processor, up to five programs for each ear may be loaded into the same processor (for a total of 10 programs). When a recipient places the sound processor transmitting coil on his or her head (i.e., achieves lock), the IntelliLink feature automatically identifies the ear that is being stimulated, and the processor selects the programs that are loaded for that ear.

For each program that is dragged to a program slot, the clinician may select multiple options that alter how the program functions for the recipient. As shown in Figure 4–26, the clinician may enable/disable the LED light on the processor by simply highlighting/deselecting the "light" icon. The clinician may also enable/disable ComPilot streaming by activating/deactivating the ComPilot streaming icon. Furthermore, the clinician may "click" on the arrow immediately to the right of each program slot number to access a drop-down menu, which provides access to numerous parameters.




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**FIGURE 4–25.** Moving programs into a program slot for the Naída CI Q70 sound processor. *Note.* Image provided courtesy of Advanced Bionics.

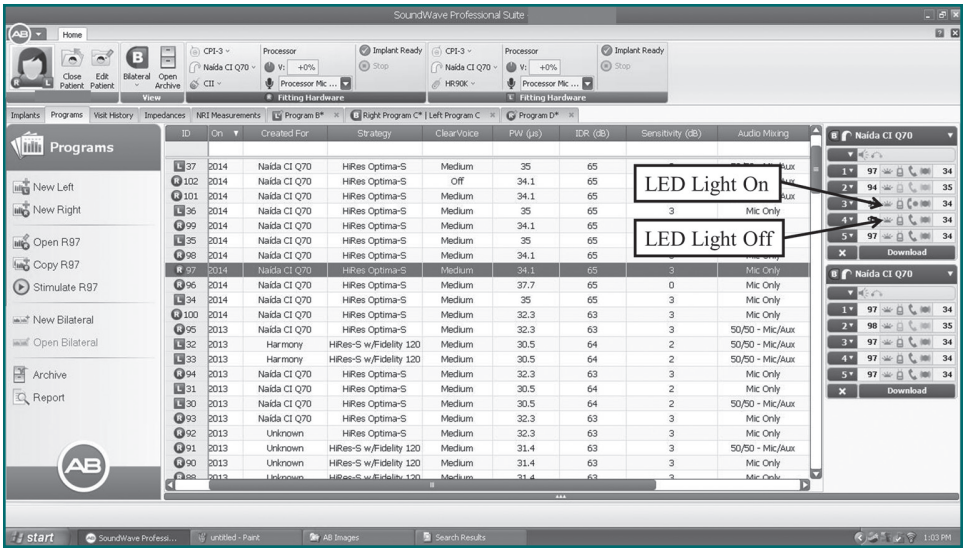



**FIGURE 4–26.** Enable or disable the LED light on the processor in the program slot. *Note.* Image provided courtesy of Advanced Bionics.

For instance, the clinician may determine the microphone source used for each program. ComPilot mixing (described later in this chapter) is also selected from this drop-down menu. Additionally, the clinician may clear a program slot, enable/disable IntelliLink, and stimulate in live speech mode from a given program slot. Also, for bilateral recipients, the clinician may confirm that the DuoPhone or ZoomControl features are enabled for a given program by noting the presence of each feature's icon (see Figure 4–24). These Binaural VoiceStream features automatically configure to the appropriate side as "transmitting" or receiving" with only one click of a button (i.e., the clinician may configure the feature by simply selecting the right or left ear as the source ear in the bilateral processor programming pane). The ease of implementing these features is a huge convenience for clinicians. Again, the ComPilot streaming icon will be highlighted for program slots in which UltraZoom is active. Finally, the clinician may "click" on the arrow to the left of the loudspeaker icon located immediately below the processor icon. This arrow opens another drop-down menu that allows the clinician to adjust the settings of the internal alarm (e.g., whether internal alarm is enabled for various functions as well as the intensity and frequency of the alarm), and the clinician may review a report of the estimated battery life (for a variety of different battery options) for each programming slot. The QuickSync feature (described earlier) may also be enabled in this menu for bilateral Naída users.

## Datalogging

At the time of this writing, Advanced Bionics has just introduced a new datalogging feature, which provides the clinician with valuable information about the recipient's usage patterns. As shown in Figure 4–27, the datalogging report reveals the number of hours that the processor is used per day. It also breaks down the usage time into hours in which lock is obtained and hours in which lock is lost (e.g., the headpiece is not on the head). It also shows the hours per day that the processor is powered down. Furthermore, the datalogging report indicates the programs the recipient uses, the types of environments in which the recipient resides, the frequency at which the recipient uses advanced features such as UltraZoom, DuoPhone, and so forth, as well as usage patterns for the ComPilot, remote control, volume control, and program button. This datalogging report is available for all Naída CI Q70 users. The clinician should

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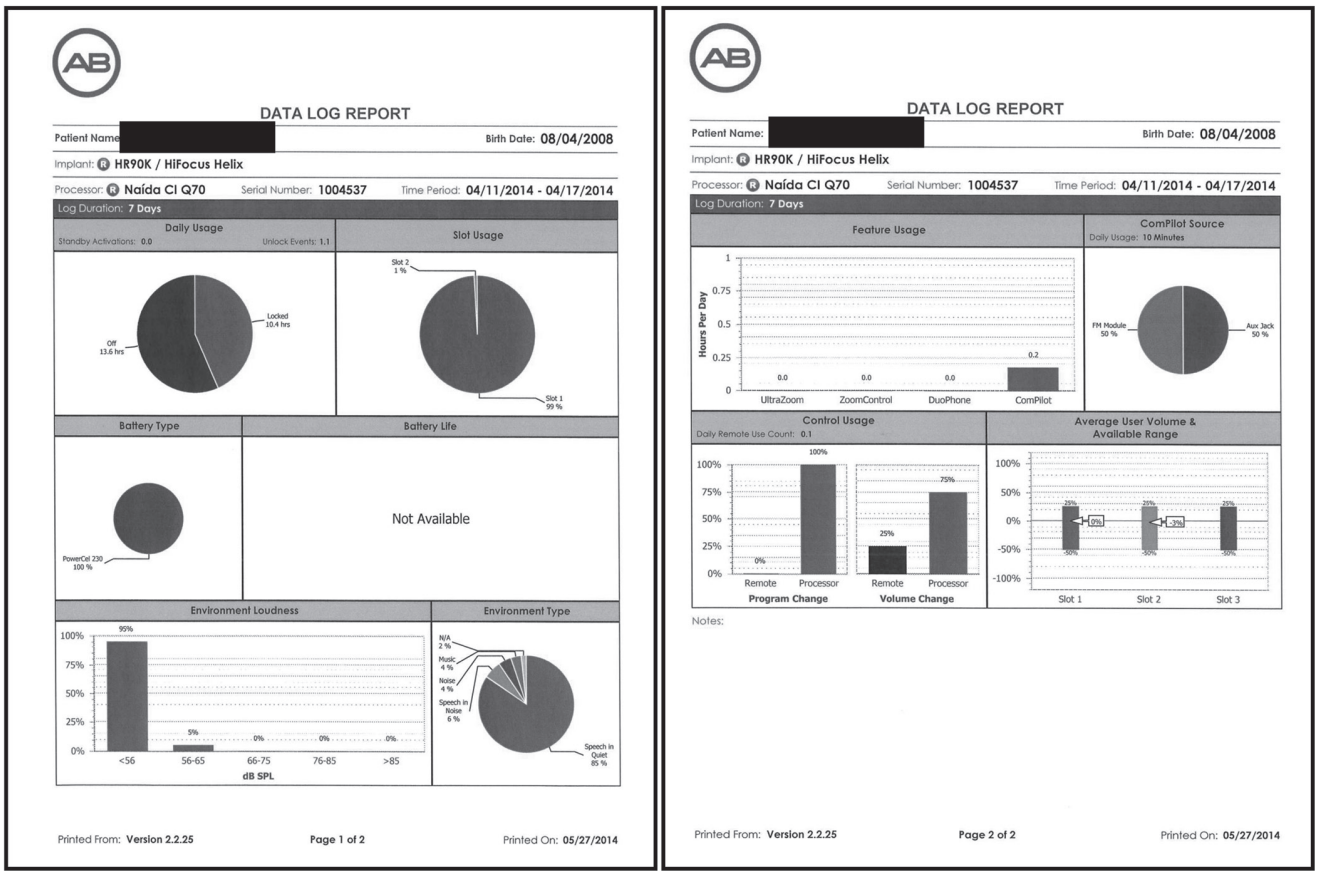



**FIGURE 4–27.** Datalogging report. *Note.* Image provided courtesy of Advanced Bionics.

review this information at each programming session in an effort to ensure the recipient (and his or her or family) is using the device appropriately (i.e., full-time use with children) and to assist in troubleshooting any recipient complaints.

## Neural Response Imaging (NRI)

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The electrically evoked compound action potential (eCAP) may be recorded in the Neural Response Imaging (NRI) platform within the SoundWave 2.2.25 software. The NRI platform in SoundWave 2.2.25 is also referred to as Flex NRI, and it possesses several upgraded features from the NRI platform in previous versions of SoundWave software. For instance, NRI measurements may be set up on multiple channels, and once the clinician initiates the measurement, the software will then automatically complete NRI measurements on each selected channel. The clinician may alter measurement parameters at any time during the process of completing the NRI assessment on each of the channels.

The NRI platform may be accessed by selecting the "NRI Measurements" tab in the SoundWave 2.2.25 software. The NRI measures are completed by delivering biphasic, rectangular electrical pulses at a rate of 30 pulses per second on the channel selected for stimulation and recording the eCAP captured at an electrode located two channels apical from the stimulated channel. When stimulating channels one and two, the recording electrode is located two channels basal to the stimulated channel. The clinician selects the minimum and maximum stimulation levels that will be used for the NRI measurement (Figure 4–28). The clinician may also select the number of measurements made between the measures made at the minimum and maximum values by enter-

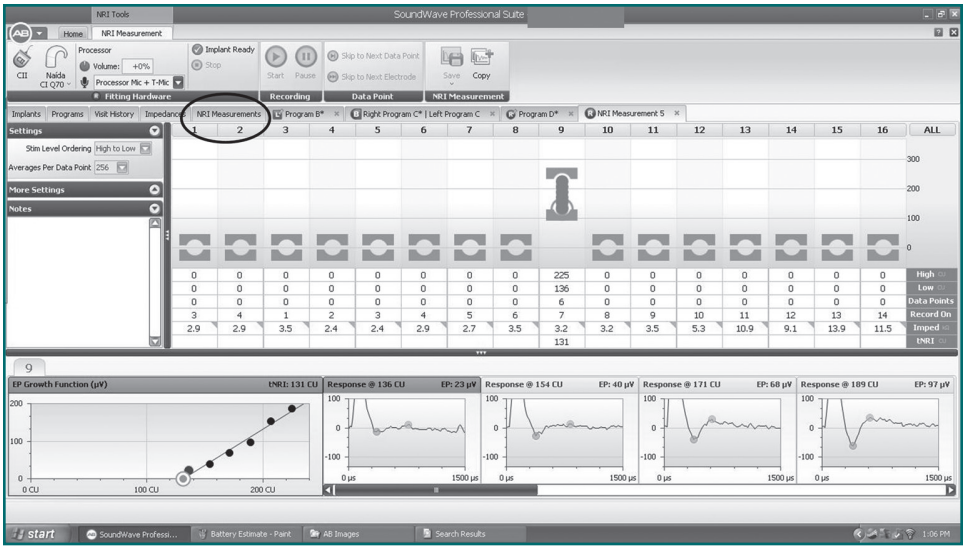



**FIGURE 4–28.** Selecting the minimum and maximum stimulation levels that will be used for the NRI measurement. *Note.* Image provided courtesy of Advanced Bionics.

ing the desired number of measures in the "Data Points" field in the table that resides underneath the NRI measurement window. The default number of measurement points is five, but the authors recommend that the distance between successive measures be no more than 25 charge units.

NRI measures can be obtained with stimulation starting at the maximum value and progressing toward the minimum value or vice versa. The authors recommend beginning the NRI at a stimulation level that should be loud but not uncomfortable for the recipient and then decreasing in 15 to 25 charge unit steps. The descending stimulation approach allows for attainment of a clear NRI response at a relatively high stimulation level and termination of the NRI measure on a given channel as soon as the clinician observes that the response is no longer present. The number of sweeps (e.g., stimulus presentations) on each channel is defined by the "Averages per Data Point" parameter. The default value is 32 sweeps, which is often sufficient to record a response. However, the authors typically select 128 sweeps and then choose to manually advance the measure to the next stimulus level once a clear response or no response has unequivocally been obtained. The recording gain of the amplifier is set to a default of 300 (i.e., amplified by a factor of 300). The amplifier gain can be set to 1,000 times of amplification, but this adjustment is rarely needed. Finally, the NRI platform alternates the polarity of the biphasic stimulus to reduce stimulus-related artifact. The clinician may select whether the stimulus leads with a cathodic (default) or anodic phase, but again, this adjustment is rarely needed.

It should be noted that NRI measurements may also be completed within the programming platform. The clinician must simply "right click" on a channel to access a menu that includes "tNRI." Then, the software automatically attempts to measure a tNRI response, which is an estimated NRI threshold based on where best fit a regression line (calculated from the amplitude of NRI responses made at the various stimulation levels) crosses 0 Charge Units on the *x*-axis (see Figure 4–28). The stimulation levels the software selects for tNRI assessment are based on the T levels and M levels determined for a given channel. The tNRI responses obtained from either the NRI Measurement platform or the NRI measurement obtained within the programming screen are imported to a recipient's program. The reader is referred to Chapter 7 for additional discussion on eCAP measurement as well as to Michelle Hughes' excellent book, *Objective Measures in Cochlear Implants* (Plural Publishing CCC Series on Cochlear Implants).

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## Hearing Assistance Technologies

### *Phonak ComPilot*

The Phonak ComPilot available for use with the Advanced Bionics Naída CI is a hearing assistance technology device that serves three general purposes: (1) streaming of audio signals from the ComPilot neck-worn wireless interface to the Naída CI Q70 sound processor, (2) adjustment of settings on the Naída CI Q70 including volume and program adjustments, and (3) bimodal streaming to the Naída CI Q70 and any Phonak-compatible hearing instrument.

The ComPilot streams audio signals by transmitting digital data at a carrier frequency of 10.6 MHz via a wireless magnetic induction link from the ComPilot neckloop to a receiving antenna located within the sound processor. Audio signals may be delivered wirelessly to the ComPilot via Bluetooth transmission. Specific use-case examples include pairing with personal electronics devices, such as a mobile telephone or a Bluetooth-enabled laptop computer. The ComPilot also possesses a microphone that captures the user's voice and transmits it to the mobile telephone so that it may be heard by the conversational partner with whom the ComPilot user is speaking. In this instance, the ComPilot is providing bidirectional streaming of the audio signal to provide "hands-free" mobile telephone use.

As described in Chapter 1 (see Table 1–4), pairing the ComPilot to Bluetooth-enabled personal electronic devices is quite simple. The user must navigate to the Bluetooth pairing menu on his or her personal audio device and place it in "search mode." Then, the user simply holds down simultaneously on the "Connect" button (<>)on the right side of the ComPilot along with the "+" button for approximately 2 seconds. This should cause the audio indicator light on the ComPilot to begin flashing blue rapidly. Within a few moments, "ComPilot" should appear on the personal audio device's menu of available Bluetooth devices to which it may couple. To complete the pairing process, the user should select the ComPilot as the desired pairing device and, if required, enter passcode "0000." The ComPilot may also be coupled to a separate base station known as the TVLink S. The TVLink S may serve as the ComPilot charger, and it also may be connected to the audio output ports of a television to provide wireless Bluetooth streaming of the television audio signal to the ComPilot interface. Once again, it is relatively simple to pair the user's ComPilot to the TVLink S. The user should simultaneously hold down on the (<>) and (+) buttons of the ComPilot for 2 seconds until the audio indicator light begins flashing blue rapidly. The user should then simultaneously hold down the power and (+) buttons of the TVLink S for 5 seconds until the indicator light flashes blue rapidly. Completion of pairing is indicated when both indicator lights turn solid blue.

The ComPilot may also be paired to the Phonak RemoteMic to allow the user to access remote microphone technology. The Phonak RemoteMic is designed to be clipped onto the shirt ideally about 6 to 8 inches from the mouth of the ComPilot's conversational partner in order to capture the talker's speech and wirelessly send it via Bluetooth to the ComPilot. The RemoteMic has a transmission range of 20 meters (60 feet). To pair the ComPilot to the Phonak RemoteMic, the user must simultaneously hold down on the (<>) and (+) buttons of the ComPilot for 2 seconds until the audio indicator light begins flashing blue. If the RemoteMic has never been paired it will automatically be in pairing mode when powered on, which is indicated by the status light flashing blue rapidly. To enter pairing mode manually, the user should press and hold the volume-up button and simultaneously power on the RemoteMic. Release the volume increase button when the status light starts flashing blue rapidly. Completion of pairing is indicated when both status lights turn solid blue.

Finally, the ComPilot features two additional modes in which the user may connect to external devices. First, an auxiliary audio cable (3.5 mm phone plug) may be used to physically connect from the audio output port (e.g., headphone jack) of a personal audio device to the audio input port at the base of the ComPilot. Additionally, a personal FM/RF receiver may be plugged into the Europlug at the base of the ComPilot. The ComPilot possesses multiple buttons that allow users to access several functions available from the device (see Table 1–4 in Chapter 1 for details).

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Within the Advanced Bionics SoundWave 2.2.25 software platform, the clinician may determine whether the ComPilot streaming is available (Figure 4–29) as well as the strength of the ComPilot signal the user receives relative to the signal from the sound processor microphone. The ComPilot mixing ratio, which governs the strength of the ComPilot signal relative to the sound processor microphone signal, may be set at 25, 50, 75, and 100% ComPilot (see Figure 4–29). A 50% ComPilot mixing ratio offers some attenuation to the signal from the sound processor microphone to preserve signal integrity and prevent loudness discomfort from the summation of the signals from the processor microphone and ComPilot. With the 50% ComPilot mixing ratio, the signal from the processor microphone and ComPilot are both attenuated by 6 dB. A ComPilot ratio of 25% actually places about 12 dB of attenuation on the ComPilot signal and 3 dB on the sound processor microphone signal to allow the user to focus more on the signal from the sound processor microphone. The 75% ComPilot mixing ratio attenuates the processor microphone by 12 dB and and the ComPilot by 3 dB to provide greater salience for the ComPilot signal. The 100% ComPilot ratio essentially mutes the microphone of the sound processor, so that the recipient only hears what is arriving from the ComPilot. Of course, in the microphone only mode with the ComPilot disabled, there is no attenuation to the signal from the processor microphone.

Finally, if the clinician pairs the ComPilot to the Naída CI Q70 sound processor(s), it may be used as a remote control to adjust the settings of the recipient's sound processor. Pairing is accomplished by connecting the ComPilot to the programming computer (by way of a USB to mini-USB cable), while the sound processors are being programmed (i.e., the recipient file is open). The volume of the sound processor may be adjusted by using the "+" and "–" buttons of the ComPilot. The processor program may also be adjusted by pushing the main (large center) button or by a short press of the Connect "<>" button ("Home Orientation" button) on the side of the ComPilot. A long press of the Connect"<>" button returns the Naída processor to program #1. When a ComPilot is paired to a recipient's sound processor(s), the "+" and "–" buttons adjust the volume of the sound processor, but they cannot be used to adjust the volume of the signal being streamed from the ComPilot.

If the ComPilot is not paired to a sound processor, then it may be used to stream audio signals to the Naída CI Q70 sound processor and also to a compatible Phonak hearing aid equipped with

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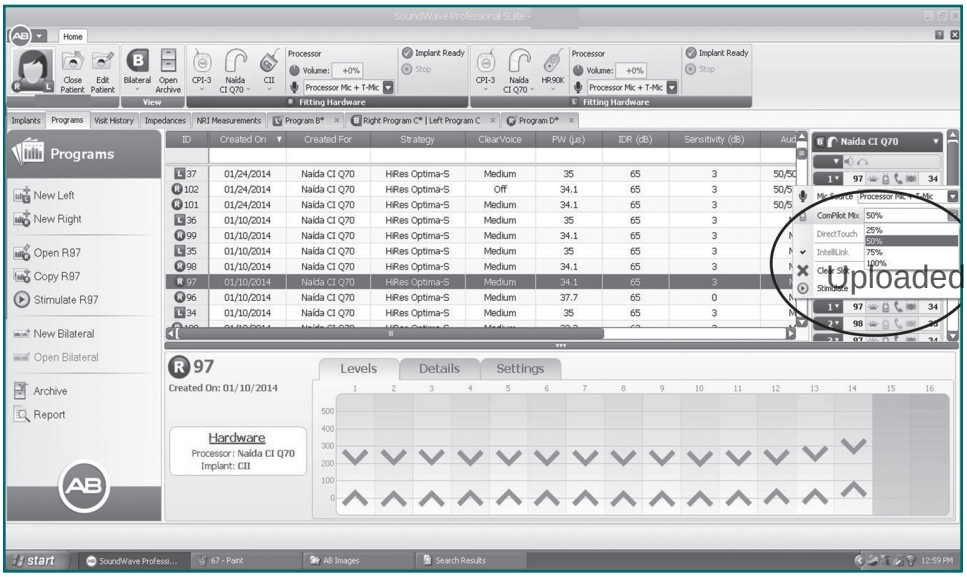



**FIGURE 4–29.** Accessing the ComPilot mixing ratio. *Note.* Image provided courtesy of Advanced Bionics.

HiBAN technology. The unpaired configuration may be desirable for bimodal users to allow for streaming to both hearing technology instruments. In the unpaired mode, the ComPilot cannot be used as a remote control to adjust the settings of the Naída CI Q70 sound processor. Adjusting the "+" and "–" buttons changes the volume of the signal streamed from the ComPilot to the sound processor. The clinician may load firmware upgrades to the ComPilot and change the language used for notifications by the ComPilot but remain in the unpaired mode by simply connecting the ComPilot to the programming computer with the SoundWave 2.2.25 software open but the recipient file closed.

### *AB myPilot*

The AB myPilot is a remote control that may be used to control the settings of the Naída CI Q70 processor(s) and also to assess the function of the Naída CI Q70 sound processor(s). The first step in successfully using the AB myPilot is to pair the myPilot to the Naída CI Q70 sound processor(s). The clinician can simply pair the myPilot to the recipient's processor within SoundWave or the processors can be paired "dynamically." Dynamic pairing is accomplished by selecting the center button of the myPilot to access the control menu and then selecting the "Pairing" function (see Chapter 1 and Table 1–4 for more details). The processors enter pairing mode once the battery is placed on powered the device, and the processor pairing window lasts 5 minutes. The myPilot will then be paired to the processor(s) if the processor(s) is within about 30 cm or less of the myPilot.

In the "standard mode," the myPilot may be used to adjust the volume/sensitivity of the sound processor (with the up and down buttons) and also to change the program of the sound processor (with the left and right buttons). The menu button (center button) allows the recipient to access a special menu that allows the user to make unilateral or bilateral adjustments, volume or sensitivity adjustments, and a special diagnostic utility that provides information regarding the status of the sound processor function and battery life. The special menu also allows the recipient to unpair/pair the myPilot to a sound processor and also change the settings of the myPilot (language, screen brightness/contrast, etc.). The diagnostic mode is referred to as "Device Status" and may be used to evaluate processor function for one processor at a time. The recipient should hold the myPilot at a closer distance to the processor he or she wishes to evaluate. It should be noted that the myPilot allows the recipient to toggle across as many as five programs.

In the direct mode, the recipient may adjust the volume control with the up and down buttons. The recipient may directly access up to five programs by pushing the left for program 1, middle for program 2, and right for program 3 buttons. A 2-second hold on the left button accesses program 4, and a 2-second hold on the right button accesses program 5. The direct mode does not provide as much flexibility for adjustment as the standard mode, but it does allow for simple and direct access to volume control and program adjustments.

## Programming Previous Generations of Advanced Bionics Implants

All recipients who have C1 and 1.2 internal devices and sound processors older than the Harmony BTE are programmed in the SCLIN programming software. Many of the basic programming strategies used in SoundWave also are present in the SCLIN software. Therefore, clinicians who are familiar with SoundWave should easily adapt to SCLIN. There are, however, a few significant differences. First, SCLIN allows C1 and 1.2 devices to be programmed with the Simultaneous Analog Stimulation (SAS) signal coding strategy (see Chapter 2 for a description). An analog stimulus must be used to measure stimulation levels with the SAS strategy. The clinician will select either an analog stimulus or a train of biphasic pulses prior to making stimulation level measurements. Second, the clinician may select from three different types of AGC: AGC1, AGC2, or AGC off, and AGC2 is recommended for the vast majority of recipients. Third, at the end of the programming session, the clinician must use a special RF Power test to estimate the amount of RF power required to sufficiently operate the user's program with the sound processor. The test presents high-level environmental sounds (e.g., clapping) whereas

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the software estimates the optimal RF power necessary to provide efficient battery life and a consistent signal. Finally, the measurement of the electrically evoked compound action potential of the auditory nerve is not available for C1 devices, but is available via the Neural Response Imaging (NRI) platform in the SoundWave software.

## KEY CONCEPTS

- The provision of appropriate M levels is one of the most important requirements for creating an optimal program for Advanced Bionics recipients.
- Many recipients perform well with T levels that are estimated from M level measurements, but some recipients have better access to low-level sounds when T levels are measured.
- Use of the T-Mic 2 as the Naída CI Q70 processor's only microphone source generally optimizes performance across a wide variety of listening situations.
- Use of advanced signal processing features, such as ClearVoice, UltraZoom, Zoom Control, and DuoPhone, are likely to improve recipient performance in difficult listening situations.
- Use of the Advanced Bionics accessories, such as the ComPilot, the Phonak Remote Microphone, and the TVLink will likely optimize user performance in several challenging situations.

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# 5

# Programming Cochlear Devices

## PROGRAMMING THE NUCLEUS 6 SYSTEM

The Nucleus 6 sound processor, the most recent release from Cochlear, possesses several new features relative to its immediate predecessor, the Nucleus 5 sound processor. Features include an option to provide electroacoustic stimulation, scene analysis with automatic adaptive directionality (SCAN), digital noise reduction (SNR-NR), wind noise reduction, datalogging, and compatibility with a portfolio of proprietary wireless hearing assistance technologies (HAT). The various components of the Nucleus 6 cochlear implant system are described in Chapter 1.

The hardware used to connect the processor to the computer (i.e., the programming pod and cable) is shown in Figure 5–1. Briefly, a programming interface, known as a pod, is connected to the

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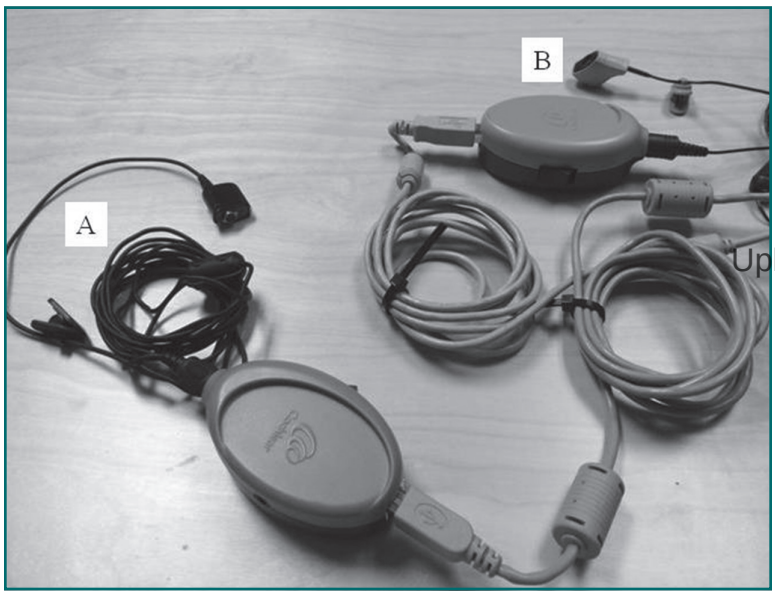



**FIGURE 5–1.** Cochlear programming Pod with cables for the Nucleus 5 and 6 (**A**) and Freedom sound processors (**B**).

sound processor with a specialized programming cable, and the pod is connected to the programming computer with a USB cable. The same pod and programming cable (Figure 5–1A) are used to connect the Nucleus 6 and Nucleus 5 sound processors to the programming computer, whereas the same pod but a different programming cable (Figure 5–1B) is used to connect the Freedom sound processor to the programming computer.

Prior to creating a program, the clinician must create a file for the recipient and identify the implant(s) the recipient has received. In the Custom Sound 4.1 software, this is accomplished by selecting the "Create" function on the right-hand side of the patient selection "start-up" page (Figure 5–2). The recipient's first and last names (Figure 5–3), the type of implant the recipient has received, and the ear that has been implanted must

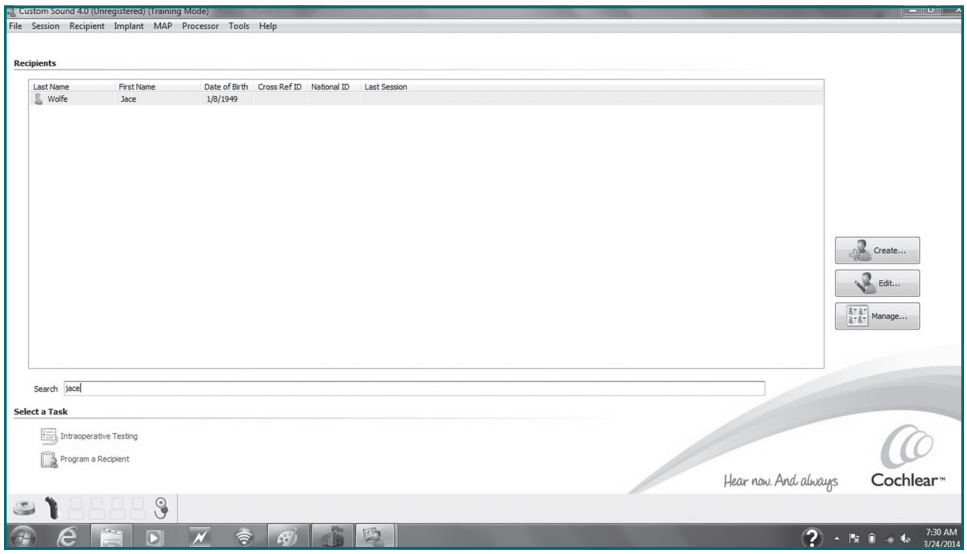
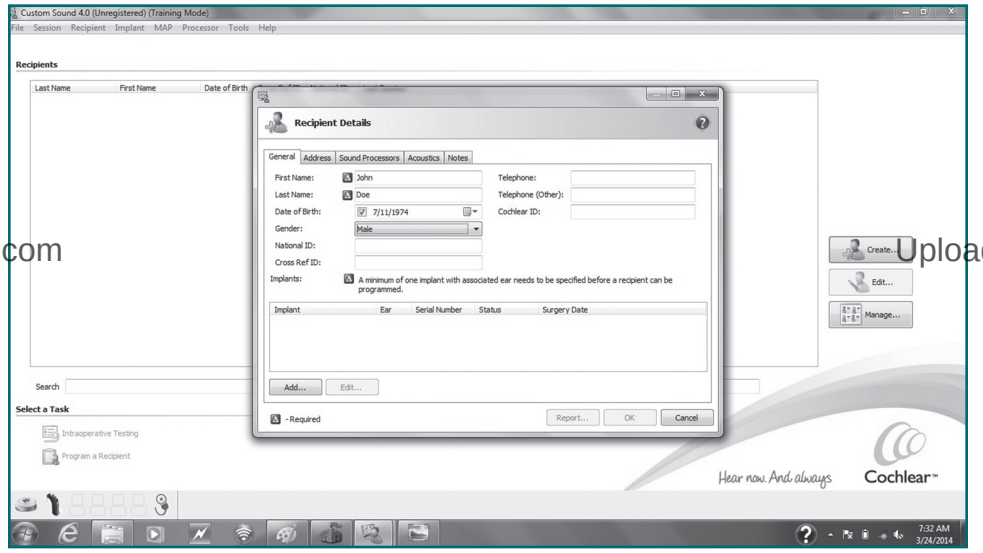



**FIGURE 5–2.** Create a file for the recipient by selecting the "Create" function. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 5–3.** In the "Recipient Details" menu, enter the recipient's first and last names as well as other demographic information; select the "Add" button when complete. *Note.* Provided courtesy of Cochlear™ Ltd.

be entered into the "Recipient Details" menu. Additional demographic information may also be entered as the clinician desires. The clinician must select the "Add" button in the "Recipient Details" menu to enter the implant the recipient has received (see Figure 5–3). A pop-up menu is provided, and from this menu, the clinician must enter the recipient implant model(s) and the ear(s) that was implanted (Figure 5–4).

In the Cochlear Custom Sound 4.1 software, the basic programming process is essentially divided into four categories (Figure 5–5): (1) Measure Impedances, (2) Open or Create MAP, (3) Set Levels, and (4) Write to Processor. The clinician may choose to incorporate other steps into the process as needed. These additional steps include: Perform Neural Response Telemetry (NRT), Bilateral Balance, and Finalize Programming. The

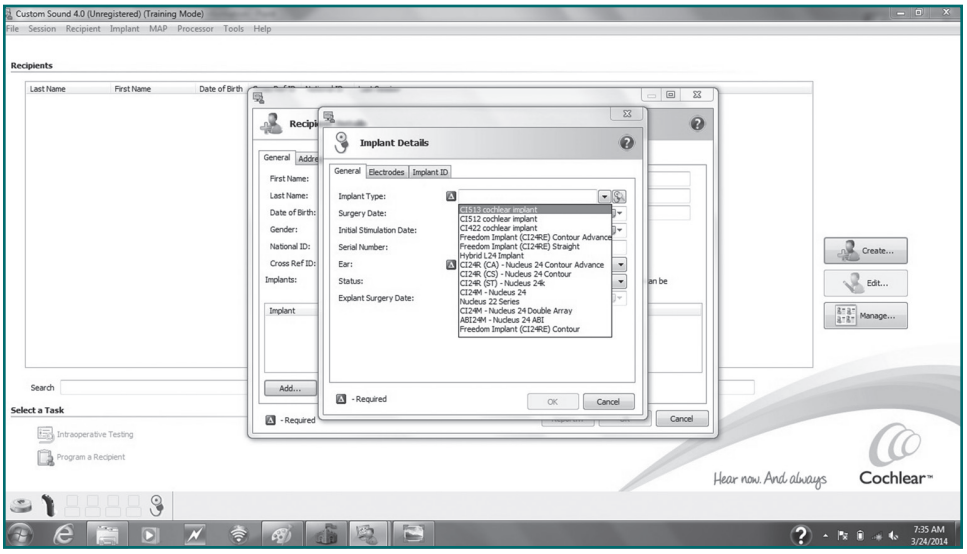



**FIGURE 5–4.** Pop-up menu for selection of the recipient implant model(s) and the ear(s) that was implanted. *Note.* Provided courtesy of Cochlear™ Ltd.

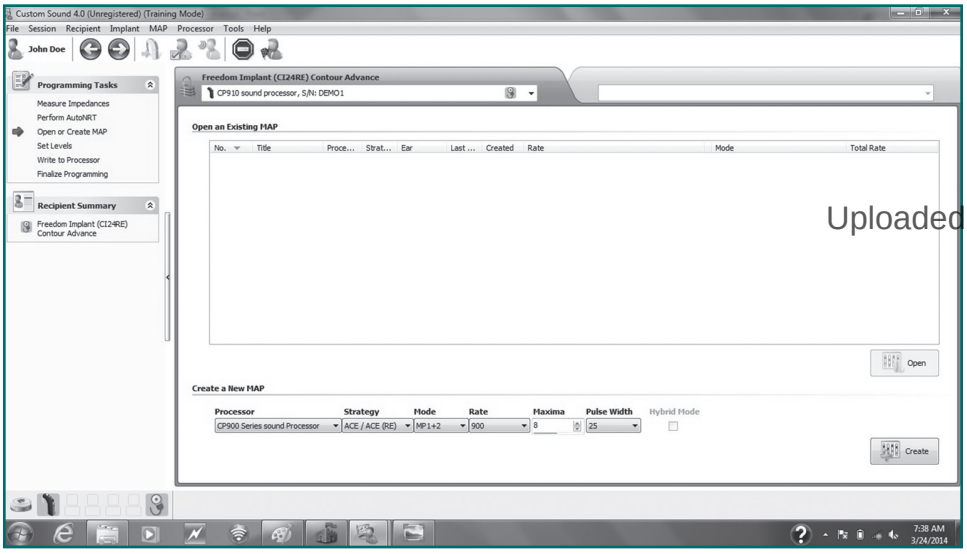



**FIGURE 5–5.** Each programming task is shown in a box on the left side of the screen. *Note.* Provided courtesy of Cochlear™ Ltd.

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following description of programming Cochlear devices will be organized to correspond with the aforementioned categories.

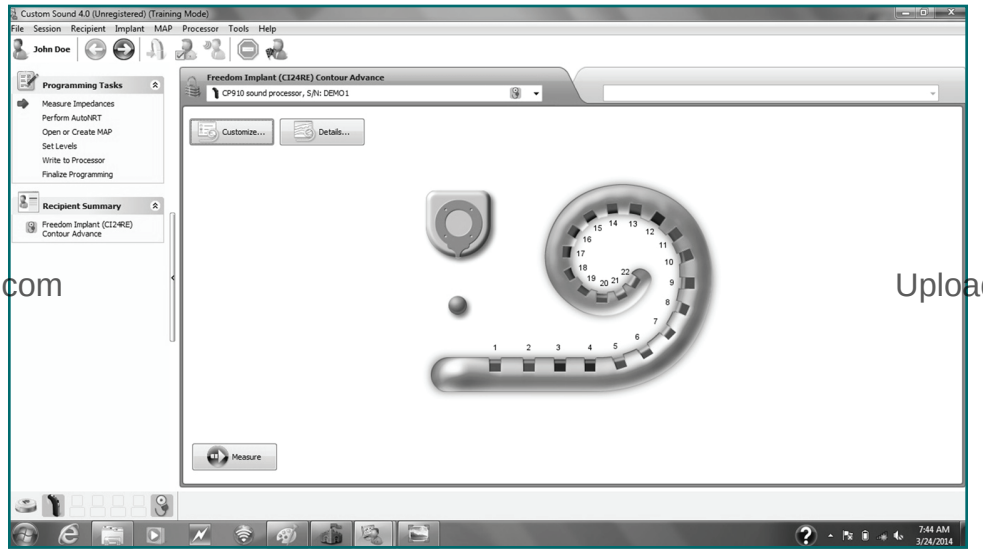
## Measure Impedances

The first step involved in programming in Custom Sound 4.1 is to measure electrode impedance in four different electrode coupling modes. Electrode impedances are measured by simply selecting the "Measure" button within the "Measure Impedance" platform (Figure 5–6). The results of the measurement are promptly displayed in the illustration of the electrode array and reference electrodes by depicting impedances falling within the normal range in a green color and any abnormal findings in a red color (see Figure 5–6). The clinician should select the "Details" button to review to electrode impedances in graphical and tabular form (Figure 5–7).

In each electrode coupling mode, a low-level electrical current (100 CL with a pulse width of 25 usec), which is inaudible for most but not all recipients, is delivered sequentially to each intracochlear electrode contact. The impedance (in kOhms) is measured across the entire circuit as the current travels from the current source to the intracochlear electrode contact and finally to one or more reference electrodes.

Impedance is first measured in the common ground mode where the low-level current is delivered sequentially to each intracochlear electrode contact. Each of the remaining intracochlear contacts simultaneously serves as the return path. Because the common ground mode electrically couples each intracochlear electrode contact to the remaining intracochlear electrode contacts within the array, it is the most sensitive mode for detecting shorted electrode contacts. As a result, common ground is the preferred electrode coupling mode to detect anomalous intracochlear electrodes.

The Monopolar 1 and Monopolar 2 coupling modes measure electrode impedance by sequentially delivering the low-level current to each intracochlear electrode contact and by evaluating the impedance in the circuit from the current source to each intracochlear electrode and extracochlear reference electrode. Measuring electrode impedance in the Monopolar 1 and Monopolar 2 modes allows the clinician to identify the status of the extracochlear reference electrodes (located on the end of the nonstimulating electrode lead and the implant case, respectively).




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**FIGURE 5–6.** Measuring impedances. *Note.* Electrodes 3, 4, and 12 are darkly shaded in this figure (abnormal impedance), whereas remaining electrodes are lightly shaded in this figure (normal impedance). Provided courtesy of Cochlear™ Ltd.

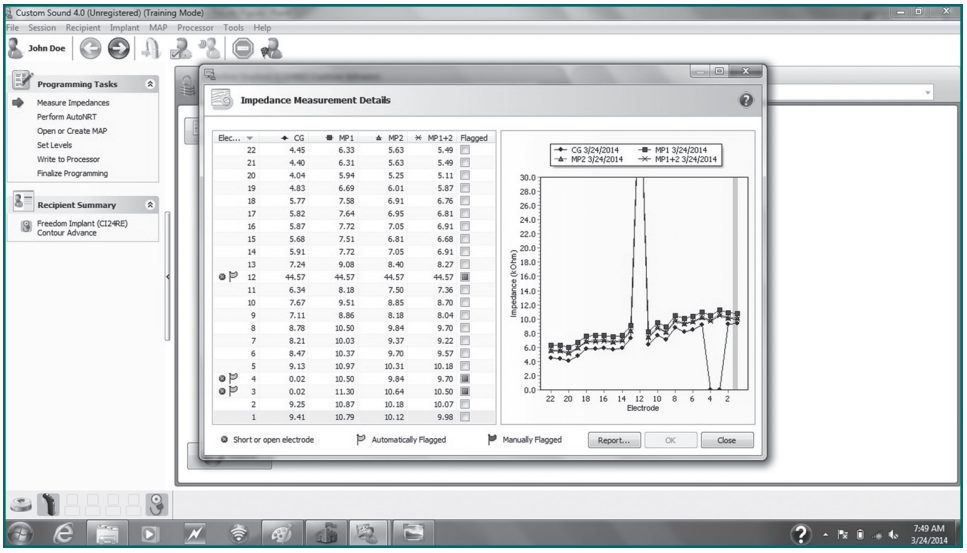



**FIGURE 5–7.** After measuring impedances, select the "Details" button to review to electrode impedances in graphical and tabular form. *Note.* Provided courtesy of Cochlear™ Ltd.

In the fourth mode, Monopolar 1+2, each intracochlear electrode is referenced to both the remote and case reference electrodes. In other words, the measure is completed by evaluating the impedance that exists in the circuit from the current source to each intracochlear electrode contact and finally to the reference, which is comprised of MP1 and MP2 which are electrically coupled to one another. The Monopolar 1+2 (MP1+2) mode is the default mode used for stimulation in the primary signal coding strategy used with Cochlear devices, the Advanced Combination Encoder (ACE) signal coding strategy (also in the Continuous Interleaved Sampling [CIS] and Spectral-Peak [SPEAK] strategies). As a result, impedance measured in the MP1+2 mode closely reflects the typical impedance present during stimulation when the implant is used on a daily basis.

According to the manufacturer, electrode impedances below 565 ohms are abnormally low and are designated as "short" circuits, whereas electrode impedances greater than 30 kohms for half-band electrode contacts (e.g., Nucleus Freedom, CI422, CI512, Hybrid) are abnormally high and are referred to as "open" circuits. It should be noted that the open circuit limit for the electrode array with full-band electrode contacts (e.g., Nucleus Straight Array and the Nucleus Double Array) is 20 kOhms. Electrodes with abnormal impedances are typically "flagged" and deactivated for programming and subsequent impedance assessments. In the "Measure Impedance" module, "flagged" electrodes are depicted in yellow for all subsequent impedance measurements. Flagged electrodes are also "grayed out" (disabled) in the programming/"Set Levels" module of Custom Sound.

At initial activation, impedance is frequently high but will generally decrease with routine implant use. Therefore, the clinician should reassess the impedance of electrodes with abnormally high electrode impedance to determine whether impedance decreases to normal levels after stimulation of the electrodes. However, shorted electrodes will typically always remain as shorted electrodes and should be permanently disabled once they are identified. It should be reiterated that shorted electrodes are identified not only by their abnormally low impedance (i.e., less than 565 ohms) but also because the common ground mode allows for detection of two or more intracochlear electrodes that are electrically connected to one another.

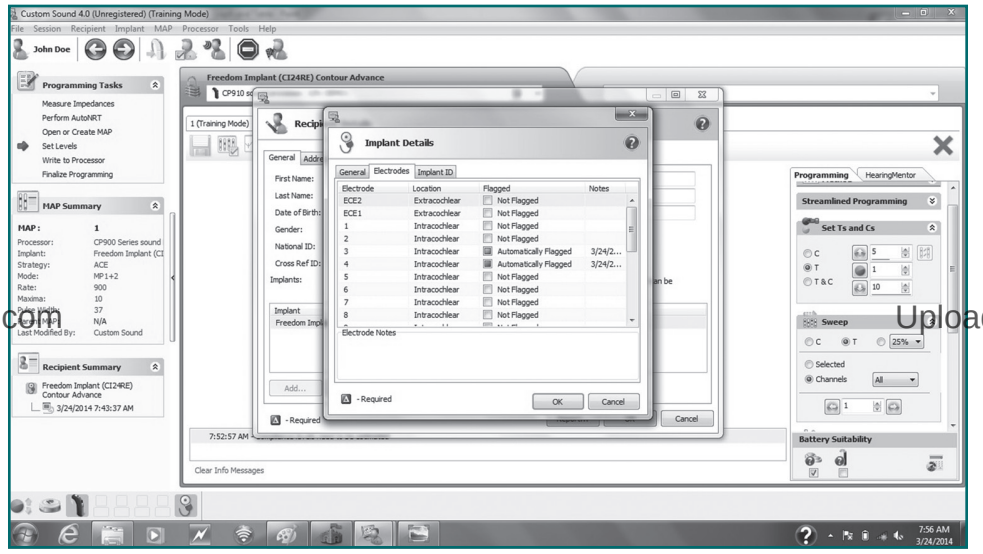
For situations in which an electrode initially had abnormal impedance prior to stimulation and was subsequently flagged, the clinician may

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reactivate the flagged electrode by accessing the "Recipient" tab at the top of the Custom Sound programming software and then selecting the "Edit" option within the "Recipient" tab. Then, the clinician must select the "Edit" option in the "Recipient Details" menu and subsequently select the "Electrodes" tab. From there, the clinician may remove the "flag" from the electrode with an abnormal impedance value (Figure 5–8). Completing this series of steps will allow for reassessment and a subsequent determination of the integrity of a suspect electrode contact.

It should be noted that some researchers of Nucleus implants reported absolute impedance values that are within the aforementioned normal values but impedance results that nonetheless are atypical/abnormal and representative of a fault within the implant. Specifically, Zwolan, Heller, and McGreevy (2012) reported on a retrospective review that they conducted with 635 recipients of Nucleus CI24R and CI24RE implants to inspect recipient performance and impedance data over the life of each recipient's implant use. They focused on electrode impedance results they defined as "atypical," which referred to results that were technically within normal limits but possessed findings that were uncharacteristic of a normal impedance assessment. The authors reported on the incidence of two types of atypical electrode impedance patterns found within Nucleus CI24R (Contour) and CI24RE (Freedom) implant recipients: (1) zig-zag (sawtooth) defined as significantly decreased impedance on three or more all even or all odd numbered electrodes, and (2) low-flat defined as significantly decreased average impedance and impedance range on five or more successive electrodes. The reader is referred to Case Studies 11 and 12 in Chapter 11 of this book for examples of both zig-zag and low-flat impedance abnormalities.

More specifically, an atypical zig-zag pattern was defined as an impedance finding in which the graphed result possesses a sawtooth appearance, with a minimum of three "spikes" that are all found on either even-numbered or odd-numbered electrodes. Furthermore, the electrode possessing the lower-value "spikes" must have impedance results that are at least 1.5 kOhms lower than both of their neighboring electrodes. The "spikes" must all fall on either even-numbered or odd-numbered electrodes. Zwolan et al. (2012) attributed the zig-zag pattern to a breach of the silicone insula-




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**FIGURE 5–8.** To reactivate a flagged electrode, select the "Recipient" tab, "Edit" option, "Edit" option again in the "Recipient Details" menu, and the "Electrodes" tab; then remove the "flag" from the electrode with an abnormal impedance value. *Note.* Provided courtesy of Cochlear™ Ltd.

tion that covers the case of the implant. All of the even-numbered electrode contacts are wired to one side of the implant case, whereas all of the odd-numbered electrode contacts are wired to the opposite side of the case. If a tear occurs on one side of the case, then body fluid ingress into that tear will potentially create a conduit for current to spread across either the even-numbered or odd-numbered electrode leads, depending on which side of the case the breach occurs. This type of fault is occasionally referred to as partial shorting.

A variety of management strategies exist for recipients who present with a zig-zag impedance finding. First, all of the electrodes with lower-numbered spike values may be disabled. When possible, the recipient may provide feedback about potential change or improvement in sound quality, and speech recognition may be evaluated to examine for improvement. Second, if the recipient notices degradation or no change in sound quality when suspect electrodes are disabled, and sound quality and speech recognition are both satisfactory, then all electrodes may remain enabled. In this case, the recipient's progress should be followed closely. Third, if the aforementioned strategies are unsuccessful, then the implant may be replaced.

According to Zwolan and colleagues (2012), a flat-low finding is defined by a number of specific characteristics. First, a flat-low finding must be determined by comparing the most recent impedance values and a "baseline" measurement obtained at least 60 days after surgery. In order to meet the criteria to be classified as "flat-low" finding, at least five electrodes must be affected. The average impedance and impedance range of the group of low impedance electrodes must both decrease by at least 30% between the "baseline" and "current" values.

For example, in a "flat-low" impedance finding, impedances can decrease to approximately one-half of the value of the impedances measured a few months postactivation. A flat-low impedance finding may result from a tear in the insulation in the middle of the case, which would allow body fluid ingress to spread to even-numbered and odd-numbered electrode contacts. If sound quality and recipient performance are satisfactory in the presence of a low-flat finding, the recipient's progress should be followed closely over time. If performance is poorer than one would expect or if sound quality deteriorates over time, implant replacement should be considered.

## Open or Create MAP

The next step involved in programming in Custom Sound 4.1 is to select the signal coding strategy, stimulation rate, maxima, and other processing parameters. This is accomplished by selecting the "Open or Create MAP" module (Figure 5–9). The clinician may select from a number of different parameters including several signal coding strategies, electrode coupling modes, stimulation rates (pulses per second per channel), maxima, and pulse width. The ACE signal coding strategy is recommended initially because previous clinical trials show that it provides superior performance over CIS and SPEAK strategies (Beynon, Snik, & van den Broek, 2003; Manrique et al., 2005; Skinner et al., 2002). Moreover, previous research suggests that most subjects have optimal speech recognition performance when using a stimulation rate of 900 pps (see below for details pertaining to a study examining performance as a function of stimulation rate for Cochlear users) and at least 9 maxima (Balkany et al., 2007; Dorman, Loizou, Spahr, & Maloff, 2002). The authors of this text recommend beginning with 10 maxima to provide sufficient detail for a speech signal embedded in competing noise and, with that number of maxima, a pulse width of 37 microseconds, which is the widest pulse width that still allows for a 900 pps stimulation rate. The authors recommend the 37 microsecond pulse width over the software default pulse width of 25 microseconds because a wider pulse width requires less current level for a given psychophysical percept, which reduces the likelihood that voltage compliance limits will be encountered. Additionally, the wider pulse width with lower current level may reduce the potential for deleterious channel interaction or nonauditory side effects (e.g., facial nerve stimulation).

Skinner, Holden, and Holden (1994, 1997) have shown that narrow frequency allocation in the midfrequency channels results in better speech recognition. With that in mind, the authors

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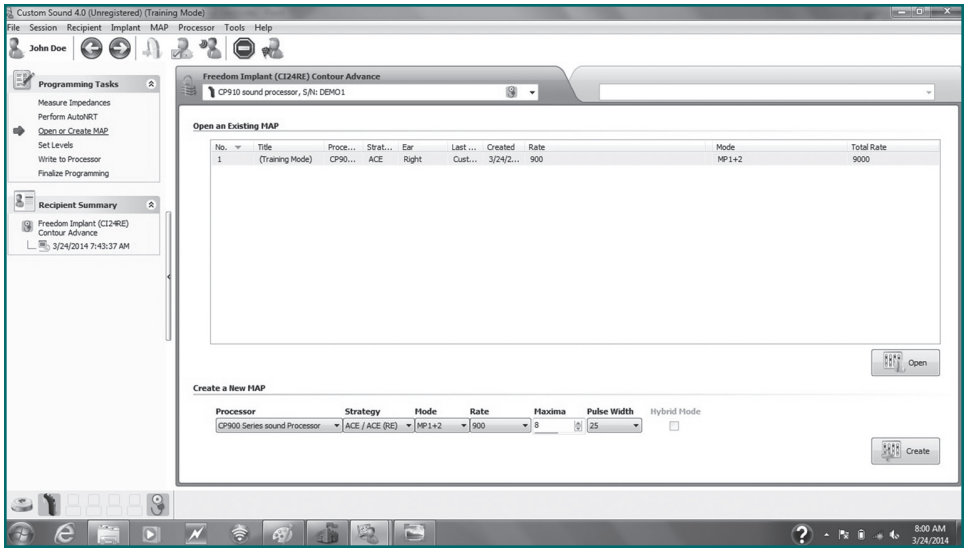



**FIGURE 5–9.** Selecting the "Open or Create MAP" module. *Note.* Provided courtesy of Cochlear™ Ltd.

also recommend setting the frequency allocation table's upper boundary out to approximately 8000 Hz for children but to 6500 to 7000 Hz for adults. The wider bandwidth is recommended for children because research has suggested that young children require an extended high-frequency response to adequately understand speech, especially when spoken by women and other children (Stelmachowicz, Pittman, Hoover, & Lewis, 2001).

Furthermore, for newly activated recipients, the authors recommend switching the volume control range from the default setting of 20% to 100%. The wider volume control range allows new recipients to select a minimal amount of stimulation, which the user may appreciate during the early acclimatization period of implant use, particularly when placing the coil on the head after a period of nonuse (i.e., placing the coil on the head after awakening in the morning). A moderate volume control range (e.g., 40%) is selected for most experienced implant users. A more detailed discussion on the effect of the volume control parameter is provided later in this chapter.

The authors recommend leaving all other processing parameters at the default settings for the initial program. If the recipient is not making satisfactory progress after approximately 1 month of use implant use, then the clinician may try alternative signal coding strategies or stimulation rates (Chapter 3). It should, however, be noted that most recipients will experience satisfactory outcomes when the signal coding parameters are set to the default settings. The authors use the aforementioned recommendations because they are supported by sound theory and because numerous recipients have experienced substantial success with the recommended settings. Additional parameters for adjustment are discussed later in this chapter, but the reader should keep in mind that in many cases, adjustments to these parameters are not necessary. Again, the provision of appropriate T and C levels is the most important factor that influences recipient success.

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Balkany et al. (2007) reported on a study that was conducted to examine the preferred stimulation rate for 55 Nucleus Freedom recipients. Subjective preference and speech recognition were evaluated for six different stimulation rates (500, 900, 1,200, 1,800, 2,400, and 3,500 pps). Sixty-seven percent of the subjects preferred a stimulation rate of 1,200 pps or lower with most subjects preferring rates of 900 or 500 pps. Only 9% of subjects preferred stimulation rates higher than 1,800 pps. Additionally, speech recognition did not improve with faster stimulation rates. The

study authors concluded that most Nucleus Freedom recipients preferred slower stimulation rates (i.e., 900 pps or lower), and there was considerable variability in the ideal stimulation rate selected across subjects. In short, this study has several clinical implications. First, Nucleus recipients tend to perform better with moderate to low stimulation rates (e.g., 250 to 1,200 pps). Second, the results of this study largely support the use of the default stimulation rate for Nucleus implants (e.g., 900 pps). Third, some recipients do perform better at alternative stimulation rates, and as a result, it may be worthwhile to consider stimulation rates other than 900 pps (i.e., default stimulation rate) for recipients who are struggling to adapt to the signal from their implant after several weeks of use.

## Setting Stimulation Levels

The majority of the work (in terms of both time consumption and in importance) involved in creating a program for a recipient is accomplished in the "Set Levels" module (Figure 5–10). This is where stimulation levels are set and where a number of programming parameters may be adjusted. It should be noted that the following discussion on the provision of stimulation levels for recipients of Nucleus implants applies to users of Nucleus 6, Nucleus 5, and Nucleus Freedom sound processors. When creating a new program in the Custom Sound 4.1 fitting platform, five channels (22, 16, 11, 6, and 1) are highlighted for measurement (Figure 5–11). The streamlined approach to programming suggests that the clinician measure stimulation levels on these five channels and interpolate the remaining values. As mentioned in Chapter 2, interpolation is only appropriate when using stimulation delivered via monopolar electrode coupling. This procedure is supported by research, which shows equivalent performance for programs created with the streamlined and conventional approaches (Plant et al., 2005). However, to detect any potential variation across the array, the authors of this textbook also recommend measuring stimulation levels on at least one channel between each of these five highlighted channels as well as on channels 2 and 3. Furthermore, if more than a 10 current level (CL) difference exists between the most closely spaced measure channels, then the authors recommend measuring a channel between the two channels in which the relatively large difference exists for T level.

It should be noted that the unit of stimulation in the CustomSound 4.1 software platform is referred to as a current level (CL) or clinical

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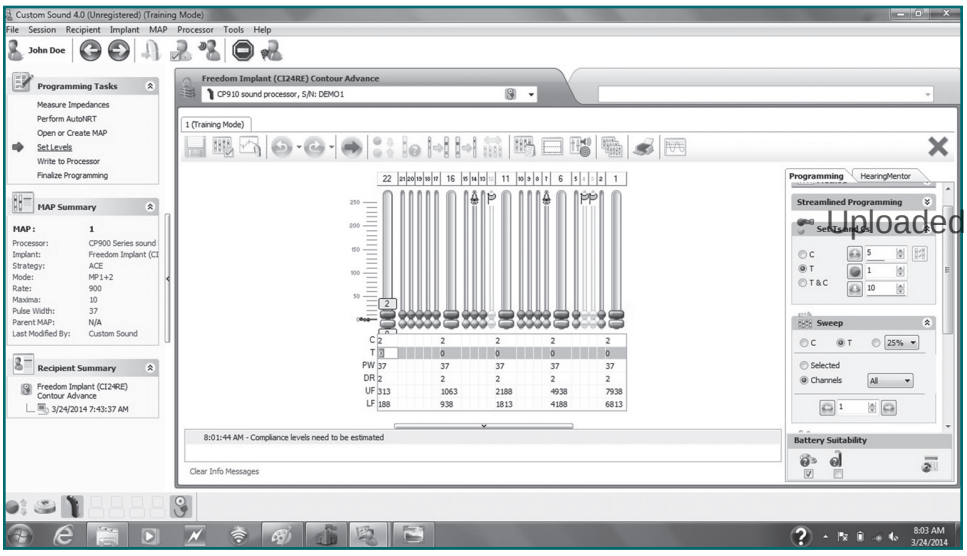



**FIGURE 5–10.** "Set Levels" module. *Note.* Provided courtesy of Cochlear™ Ltd.

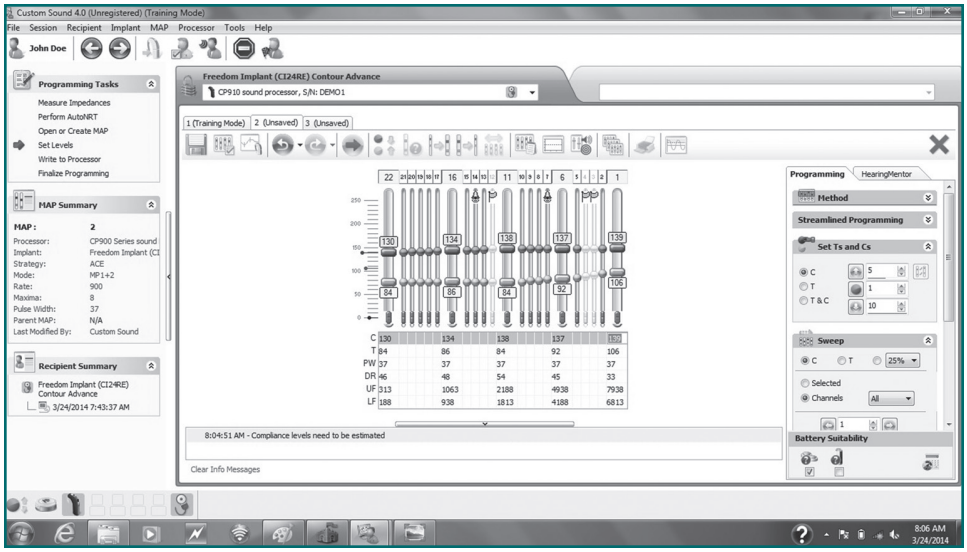



**FIGURE 5–11.** Five channels highlighted (22, 16, 11, 6, and 1) for measurement of stimulation levels. *Note.* Provided courtesy of Cochlear™ Ltd.

unit. The abbreviation, "CL," will be used for the remainder of this chapter. The CL represents the current amplitude of the programming stimulus at the specified pulse width. The programming stimulus is a train of biphasic, rectangular pulses presented at a particular stimulation rate (e.g., default stimulation rate of 900 pulses per second). Stimulation level changes occur logarithmically and range from 0 CL (an amplitude of 0 microamperes/phase) to 255 CL (an amplitude of 1.75 milliamperes/phase) at a given pulse width. The pulse width may vary from 9 to 400 microseconds/phase.

At the following appointment, the clinician should attempt to measure stimulation levels across a variety of channels so that, by one month of implant use, measurements are made on every channel at least one time. This action allows the clinician to determine any channels that may possess particularly unusual levels. Alternatively, the clinician may measure stimulation levels on the same subset of channels at each session and then balance loudness at C level across the electrode array and sweep at T level across the array to ensure audibility across all channels. The latter approach may not be feasible with children, but it achieves the objective of identifying anomalous channels. It should be noted that measured channels should be highlighted, so the clinician knows the channels on which T and C levels have been obtained. Highlighting measured channels is achieved by "clicking" (single click) on the channel or selecting "v" on the keyboard while the channel is being measured. A highlighted channel is represented by a yellow square (rather than white) encompassing the channel number.

As discussed in Chapters 2 and 3, the measurement of accurate T and C levels is crucial in providing a high-quality program. For the Nucleus 6 system, the T level is the lowest level of stimulation that the user can detect 100% of the time. At default settings, single-channel T and C level measurements are made with the presentation of a train of biphasic pulses with a stimulation rate of 900 pps and a duty cycle of 500 msec. Recipients typically perceive these stimuli as similar to pure tones used during assessment of audiometric threshold. Typically, the authors present two trains of pulses each time the signal is presented (i.e., 500 msec on—500 msec off—500 msec on). However, for a recipient who is experiencing difficulty distinguishing the signal from his or her tinnitus, the authors may vary the number of pulse trains (usually between 1 to 5) that are presented and ask the recipient to count the number of "beeps" he or she hears. As previously

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mentioned, the "count-the-beep" method allows the clinician to determine whether the recipient is truly hearing the test signal, because he or she must report the correct number of pulse trains presented in order to indicate audibility.

Appropriate alternatives for measuring T levels were discussed in detail in Chapter 3, but typically, the authors of this book use a "clinical-bracketing" (i.e., Hughson-Westlake) approach with two stimulus presentations at each CL. At initial stimulation, a 5 CL ascending step size and a 10 CL descending step size is used. For experienced adults and attentive children, ascending and descending step sizes may be set to 2 and 4 CL, respectively. Furthermore, small step sizes should generally be used for all recipients who have narrow dynamic ranges (e.g., less than 15 CL).

When T levels are set too high, the user may complain of ongoing noise. This noise may arise from two sources. First, if T levels are set too high, the recipient may perceive a constant "buzzing" or "humming" sound secondary to supra-threshold low-level electrical stimulation. This is an unwanted percept, and the clinician should attempt to eliminate it by decreasing T levels. Second, this noise may arise from the perception of ambient noise that normal-hearing persons hear every day but that newly implanted persons may not have heard for quite some (i.e., perception of ambient noise ranging 25 to 50 dB SPL, such as the room's ventilation system or the fan on a computer). This percept is actually quite typical and intended, and the clinician should encourage the recipient to attempt to acclimate to the low-level sound(s).

For many recipients, it is difficult to determine if the constant, low-level noise they areperceiving is attributed to excessive T-level stimulation or to audibility for low-level ambient sounds. One way to determine the culprit is to take the recipient inside a sound-treated audiometric test booth to see whether he or she is able to detect the sound. If the sound is eliminated, then it is likely the recipient was hearing low-level room noise, and he or she should be counseled to acclimate. Another way to troubleshoot this complaint is to increase the floor of the instantaneous input dynamic range (IIDR) by increasing the T-SPL parameter (e.g., from the default of 25 dB SPL to 35–40 dB SPL) or decreasing the sensitivity. If the unwanted perception persists, the clinician may infer that the recipient was hearing constant suprathreshold electrical stimulation, and T levels should be decreased globally. However, if the constant sound decreases or is diminished, the clinician may infer that the recipient was simply responding to newly audible ambient room noise, and the clinician should encourage the recipient acclimate to the normal ambient noise. It is worth mentioning that a constant noise secondary to excessively high T levels is possible with Nucleus recipients, but the use of the ACE (*n*-of-*m*) signal coding strategy does reduce the likelihood of this becoming a significant problem as what might occur with CIS-based signal coding strategies.

Of course, it should be noted that the clinician should evaluate warble tone thresholds once a stable program is established with the goal of aided sound-field thresholds falling between 15 to 30 dB HL. Children should have warble tone thresholds sound-field thresholds no greater than 25 dB HL, while adults should have warble tones sound-field detection thresholds of no greater than 30 dB HL. If sound-field thresholds are substantially better than 15 dB HL, it may be prudent to consider slightly decreasing T level stimulation. If sound-field detection thresholds are poorer than the aforementioned goals, then the clinician should remeasure T levels with the goal of increasing T level on the same channels that correspond to the frequencies at which sound-field thresholds were elevated.

The upper-stimulation levels, C levels, should be perceived by the user as "loud but comfortable." Approaches to measuring C levels were discussed in Chapter 3, but briefly, C levels may be set using a variety of techniques. When activating adult recipients, the authors of this text typically increase C levels globally from the T level measurements in live speech mode (e.g., the processor microphone is activated so the recipient is able to hear speech and ambient environmental sounds) until the recipient reports that an optimal loudness and sound quality have been achieved. Then, loudness balancing is conducted two channels at a time across the array to ensure that loudness is equalized from channel to channel. The provision of equal loudness across channels typically optimizes sound quality and speech recognition.

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After loudness balancing is completed, the C level profile (e.g., shape across the array) is usually flatter than the T-level profile. In fact, as a general rule for Cochlear recipients, the C level profile is usually flatter with less variation across the array than the T level profile. After balancing is completed, the authors will once again adjust C levels globally (in 1–2 clinical-unit steps) in the live speech mode until the subject reports that loudness and sound quality are optimal. It is important to note that the authors typically will increase C levels to a point that just exceeds the recipient's most comfortable listening level in live speech mode or to a point that begins to degrade sound quality. Then, C levels are globally decreased back to an overall level that provides optimal listening comfort and sound quality. Increasing C levels to a level that just exceeds optimal comfort and sound quality allows the recipient to be certain that an optimal listening level and quality has been achieved.

For experienced recipients, the electrical dynamic range (range in CL between T level and C level) typically ranges between 25 to 30 to 60 CL with most recipients possessing an electrical dynamic range around 40 to 50 CL (Botros, Banna, & Maruthurkkara, 2013). Some adult recipients who have a long duration of high-frequency hearing loss will have a narrower electrical dynamic range (15 to 30 CL) for high-frequency electrodes. The clinician should be leery of electrodes for which the electrical dynamic range is significantly different from the typical range of 30 to 60 CL. Although the authors provide clinical services for recipients who have electrical dynamic ranges approaching 70 to 80 CL who achieve excellent performance with their cochlear implant, these cases are certainly the exception. The clinician should be especially guarded against electrical dynamic ranges that are wider than normal in children as studies have suggested that clinicians may be more apt to provide too much stimulation for children relative to adults (Zwolan & Overstreet, 2005).

Likewise, the clinician should be concerned when the electrical dynamic range is very narrow (e.g., less than 20 CL), particularly when it is narrow across the array. In these cases, it may be prudent to attempt to gradually increase the electrical dynamic range over an extended period of time (i.e., increase the electrical dynamic range by 2 to 3 CL every 4 to 6 weeks over a 6-month period). Of course, some recipients may not be able to tolerate this type of increase, even when the clinician attempts to provide it gradually across time. In these cases, the clinician should avoid providing a signal that is uncomfortable. Regardless of the electrical dynamic range, the clinician should gather outcome measurement data to ensure that a recipient's program is appropriate. Outcome measurements should include word recognition, sentence recognition in noise, sound-field warble tone thresholds, and an assessment of subjective benefit/performance for recipients who can complete these measures. For younger children, outcome measures should include subjective questionnaires completed by caregivers (e.g., LittlEars, PEACH, etc.) as well as informal and formal assessment of speech, language, and auditory skill development by the child's Listening and Spoken Language Specialist or speech-language pathologist. Concern pertaining to deviant stimulation levels and/or an electrical dynamic range is heightened when outcomes measurements suggest poor performance and progress.

Given the aforementioned discussion of stimulation levels and electrical dynamic range, it is helpful to administer a measure that serves as an objective guide to where C level should be set. The electrically evoked stapedial reflex threshold (ESRT) has been shown to correlate strongly to C level (Brickley et al., 2005; Hodges et al., 1997; Lorens, Walkowiak, Piotrowska, Skarzynski, & Anderson, 2004). The authors typically measure the ESRT by presenting three pulse trains (500 msec on-off duty cycle) and examining for a time-locked change in admittance (Figure 5-12). For adults who are able to hold steady and quiet for a relatively long period in order for the measure to be completed, the authors use an up-2/down-4 CL bracketing technique to acquire threshold. For younger children who are less likely to hold still for a long period of time, the authors ascend and descend by 3 to 5 and 6 to 10 CL, respectively. For cooperative recipients, the ESRT is measured for all even channels as well as for channel 1. For recipients who have a difficult time holding still for several minutes at a time, the ESRT is mea-

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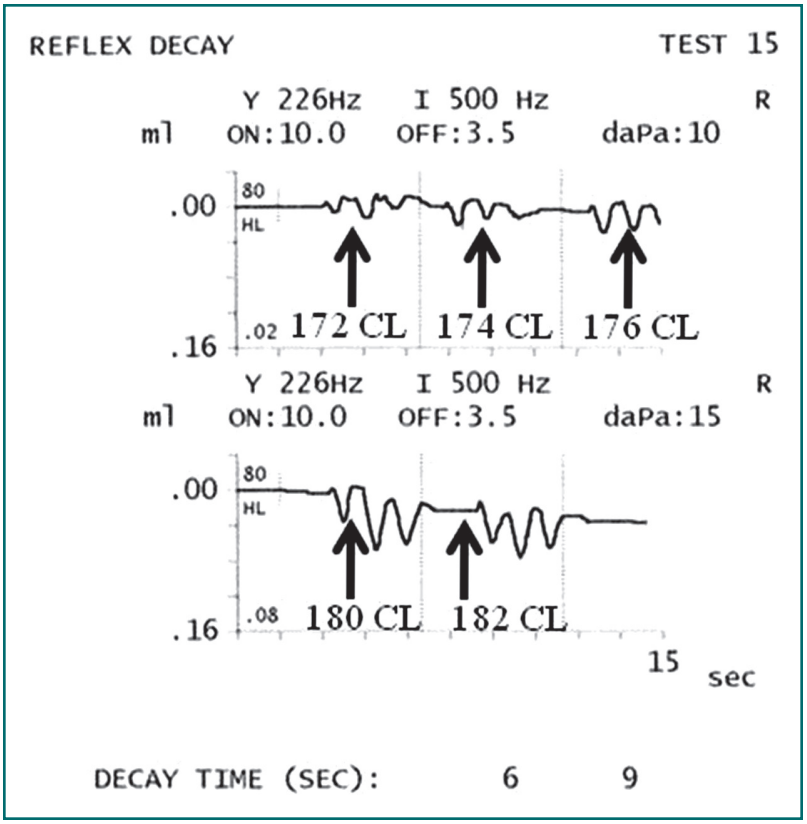



**FIGURE 5–12.** Measurement of the electrically evoked stapedial reflex threshold (ESRT) by presenting three pulse trains (500 msec on-off duty cycle) and examining for a time-locked change in admittance.

sured on channels 22, 3, 11, 1, 16, and 6. Then, the ESRT is completed on remaining channels as time and the recipient's cooperation allow. Unmeasured channels are interpolated based on the values set for the measured channels.

Once the ESRT is obtained for the desired channels, global C levels are decreased substantially (e.g., typically by 40 to 50 CL), the processor is activated in live speech mode, and then global C Levels are adjusted to the recipient's most comfortable listening level while he or she listens to speech and environmental sounds. Typically, optimal performance and sound quality are achieved when C levels are 10 to 20 CL below the ESRT values. When C levels are more than 20 CL below the ESRT for experienced cochlear implant users (>3 months of experience), the clinician should attempt to gradually increase stimulation levels over an extended time (e.g., several weeks to several months) until they are within 15 to 20 CL of the ESRT. On the other hand, if C levels exceed the ESRT, the clinician should counsel the recipient to accept a lower-stimulation setting that is within 5 CL of the ESRT. In most cases, once the recipient has time to adjust to the lower, ESRT-based C levels, performance will improve. In addition, the clinician should avoid setting C levels above ESRT, as doing is more likely to lead to overstimulation (i.e., discomfort, excessive electrical stimulation, and/or distortion of speech).

Although the aforementioned techniques are the authors' preferred strategies to set C levels, other methods do exist. The clinician may use psychophysical loudness scaling to set C levels to a "loud but comfortable" percept in several channels across the electrode array. The C levels for unmeasured channels are interpolated based on measured values. Then, C levels are decreased

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and the program is activated in live speech mode. Next, the clinician globally increases C levels to the recipient's most comfortable listening level while in live speech mode.

The reader should remember that increases in stimulation in the Custom Sound 4.1 software occur in a logarithmic fashion and may cause relatively large increases in loudness growth with small changes to C levels, particularly as stimulation levels approach the recipient's upper level of loudness comfort. While making global adjustments in live speech mode, small increases (e.g., 1–2 CL) may result in fairly large changes in loudness, clarity, and sound quality. As a result, it is important to make global adjustments in precise steps. Finally, the stimulation levels the recipient experiences while in live speech mode or while in everyday use are influenced by the volume control. The effect of the volume control parameter is discussed in detail in a later section of this chapter. The reader should pay special attention to the influence of the volume control on stimulation level.

### *Parameters That Influence How Acoustic Inputs Are MAPped into the Recipient's Electrical Dynamic Range*

The previous section described how upper-stimulation levels should be set for users of Cochlear Nucleus implants. In summary, T levels represent the lowest level of electrical stimulation the recipient can detect for each channel, and C levels represent the level of electrical stimulation at each channel that is loud but comfortable to the recipient. Collectively, the T and C levels define the user's electrical dynamic range (EDR). As mentioned in Chapter 2, a significant challenge facing engineers and designers of cochlear implants is the task of delivering a wide range of acoustic inputs (e.g., from whisper soft sounds of 20 dB SPL to very loud sounds of around 100 dB SPL,) into the recipient's relatively narrow EDR (typically around 20 dB). The following section describes several parameters within the Cochlear Custom Sound programming platform that influence how a wide range of desirable acoustic inputs are delivered to an appropriate location within the narrow range of a recipient's electrical dynamic range. These parameters are all interrelated, so it is somewhat difficult to explain the effect that a given parameter has on the sound a recipient receives without having a thorough understanding of each of the other parameters. As a result, it may be necessary for the reader to review to following section more than once to acquire a full understanding of the influence of each parameter on the recipient's listening experience as well as how each parameter is interconnected in the sound processing of Cochlear sound processors.

It should be noted that the following discussion applies to the programming of Nucleus 6, Nucleus 5, and Nucleus Freedom sound processors. Also, most of the following programming parameters may be adjusted in the "Parameters" tab, which is located in the "Set Levels" module (Figure 5–13).

**IDR, IIDR, T-SPL, C-SPL, AGC, Sensitivity, and Autosensitivity Control.** The input dynamic range (IDR) is not actually a programmable parameter found in the Custom Sound programming software, but it is imperative to define it here and distinguish it from another parameter, the instantaneous input dynamic range (IIDR), which is an important parameter within Custom Sound. Cochlear considers the IDR to refer to the maximum range of acoustic inputs that may be electronically captured by the sound processor without clipping or significant distortion. By this definition, the low end of the IDR for broadband sounds is determined by the noise floor of the microphone and is approximately 23 dB SPL. The upper end of the IDR may be determined by the limits of the A/D converter (i.e., the number of bits of the DSP). The upper range of the IDR of the Nucleus 6 sound processor is just over 100 dB SPL for broadband inputs, which results in an overall IDR of just shy of 80 dB. It should be noted that the lower end of the IDR is actually lower for narrowband inputs (i.e., tones), especially in the mid-frequency range (1000 to 3000 Hz). This is due to the fact the noise (i.e., microphone noise) present in any one channel is lower than the noise summed across all channels simultaneously for broadband inputs. As a result, the lower end of the IDR may reach as low as 9 dB SPL for narrow-

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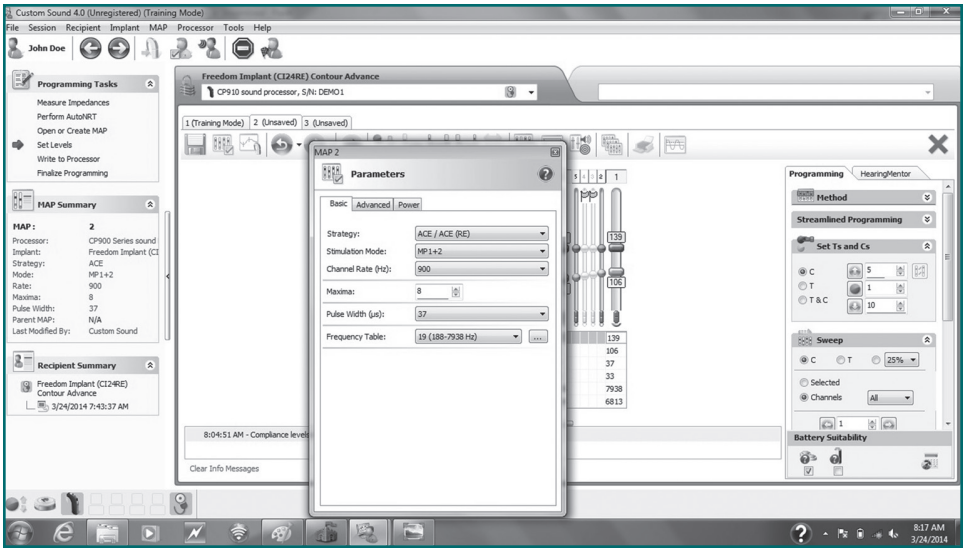



**FIGURE 5–13.** Parameter adjustments may be done in the "Parameters" tab located in the "Set Levels" module. *Note.* Provided courtesy of Cochlear™ Ltd.

band inputs in the mid-frequency portion of the speech range.

As mentioned in Chapter 2, the EDR of a typical recipient is usually around 20 to 30 dB, and as a result, some form of compression processing is needed to place this wide range of potential acoustic inputs (approximately 80 dB SPL for broadband inputs) at the front end of the processor into the narrow EDR. The signal processing engineers at Cochlear have developed a method to provide minimal fast-acting compression to a static speech signal while also allowing access to wide range of dynamic signals (e.g., 20 to 100 dB SPL for broadband inputs). In order to understand how this is accomplished, it is important to define some additional terminology.

The IIDR is a parameter label that is specific to Cochlear Ltd. implants and is defined as the range of short-term, instantaneous fluctuations that are mapped without compression (or other types of attenuation, such as Autosensitivity™ Control [ASC]) into the recipient's electrical dynamic range at a given point in time. The default IIDR of contemporary Cochlear processors (Nucleus 5, Nucleus 5, and Nucleus Freedom) is 40 dB, which closely corresponds to the dynamic range of conversational level speech. More specifically, when a talker speaks at an average conversational level, the peaks of the speech signal are near 65 dB SPL, whereas the minima of the speech signal fall near 25 dB SPL.

The default IIDR is defined by two parameters referred to as T-SPL and C-SPL. At default settings, the T-SPL is set at 25 dB SPL, which results in T-level stimulation for acoustic inputs of 25 dB SPL, while inputs below 25 dB SPL receive stimulation below T level. If T levels are set appropriately (i.e., at a level that is just audible to the recipient), then inputs below 25 dB SPL should be inaudible to the user, whereas inputs at 25 dB SPL should be audible but very soft to the user.

At default settings, C-SPL is set to 65 dB SPL, which results in acoustic inputs at 65 dB SPL receiving C level stimulation if the volume control of the sound processor is set to maximum position. All inputs that exceed 65 dB SPL are subjected to fast-acting infinite compression from a fast-acting automatic gain control (AGC) compressor with a high compression ratio. Acoustic inputs between 25 to 65 dB SPL receive electrical stimulation that falls between the T and C Levels of the recipient.

It should be noted that the AGC of the Nucleus 6 sound processor is more sophisticated than the AGC of the Nucleus 5 sound processor. In

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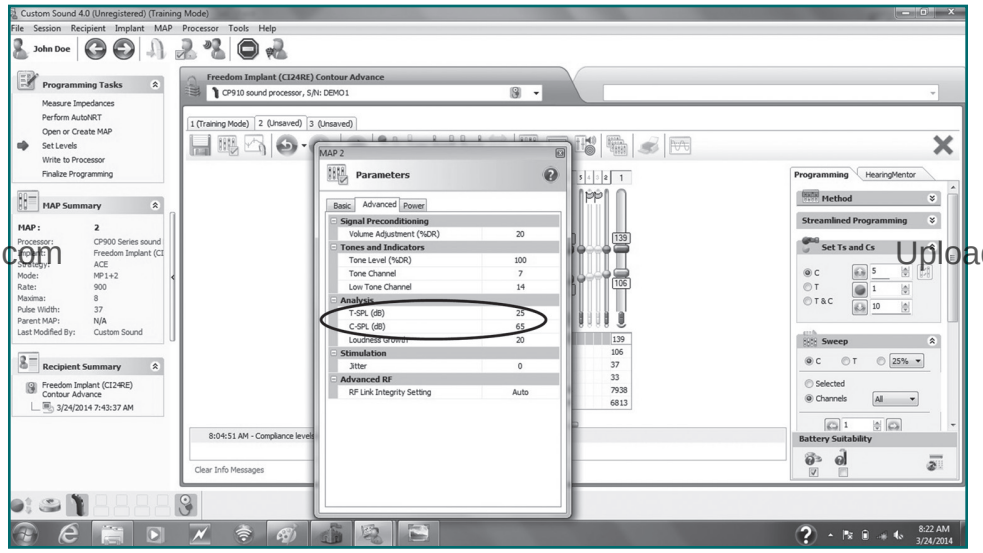
the Nucleus 5 sound processor, the AGC is a fast-acting mechanism that quickly attenuates high-level sounds with compression with a relatively high compression ratio. This may be undesirable if it distorts intensity/temporal cues that exist for high-level sounds. As a result, Cochlear developed a dual-action AGC for the Nucleus 6 sound processor. Moderate-level inputs are subjected to medium-speed (e.g., medium-acting time constant) compression with a modest compression ratio, whereas high-level sounds are subjected to fast-acting compression with relatively high compression ratios. This dual-action AGC is designed to minimize distortion in response to the dynamic nature of everyday speech and environmental sounds.

The clinician may adjust the T-SPL (from 9 to 50 dB SPL) and C-SPL (from 65 to 84 dB SPL) parameters within the "Advanced" tab of the "Parameters" menu (Figure 5–14). However, as long as the clinician counsels patients to allow for an adjustment period to new sounds, adjustments to T-SPL and C-SPL parameters are rarely needed. Indeed, a significant adjustment to these parameters alone may cause significant detriment to the listening experience of the recipient.

There are very few situations in which the clinician may consider adjusting the T-SPL/C-SPL parameters. For example, when adults cannot acclimate to hearing low-level ambient noise (e.g., ventilation systems), the T-SPL parameter may be slightly increased to reduce the salience of low-level sounds. However, in most cases, adults do very well with a T-SPL of 25 dB, and newly implanted adults who complain of hearing low-level sounds will typically acclimate to their newfound audibility with a short period of implant use. It should be stressed that it is unlikely that the T-SPL should ever be increased for young children.

Adjustments to C-SPL are generally not desired because decreases will result in compression of everyday inputs, and increases of the C-SPL without adjustment of other parameters (e.g., loudness growth, C Levels, etc.) will often make conversational speech too soft. However, if C-SPL is increased, then it is necessary for the clinician to remeasure C levels to ensure that speech and environmental sounds are perceived with an appropriate loudness. It may also be beneficial to decrease the "Loudness Growth" from 20 to 15 after C-SPL is increased but before the C levels are remeasured. Again, it is worth reiterating that the C-SPL and "Loudness Growth" parameters rarely, if ever, need to be adjusted to optimize recipient performance.

The default IIDR of 40 dB with a T-SPL and C-SPL of 25 and 65 dB SPL, respectively, is very appropriate for use for listening to average con-




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**FIGURE 5–14.** Adjust the T-SPL and C-SPL parameters (*circled*) in the "Advanced" tab of the "Parameters" menu. *Note.* Provided courtesy of Cochlear™ Ltd.

versational level speech in a quiet environment because the minima of the speech signal, which occur near 25 dB SPL, should be audible for the recipient. Likewise, little to no compression should be provided to the more intense components of speech, and as a result, the intensity cues of speech should be fairly well preserved. However, this IIDR would not be appropriate for all situations, as some components of soft-spoken speech would fall below 25 dB SPL and would not elicit stimulation. Also, loud speech and other high-level sounds would be subjected to fast-acting, high-level compression, which would likely compromise the identification and sound quality of these sounds.

In order to provide the recipient with access to a wide range of acoustic inputs without distortion and excessive compression, the IIDR window may be shifted in order to capture inputs below 25 dB SPL and to prevent excessive compression for all inputs exceeding 65 dB SPL. The sensitivity parameter controls the overall gain applied to the input signal from the microphone or direct auditory input (DAI). The sensitivity setting, which ranges from a minimum of "1" to a maximum of "20," may be adjusted by the recipient (if the clinician provides the recipient with control over the sensitivity parameter), or it may be set to a fixed level by the clinician. The default sensitivity setting is "12," which results in the IIDR being positioned from 25 dB SPL (T-SPL) to 65 dB SPL (C-SPL). Figure 5–15 illustrates the effect of the sensitivity setting on the positioning of the IIDR. As shown, increasing the sensitivity setting moves the IIDR to a lower level, which provides the user with better access to low-level inputs but also results in a higher likelihood of high-level/excessive compression for moderate level inputs (e.g., average conversation level speech). In contrast, decreasing the sensitivity setting prevents moderate to high-level inputs from being over-compressed, but low-level inputs will be inaudible to the recipient.

ASC is a type of input processing that automatically adjusts the sensitivity setting when the ambient noise level at the processor input exceeds a default activation threshold of 57 dB SPL. In most moderate to high-level noise environments, talkers raise their voices so that their level exceeds the competing noise and the recipient is able to listen at a positive signal-to-noise ratio (SNR). ASC prevents excessive compression of the speech signal in noisy environments and enhances the likelihood of preserving a positive SNR in noisy listening conditions. Therefore, the total range of input levels the processor can encode without significant clipping or distortion is much wider than the IIDR.

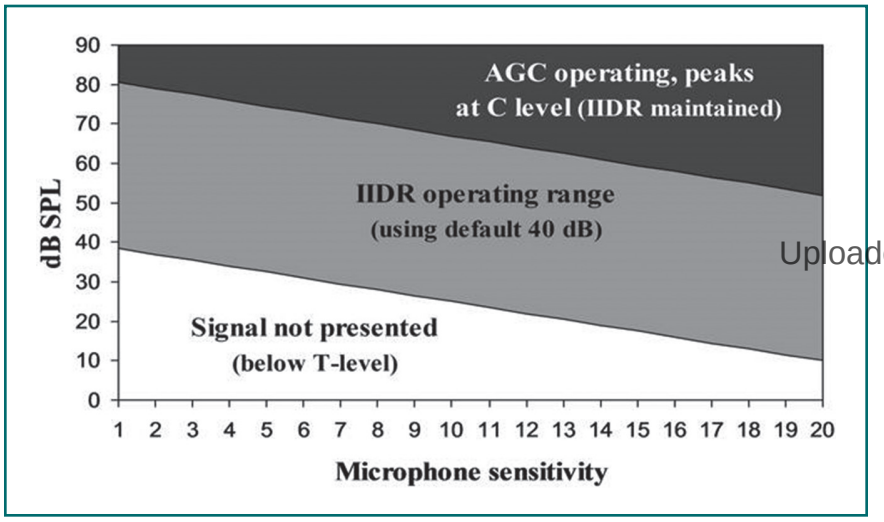



**FIGURE 5–15.** Effects of the sensitivity setting on the positioning of the instantaneous input dynamic range (IIDR) parameter. *Note.* Provided courtesy of Cochlear™ Ltd.

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Numerous studies have shown that use of ASC results in an improvement in the recipient's ability to understand speech in noise with no change in speech understanding in quiet (Gifford & Revit, 2010; Wolfe et al. 2009; Wolfe, Schafer, John, & Hudson, 2011; Wolfe et al., 2012). As a result, the clinician should consider enabling ASC for the recipient's primary program to be used in most typical listening situations. It should be noted that ASC is enabled/disabled in the "Write to Processor" module, which is discussed later in this chapter.

In most cases, Nucleus recipients do not have access to manual adjustment of the sensitivity control, and ASC (along with ADRO, which will be discussed later) is used with a fixed default sensitivity setting of "12" to provide access to a wide range of speech and environmental sounds. Clinicians may program the CR210 and CR230 remote controls to allow the recipient to adjust the volume control or the sensitivity. However, in the typical case, remote controls are programmed to allow for adjustment of the volume control. Figure 5–15 does show the effect of a manual sensitivity adjustment on the recipient's IIDR. It should be noted that a fixed sensitivity setting of "12" with ASC should always be used with young children and other recipients who are unable to appropriately adjust sensitivity settings manually. Because sensitivity is typically a fixed parameter, but the volume parameter is oftentimes recipient-adjustable, the following section of this chapter describes, in detail, the effect of the volume control.

The reader may ask when it would be appropriate to manually adjust the sensitivity control. Examples of situations in which recipients report that manual adjustment may be helpful include: (1) when speech and noise are too soft with ASC responding to background noise, it may be helpful to increase the sensitivity setting, (2) when better access to sounds originating from a distance (e.g., nature sounds when on a walk) is desired, it may be helpful to increase the sensitivity setting, and (3) when competing noise in a moderate-level noise environment continues to cause difficulty with speech recognition, it may be helpful to decrease the sensitivity setting even further. However, it should be stated again that the need for manual sensitivity adjustments are very infrequent, when ASC is active, and as a result, the authors prefer to fix sensitivity setting at "12" and enable ASC for automatic adjustment of the sensitivity setting (Wolfe et al., 2009, 2011, 2012).

It should be noted that additional types of input processing also influence how acoustic inputs are delivered to a recipient's electrical dynamic range. For instance, ADRO is another adaptive form of input processing, which seeks to optimize the delivery of acoustic inputs to improve audibility for low-level inputs, while optimizing comfort for high-level inputs. Whisper is another type of input processing that aims to enhance the salience of low-level inputs. ASC, ADRO, Whisper, and other type of input processing are described in a later section of this chapter.

**Volume Control.** The volume control parameter has a large influence on the stimulation levels the recipient uses in his or her programs, and subsequently, it has the potential to substantially impact speech recognition and sound quality. As a result, it is imperative that the clinician has a thorough understanding of the function of the volume control parameter as well as the impact of the volume control setting on the recipient's daily listening experience.

The volume control may be adjusted across 10 steps (from 1–10). The range over which the volume control adjusts C levels is programmable from 0 to 100% and is controlled by the "Volume Adjustment (%DR)" parameter, which is found in the "Advanced" tab of the "Parameters" menu (see Figure 5–14). It is important to remember that when a recipient is using his or her sound processor in a quiet environment at default settings (i.e., T-SPL, C-SPL, and sensitivity settings are all set to default), acoustic inputs at 25 dB SPL result in the provision of stimulation at T level, and acoustic inputs at 65 dB SPL and higher result in the provision of stimulation corresponding to the volume control setting (it should be noted that inputs exceeding 65 dB SPL are subjected to high-level compression and also result in stimulation corresponding to the volume control setting). The volume control parameter essentially has the effect of adjusting the upper level of stimulation the recip-

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ient receives (i.e., the amount of stimulation for inputs >65 dB SPL). When the volume control is set to the maximum setting of "10," the recipient is provided with stimulation at the C level set for his or her MAP. Decreasing the volume control from the maximum position decreases the stimulation level the recipient receives for inputs of 65 dB SPL and higher.

The extent by which the volume control setting adjusts the stimulation level that the recipient receives for inputs equal to or exceeding 65 dB SPL is determined by the parameter referred to as "Volume Adjustment (%DR)" (hereafter referred to as "Volume Adjustment"). When the Volume Adjustment parameter is left at the default setting of 20%, decreasing the volume control from the maximum (i.e., 10) to the minimum (i.e., 1) results in a global reduction in C levels that corresponds to 20% of the recipient's electrical dynamic range. Therefore, a participant with an electrical dynamic range of 30 CL would experience a 6 CL decrease in stimulation when adjusting the volume control from 10 to 1 (30 CL * 20% = 6 CL). Table 5–1 provides examples of the effect of the volume control and the Volume Adjustment on the stimulation levels a recipient receives.

Several important considerations can be gleaned from Table 5–1. Specifically, wider Volume Adjustment ranges not only result in an overall wider effect of the volume control range, but each step adjustment of the volume control results in a relatively large change in the stimulation the recipient receives for moderate to high-level inputs. Note that for recipient #2 in Table 5–1, a Volume Adjustment range of 100% results in a change of 5.56 CL for each step adjustment of the volume control, while a Volume Adjustment range of 20% results in a change of 1.11 CL for each step adjustment of the volume control. In other words, small Volume Adjustment settings (i.e., 30% or less of DR) allow for more precise control over the stimulation the recipient receives for moderate to high-level inputs. Large changes to volume settings should not be necessary for experienced

**Table 5–1.** Stimulation (in CL) Recipient Receives for Inputs Equaling or Exceeding 65 dB SPL

| Recipient | T Level | C Level | DR | Volume Adjustment (% DR) | VC Setting | | | | | | Step Size/VC Setting |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | | | | | 1 | 2 | 4 | 6 | 8 | 10 | |
| 1 | 120 | 135 | 15 | 100% | 120 | 121.7 | 125 | 128.3 | 131.7 | 135 | 1.67 |
| 1 | 120 | 135 | 15 | 50% | 127.5 | 128.3 | 130 | 131.7 | 133.3 | 135 | .83 |
| 1 | 120 | 135 | 15 | 20% | 132 | 132.3 | 133 | 133.7 | 134.3 | 135 | .33 |
| 1 | 120 | 135 | 15 | 0% | 135 | 135 | 135 | 135 | 135 | 135 | 0 |
| 2 | 120 | 170 | 50 | 100% | 120 | 125.6 | 136.7 | 147.8 | 158.9 | 170 | 5.56 |
| 2 | 120 | 170 | 50 | 50% | 145 | 147.8 | 153.4 | 158.9 | 164.5 | 170 | 2.78 |
| 2 | 120 | 170 | 50 | 20% | 160 | 161.1 | 163.3 | 165.6 | 167.8 | 170 | 1.11 |
| 2 | 120 | 170 | 50 | 0% | 170 | 170 | 170 | 170 | 170 | 170 | 0 |
| 3 | 120 | 220 | 100 | 100% | 120 | 131.1 | 153.3 | 175.6 | 197.8 | 220 | 11.11 |
| 3 | 120 | 220 | 100 | 50% | 170 | 175.6 | 186.7 | 197.8 | 208.9 | 220 | 5.56 |
| 3 | 120 | 220 | 100 | 20% | 200 | 202.2 | 206.7 | 211.1 | 215.6 | 220 | 2.22 |
| 3 | 120 | 220 | 100 | 0% | 220 | 220 | 220 | 220 | 220 | 220 | 0 |

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implant users as long as C levels are set appropriately, and as a result, the Volume Adjustment range should be relatively narrow (e.g., 30–40%). A narrower Volume Adjustment range will allow experienced recipients to have more precise control over the stimulation they receive for moderate to high-level inputs. However, broader control over the C levels may be helpful to new implant users, particularly when they first put the processor on in the morning or after a period of nonuse.

Review of Table 5–1 also indicates that the effect of the volume control varies as a function of the size of the recipient's EDR. As shown, for recipients with narrow EDR, each step of the volume control adjustment results in a small change (in CL) in the stimulation provided to the recipient. In contrast, for recipients with wide EDR, each step of the VC results in a relatively large change (in CL) of stimulation. As a result, the clinician should consider the recipient's EDR when selecting the Volume Adjustment parameter.

At initial activation, the authors typically provide a wide volume control range to allow the recipient or caregiver to adjust the volume down by 100% of their dynamic range. Parents are advised to start with the volume setting of "0" when the child first wakes up from sleep. The external transmitting coil should then be placed on the child's head and the volume of the processor should be increased to the desired setting (typically "10") over the next few seconds. If the child reacts negatively to the signal, the volume control should be adjusted to a comfortable setting, and the parent should attempt to increase the volume at a later point in the day. If the child shows no reaction, the next time the processor is placed on the child's head, the parent should begin at a volume setting of approximately "2" to "3" and gradually increase the starting volume setting over the next few days or weeks. Once the recipient is able to tolerate the program with the volume control at a stable position, the range of the Volume Adjustment parameter should be decreased or the control should be disabled (i.e., set to 0%).

For most children who perform well and have stable stimulation levels, the authors of this text disable the volume control. If the family reports that the child occasionally reports that stimulation from the implant is too loud, then we will provide a narrow range for adjustment (e.g., 20 to 30%) to allow the caregiver to provide a small decrease in stimulation in relatively precise steps. The family is encouraged to contact the clinic if the volume control must be substantially decreased to facilitate the child's comfort or if the child consistently complains of loudness discomfort. Parents are able to conveniently adjust the volume setting via the CR210 or CR230 wireless assistant; however, these changes must be warranted to justify alteration to the child's carefully programmed upper-stimulation levels. As previously discussed, the clinician should consider disabling the volume control for young children who have stable program levels. This may be accomplished by disabling the volume control in the Processor Configuration menu (to be discussed later) or by setting the Volume Adjustment range to 0% in the "Advanced Parameters" section of the "Set Levels" screen (see Figure 5–14). The sensitivity control may be locked in the Processor Configuration menu (to be illustrated later) and should be routinely disabled for young children. There may be, however, some circumstances in which older children will desire to access the volume control. The volume control buttons of the Nucleus 6 can be locked by using the CR230 wireless assistant. Likewise, the control buttons of the CR210 and CR230 also may be locked.

For adults, the C levels are adjusted to provide a most comfortable loudness percept with the volume control set at a setting of 6 or 7, which allows the user to adjust the volume upward or downward from the recommended setting established during programming. The Volume Adjustment parameter is usually set to 30 to 40%. For young children, comfortable upper-stimulation levels are set with the volume control at the maximum setting (e.g., 10) to prevent increases in the volume control and a potentially uncomfortable loudness percept. As previously mentioned, for experienced pediatric recipients, the Volume Adjustment parameter is typically set between 0 and 20%.

It is the clinician's responsibility to ensure that adults and older children understand the functions of the sensitivity and volume parameters to enable them to appropriately address difficulties they may experience across a variety of listening environments. The authors prefer to pro-

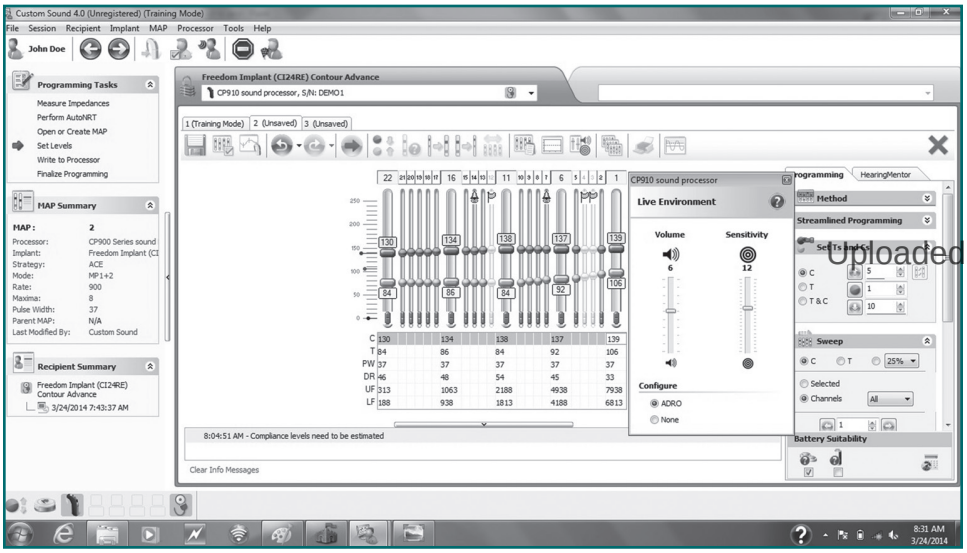
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vide older children and adult recipients with the ability to adjust the volume control, but we tend to disable the manual sensitivity control and enable ASC. However, the manual sensitivity control is occasionally enabled for recipients who encounter a wide range of listening environments and who request the freedom to exert more control over the manner in which their processor functions across these environments. It is critical that these recipients understand how to appropriately use the volume and sensitivity controls.

Once again, sensitivity influences the audibility of soft sounds and compression of loud sounds, whereas the volume control adjusts the stimulation the recipient receives for moderate to high-level inputs. Therefore, if patients feel that the overall loudness of a listening environment is too soft or loud, they should adjust the volume control up or down, respectively. However, if soft sounds from a distance are inaudible, especially in quiet situations, they can increase the sensitivity. In contrast, if background noise is preventing a user from understanding a talker located in close proximity, the recipient may choose to decrease the sensitivity setting. Understanding the function of these parameters will give experienced users the ability to address several common difficulties in quiet and noisy listening situations. It should be reiterated that when stimulation levels are set appropriately and ASC is used, there is essentially no need to adjust the sensitivity setting manually. With ASC, most users are best served by adjusting the volume control instead.

Also, in the Nucleus 5 and 6 sound processors, the sensitivity parameter affects the strength of the signal delivered from the sound processor microphone as well as the accessory or telecoil signal. For all earlier generation processors, the sensitivity adjustment affects only the strength of the signal from the microphone. Finally, it should be noted that the volume control and sensitivity settings may be adjusted in the "Set Levels" module to allow the recipient to experience the effect of these controls while listening to speech and environmental sounds during the programming process (Figure 5–16).

**Frequency Allocation.** As the name implies, the frequency allocation parameter determines how input acoustic frequencies are allocated or assigned across the active channels. At default settings, the frequency range allocated across all 22 channels spans from 188 to 7938 Hz. The bandwidth of the low-frequency channels, 14 to 22, are narrow (125 Hz) relative to the mid-frequency channels, 10 to 13, which have a bandwidth of




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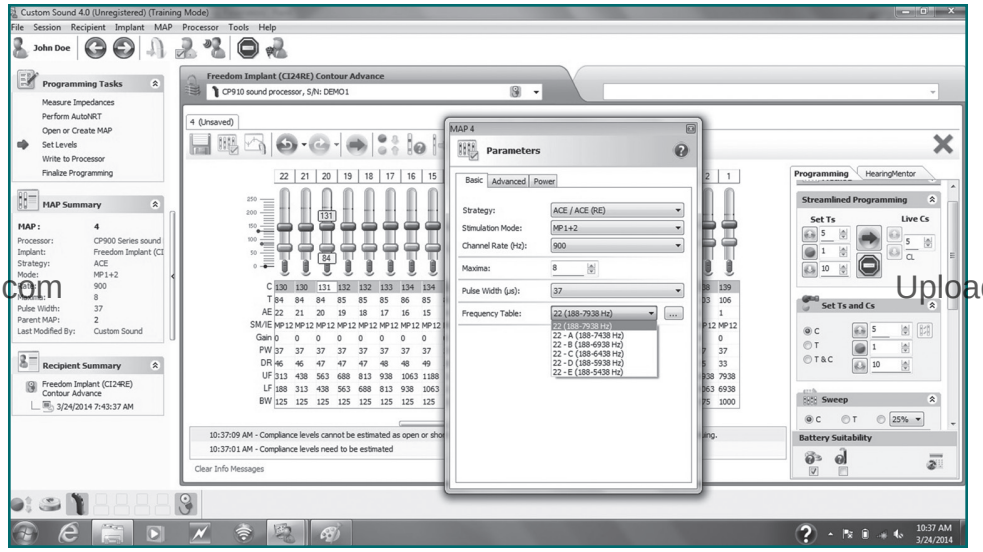
**FIGURE 5–16.** Volume control and sensitivity setting adjustments in the "Set Levels" module. *Note.* Provided courtesy of Cochlear™ Ltd.

250 Hz and high-frequency channels, 1 to 9, which have bandwidths ranging from 1000 to 375 Hz. The decision to provide narrower bandwidth channels in the low-frequency portion of the system is predicated upon the fact that the auditory filters of the peripheral auditory system are also narrower in the lower frequencies. Additionally, narrower channels are needed throughout the low and mid-frequency range in order to distinguish formant frequencies of periodic vowels and consonants, which is essential for recognition of these phonemes. It is worth noting that the width of low-frequency channels is linearly spaced, while the width of high-frequency channels is logarithmically spaced. This spacing is based off of psychoacoustic modeling of the normal auditory system (see Bark Scale in Zwicker, 1961).

The frequency allocation parameter may be adjusted within the "Frequency Table" drop-down menu (Figure 5–17). The default frequency allocation table is #22. As shown in Figure 5–17, the clinician may reduce the upper bandwidth of the allocation range by selecting from one of five alternative frequency allocation tables; however, this adjustment is not imperative. The authors of this book routinely leave the frequency allocation table at the default setting for pediatric recipients in light of research that suggests children need a wider bandwidth for the recognition and production of speech (Stelmachowicz et al., 2001). However, the authors do frequently adjust the frequency allocation table to #"22-B" for adult recipients, who are not as reliant on an extended bandwidth as compared to young children. Adjusting the frequency allocation table to #"22-B" does allow for a narrower bandwidth across several of the mid-frequency channels, which may improve the ability to distinguish speech sounds in that frequency range. Additional research of the frequency allocation parameter is needed to determine optimal settings for pediatric and adult recipients.

The clinician may also manually control the assignment of input acoustic frequencies to each individual channel by selecting the rectangle icon to the right of the "Frequency Table" drop-down menu (Figure 5–18). As shown in Figure 5–18, the "Frequency Bands" menu gives the clinician access to a table in which the low- and high-frequency cutoffs of each channel may be determined by manually entering the desired value (in Hz) within each respective field.

The manufacturer suggests that the manual control of frequency allocation be used with caution. Indeed, the recipient's performance may be




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**FIGURE 5–17.** Adjusting the frequency allocation parameter in the "Frequency Table" drop-down menu. *Note.* Provided courtesy of Cochlear™ Ltd.

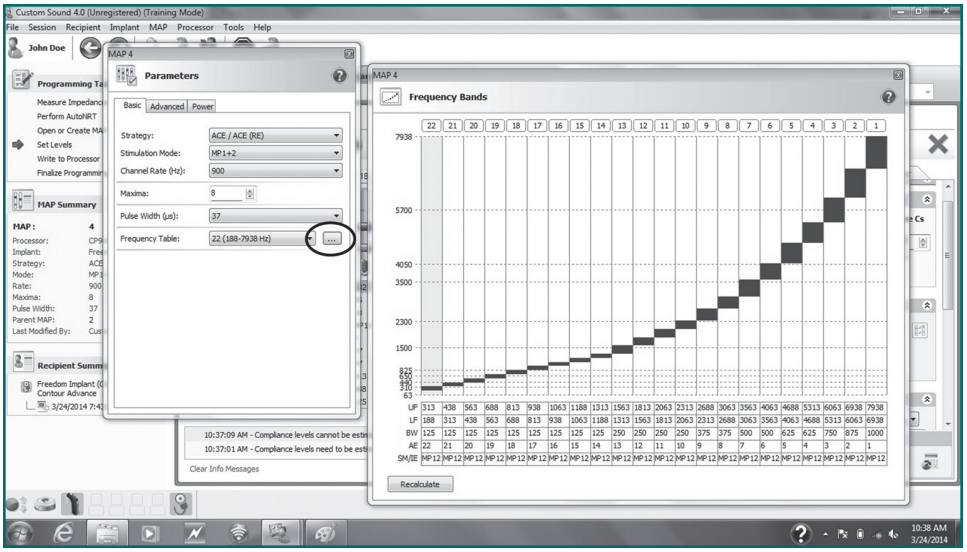



**FIGURE 5–18.** Select the rectangle icon (*circled*) to the right of the "Frequency Table" drop-down menu to manually control the input frequencies to each channel. *Note.* Provided courtesy of Cochlear™ Ltd.

compromised if adjustments are made in a haphazard manner. With that said, there are two common uses for this module. First, clinicians may use this feature to control the cutoff between acoustic and electric stimulation for recipients who have usable residual hearing and utilize electro-acoustic stimulation (discussed in Chapter 10). Second, some legacy sound processors (e.g., ESPrit 3G, Spectra, etc.) used frequency allocation that was quite different from the defaults used in current sound processors. When these recipients upgrade to contemporary processors, they often reject the frequency allocation tables used with the current defaults. In such a case, the clinician may manually re-create a frequency allocation that is similar to what the recipient used with his or her older sound processor. Again, additional research is needed to determine how clinicians may be able to adjust frequency allocation settings to optimize individual recipient performance.

**Loudness Growth.** The loudness growth parameter determines the steepness of the logarithmic function used to map input sounds into the recipient's electrical dynamic range. This parameter, which is also referred to as the Q-value, affects the slope of the amplitude growth function and determines the proportion of the user's EDR that is assigned to the upper 10 dB of the sound processor's input dynamic range. Lowering the loudness growth parameter (e.g., Q-value) results in a steeper loudness growth function slope, which should enhance the salience of low-level acoustic inputs (Figure 5–19). In other words, decreases and increases to loudness growth result in the mapping of inputs at higher or lower levels, respectively, in the user's electrical dynamic range. The loudness growth default setting is 20 but may be adjusted across a range from 10 to 50 within the "Advanced" tab of the "Parameters" menu (Figure 5–20).

The authors of this text rarely adjust the loudness growth setting. However, a theoretical example of when a clinician may choose to alter the loudness growth parameter would be if the recipient does not have adequate access to low-level sounds. In this case, the clinician may decrease the loudness growth parameter from 20 to 10 to provide more salience for soft sounds. However, it is worth noting again that it is very unlikely that the clinician will ever need to adjust this parameter. In general, access to low-level speech sounds and loudness normalization are satisfactory when stimulation levels are set appropriately.

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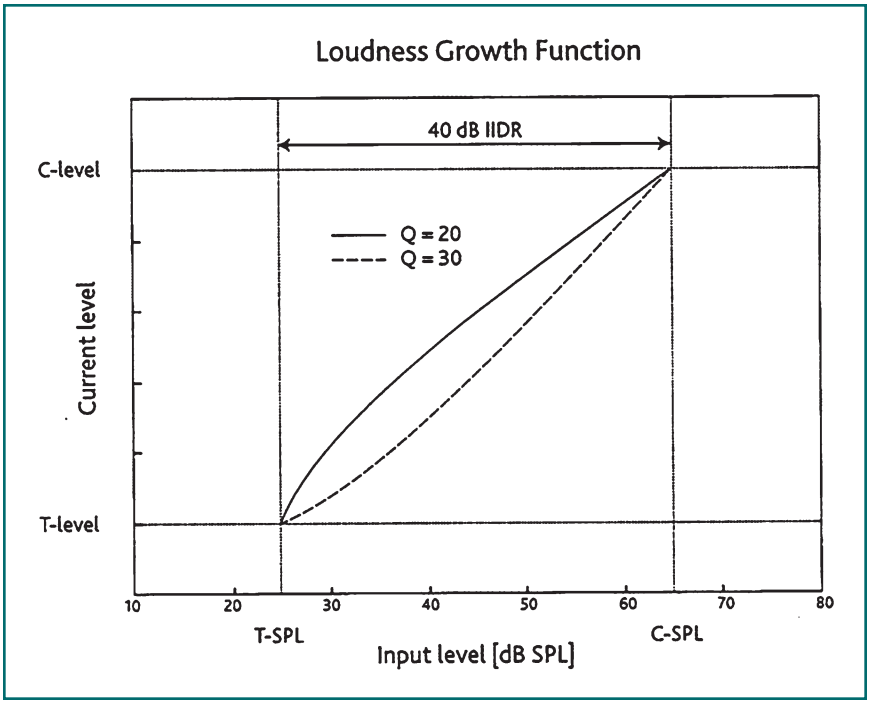
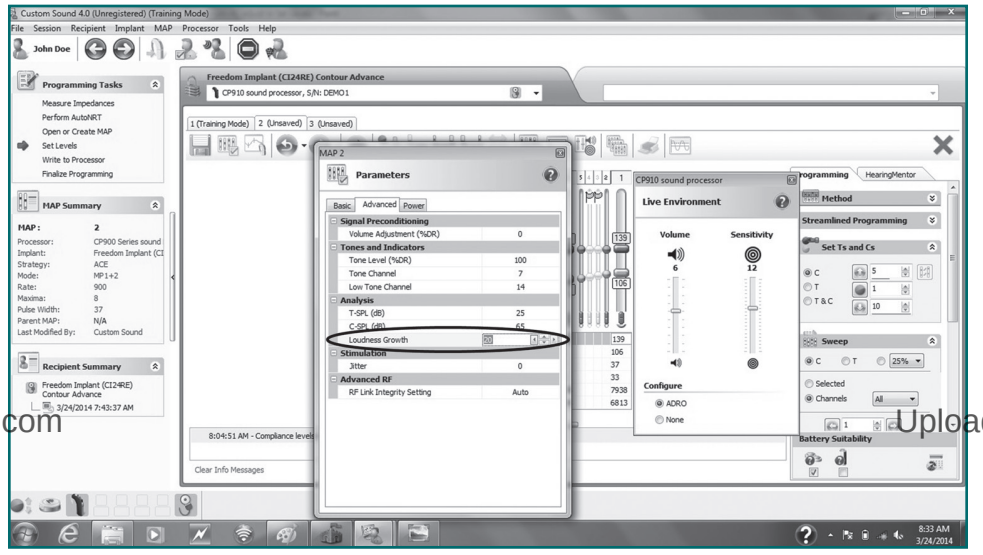



**FIGURE 5–19.** The loudness growth function is determined by the relationship between the input level and the current level. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 5–20.** Adjust the loudness growth function (*circled*) in the "Advanced" tab of the "Parameters" menu. *Note.* Provided courtesy of Cochlear™ Ltd.

**Jitter.** The jitter parameter is available only for the SPEAK signal coding strategy used by Nucleus 22 implant recipients. At low, fixed stimulation rates (e.g., 250 pps or lower), the recipient could perceive a continuous, low-pitched tonal sound that corresponds to the frequency of the stimulation

rate. In essence, the auditory nerve is responding in a phase-locked manner to multiples of the stimulus rate, which is a normal auditory system function for pitch coding. Jitter causes the stimulation rate to vary randomly by a certain percentage (e.g., 10%) around the stimulation rate. This random variation around the dominant stimulation rate eliminates the user's perception of a continuous low-pitched sound. The need to adjust the Jitter parameter is quite rare. When needed, the jitter parameter may be adjusted in the "Advanced" tab of the "Parameters" menu.

**Channel Gains.** Channel gains allow for frequency-specific adjustments of +10.5 dB/−12 dB to the input signal from the sound processor microphone. When using ACE, changes to channel gain will influence whether or not the channel will be selected as a maxima because the gain adjustment is implemented at the input before the signal is processed for coding. High-frequency acoustic information contributes greatly to speech clarity, so it is not desirable to reduce the gain for high-frequency channels because doing so would reduce the likelihood that the high-frequency channels would be selected for stimulation. As a result, adjustment of channel gains in *n*-of-*m* strategies, such as ACE, should be avoided. Adjustments to stimulation levels to address sound quality issues would be preferred. The effect of adjusting channel gains within the CIS signal coding strategy is not as deleterious, but again, it is usually unnecessary when stimulation levels are set appropriately. When needed, channel gains may be adjusted by selecting the "Channel Gain" icon on the icon bar of the "Set Levels" module (Figure 5–21) or by accessing the channel gain in the table below the stimulation levels in the "Set Levels" module (see Figure 5–21).

**Voltage Compliance.** Voltage compliance limits are measured in Custom Sound 4.1 by clicking on the icon circled in Figure 5–22. Voltage compliance levels, indicated by red dashes on each channel (limit on channel 22 shown with arrow on Figure 5–22), show the maximum current amplitude available for stimulation. Voltage compliance levels are limited by the voltage capacity of the processor's power source (i.e., battery). For the Nucleus 6, Nucleus 5, and Nucleus Freedom processors, Cochlear recommends that five or fewer channels are outside of voltage compliance limits by less than 10% on each channel. If these criteria are not met, the recipient will likely experience poor performance and sound quality because the loudness relationships across the channels will be

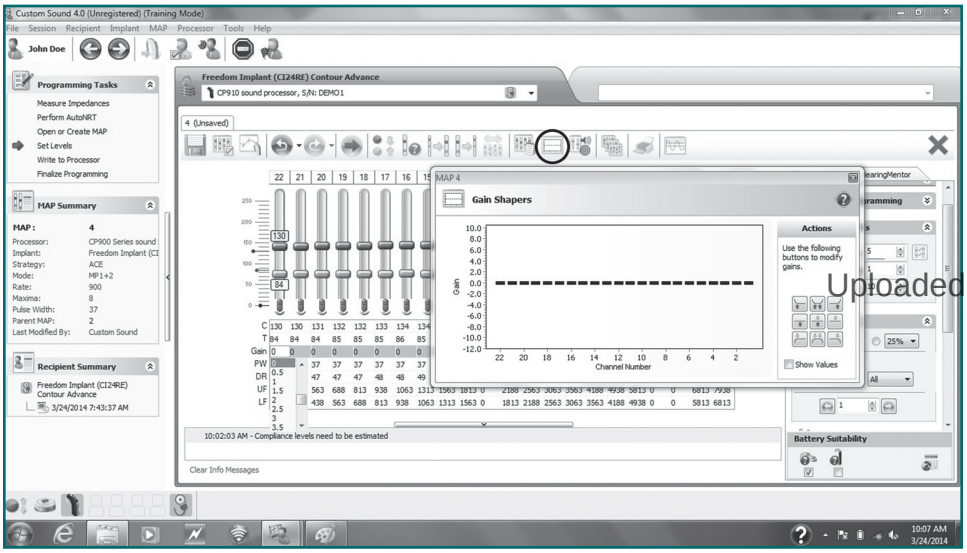



**FIGURE 5–21.** Adjust channel gains by selecting the "Channel Gain" icon (*circled*) in the "Set Levels" module or by accessing channel gain below if the stimulation levels within the "Set Levels" module. *Note.* Provided courtesy of Cochlear™ Ltd.

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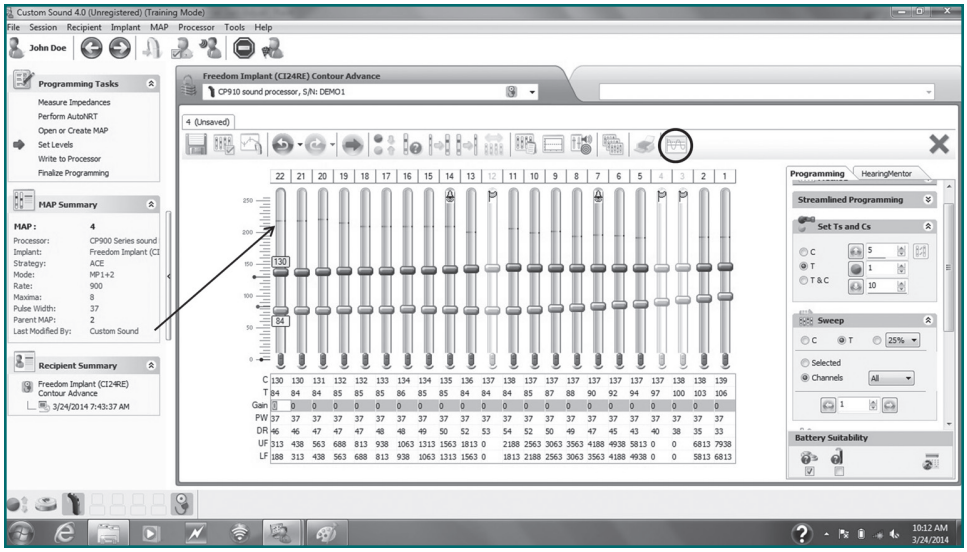



**FIGURE 5–22.** Access the voltage compliance (*arrow pointing to compliance*) by clicking on the circled icon. *Note.* Provided courtesy of Cochlear™ Ltd.

compromised. Also, use of maximum current levels increases the likelihood of channel interaction and decreases battery life.

Voltage compliance issues should be easy to address through programming adjustments. Specifically, the clinician should increase the pulse width of the compromised channel and remeasure T and C levels. The increase in pulse width should decrease the current required to elicit a desired loudness percept. Again, the authors of this text routinely increase the pulse width from the default of 25 usec to 37 usec, which reduces the likelihood that voltage compliance limits will be encountered while still allowing for a stimulation rate of 900 pps and 10 to 11 maxima. Voltage compliance problems were frequently encountered when using the Nucleus Freedom sound processor with the two zinc-air mini controller as the power source. However, the transmitting properties of the Nucleus 6 and Nucleus 5 sound processors are more efficient, and the programs of most all recipients can be supported by a two-zinc-air battery pack. Higher voltage compliance levels are present with the use of rechargeable batteries because they have a higher voltage capacity than the two 675 zinc-air modules.

**Prediction of Stimulation Levels from Other Programs.** In Custom Sound 4.1, the question mark icon (Figure 5–23) can be used to estimate stimulation levels based on measured T and C levels in another program. Once a program is complete, the clinician can begin a new program with the same signal coding strategy but a different stimulation rate or pulse width. Once the T and C levels are measured for one channel in the new program, the T and C levels of remaining channels can be predicted by clicking on the question mark icon. This feature allows for efficient creation of several programs with varying stimulation rates or a change in pulse width, which will aid the user in determining the optimal stimulus parameters for long-term use.

**Double Channel Mapping.** The Custom Sound 4.1 programming software also allows double channel mapping. This mapping involves directing the outputs of two channels into one intracochlear electrode contact, which may be desirable when the recipient has a restricted number of channels due to disabled electrodes. When patients have restricted channels, the range of input frequencies is limited (i.e., 5500 versus 7000 Hz). Double channel mapping results in an increased bandwidth assigned to each electrode contact and an increased overall bandwidth for the recipient. Double channel mapping is accomplished by right-clicking on the channel the clinician wishes to link and selecting "Double Channel" from the

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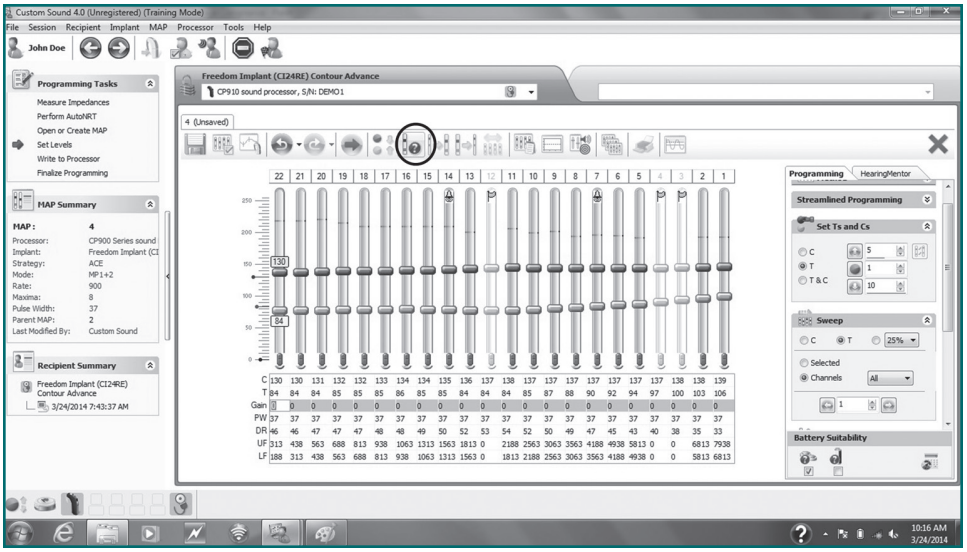
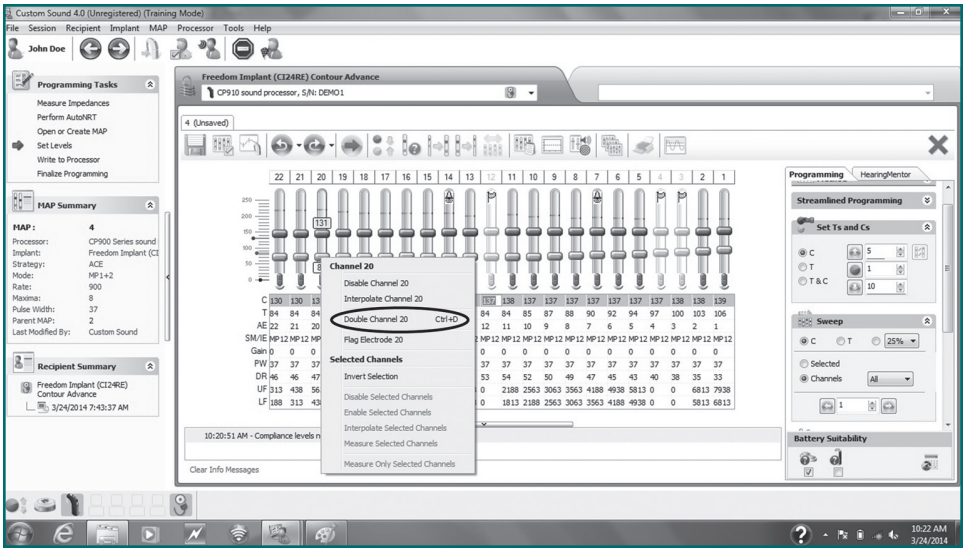



**FIGURE 5–23.** Click on the question mark icon (*circled*) to estimate stimulation levels based on measured T and C levels in another program. *Note.* Provided courtesy of Cochlear™ Ltd.




STUDENTS-HUB.com **FIGURE 5–24.** Right-click on the channel that you wish to select to select "double channel mapping" (*circled*). *Note.* Provided courtesy of Cochlear™ Ltd. Uploaded By: tuka abuzneid

list of options (Figure 5–24). Additionally, the clinician may choose the channels that are linked by selecting in the table (located below the stimulation level grid within the "Set Levels" module) the "active electrode" that corresponds to the desired channel. It should be noted that the active electrode ("AE") and reference electrode/coupling mode (SM/IE) are not present by default in the table of the "Set Levels" module. However, the clinician may enable these parameters to be included within the table by selecting "Tools" at the top of the Custom Sound software and then selecting the "My Preferences" tab. From there, the clinician must access the "Display" tab and select the parameters that he or she desires to be displayed in the table (Figure 5–25).

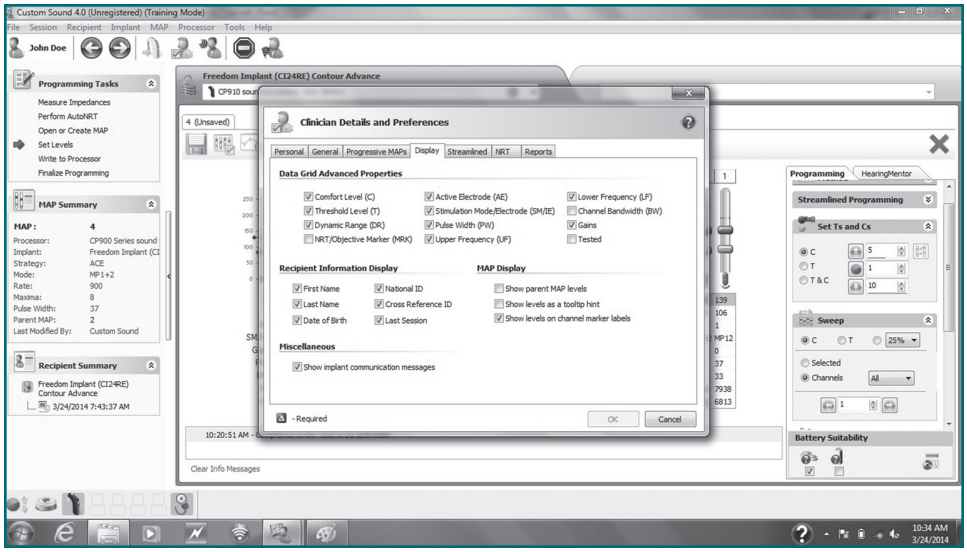



**FIGURE 5–25.** To enable the active electrode ("AE") and reference electrode/coupling mode (SM/IE) in the "Set Levels" module, select "Tools," "My Preferences," "Display," and the desired parameters to be displayed in the table. *Note.* Provided courtesy of Cochlear™ Ltd.

**Channel-to-Electrode Assignment.** At default settings, the Cochlear system allocates 22 channels to the 22 active intracochlear electrode contacts. Electrode 1 (and channel 1) is assigned to the most basal contact, and the electrode contact/channel number becomes progressively higher toward the apical end of the cochlea. The audiologist can reassign a channel to a different electrode contact by clicking on the active electrode coordinate within the parameters/data table in the "Set Levels" and selecting the desired allocation. This procedure may be warranted on two rare occasions. First, it may be beneficial for recipients who do not have the expected tonotopic organization as determined by sweeping the programming stimulus across the electrode array. In this case, alterations are made to the channel-to-electrode assignment with the intention of restoring tonotopic organization. Second, if the recipient has a common cavity cochlea, the surgeon may choose to insert the electrode array through a cochleostomy made near the horizontal semicircular canal (i.e., opposite the typical direction). This procedure requires the clinician to invert the channel-to-electrode assignment, which places the higher-numbered electrodes at the basal end of the cochlea and the lower-numbered electrodes at the apical end. It should be noted that the clinician may also activate an icon button (circled in Figure 5–26) that automatically reverses the order in which channels are assigned as a function of frequency (i.e., high-frequency acoustic inputs are assigned to the most apical electrode contacts, whereas low-frequency acoustic inputs are assigned to the most basal electrode contacts). The clinician would only activate this icon/button if the implant surgeon indicated that the electrode array was inserted in an inverted fashion.

**Additional Clinician-Adjustable Parameters.** The clinician may also make several other adjustments by selecting icons within the "Set Levels" module. Figure 5–27 identifies the icons which have not yet been discussed.

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## Write to Processor

Once a satisfactory MAP has been created in the programming platform, the clinician must create programs for the recipient and load the programs onto the recipient's sound processor. This is accomplished in the "Write to Processor" section of the Custom Sound software (Figure 5–28).

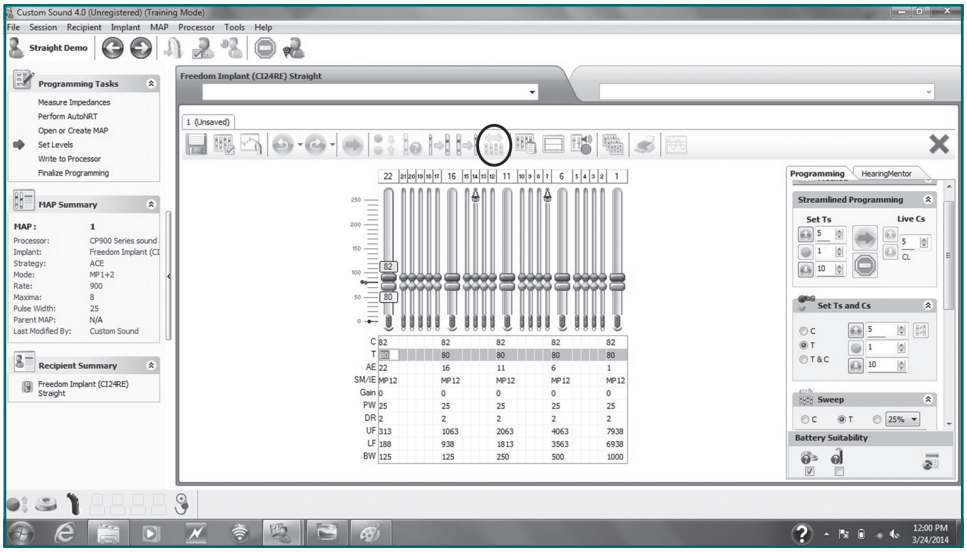


**FIGURE 5–26.** Icon for inverting the channel-to-electrode assignment (*circled*). *Note.* Provided courtesy of Cochlear™ Ltd.

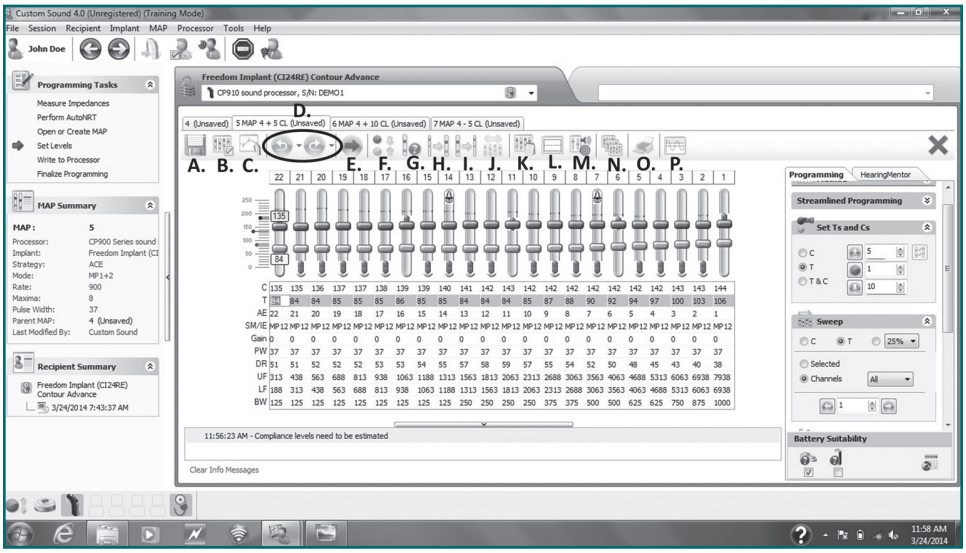



**FIGURE 5–27.** Other adjustments in the "Set Levels" module: **A.** Saves the current program; **B.** Additional MAP Parameters; **C.** Acoustic parameters for recipients using electroacoustic stimulation; **D.** Undo/Redo buttons—Used to correct mistake made during programming (i.e., revert back to settings prior to the most immediate programming change); **E.** Go Live—Activates sound processor microphone for live speech mode; **F.** Hug—Superimposes C Levels to the T Level value across all channels, **G.** Predict MAP Levels; **H.** Allows all channels to be measured, rather than just the five channels selected in the initial streamlined programming mode; **I.** Makes only selected (*highlighted*) channels measurable; **J.** Reverse Electrode Order; **K.** Notes—Allows clinician to save notes associated with a particular program; **L.** Gains; **M.** Configure Live Environment—Allows clinician to adjust volume and sensitivity controls for live speech programming mode and also allows for enabling/disabling of ADRO; **N.** Progressive MAPs—Creates three additional programs based on the current program, one program is 5 CL greater than the current program, another is 10 CL greater, and a third is 5 CL less than the current program. This button is intended to quickly create progressive programs for newly implanted recipients; **O.** Print—Prints current program parameters; **P.** Estimate Compliance Levels—Measures voltage compliance limits for each channel. *Note.* Provided courtesy of Cochlear™ Ltd.

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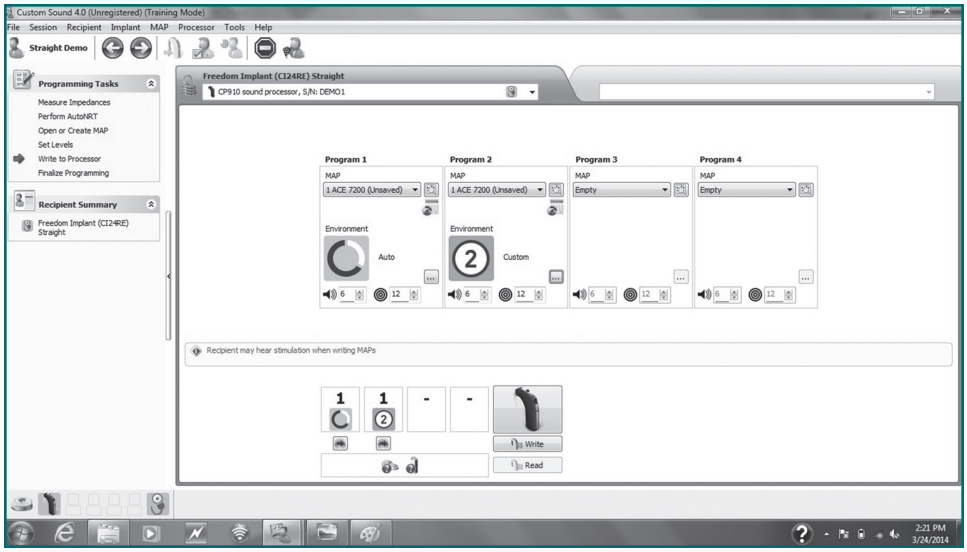



**FIGURE 5–28.** The "write to processor" screen. *Note.* Provided courtesy of Cochlear™ Ltd.

These programs may vary in stimulation levels or in preprocessing features. For example, new recipients may receive a different MAP in each program position so that they may have successively higher C levels to allow them acclimate gradually to higher levels of stimulation. Once the clinician believes that stimulation levels are appropriate and stable, the different programs will have identical stimulation levels (e.g., MAPs), but they may each possess different types of Smart Sound iQ preprocessing. Up to four programs may be loaded into a Nucleus 6 sound processor. The Nucleus 6 sound processor must be designated for use with either the right or left ear. In the default mode in the version of Custom Sound 4.1 that is used in the United States at the time of this writing, four user programs are populated from the MAP that was most recently created/selected, which include: (1) Everyday: Includes Autosensitivity Control (ASC) + Adaptive Dynamic Range Optimization (ADRO) input processing along with the conventional microphone mode; (2) Café: Includes ASC+ADRO+Zoom input processing; (3) Groups: Includes ASC+ADRO+Beam input processing; (4) Music: Includes ADRO+Whisper input processing. Also at the time of this writing, the Custom Sound 4 software platform will designate two default programs for clinics outside of the United States: (1) SCAN: Includes acoustic scene analysis with the automated provision of ASC, ADRO, Beam, Zoom, SNR-NR, Whisper, and WNR as indicated by the results of the scene analysis; (2) SCAN is disabled and the standard microphone mode is active. ASC, ADRO, SNR-NR, and WNR all active.

These programs are shown in Figure 5–28. Also, as shown in this same figure, the default volume control setting is "6" and the sensitivity setting is fixed at "12." It should be noted that the volume control setting will change from a setting of "6" if the clinician adjusted the volume setting in the "Set Levels" screen. The clinician may also adjust the volume and sensitivity settings in the "Write to Processor" module.

The clinician may review and adjust the type of input processing that is available within each program by selecting the small square icon toward the lower right-hand corner of each program box (Figure 5–29). These different types of input processing are described in detail below. The clinician may also change the name of each program except for the Auto (e.g., SCAN) program (which is fixed as "Auto") by accessing the drop-down menu in the "Remote Assistant Icon" section of the "Program Settings" menu (Figure 5–30). This allows for the provision of different icons/labels

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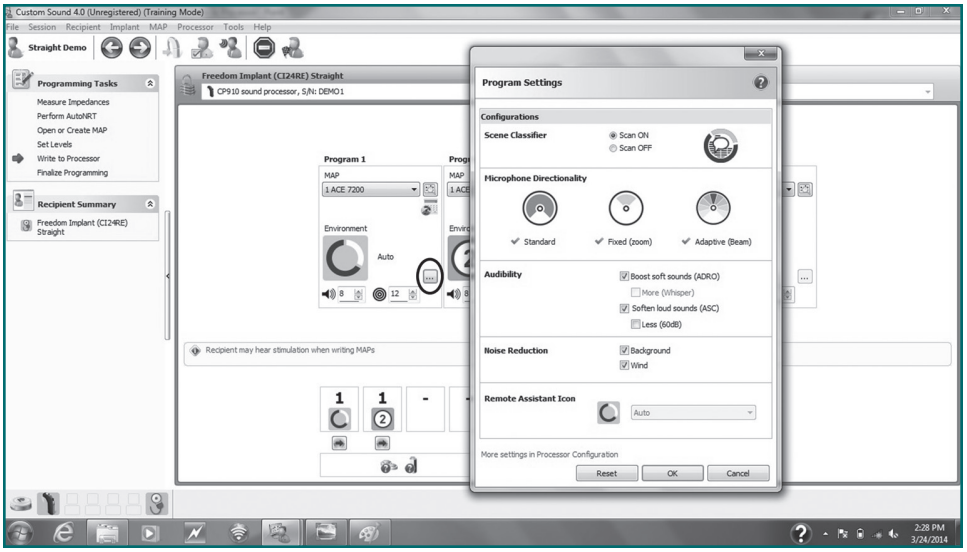
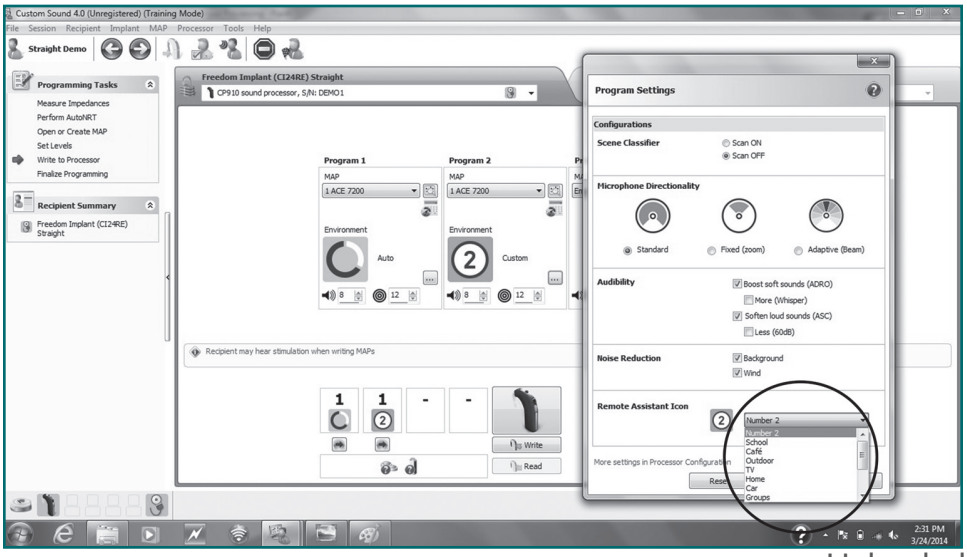



**FIGURE 5–29.** To review and adjust the input processing, select the small square icon (*circled*) toward the lower right-hand corner of each program box. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 5–30.** To change the name of each program, access the drop-down menu (*circled*) in the "Remote Assistant Icon" section of the "Program Settings" menu. *Note.* Provided courtesy of Cochlear™ Ltd.

for programs as they are displayed on the CR230 remote control. For instance, a program may be labeled numerically (e.g., Program 2) or descriptively (e.g., "Car," "Café," "Home"). The input processing schemes present in a program do not automatically change when a clinician changes the program descriptor. In other words, if the clinician changes the descriptor of the "Music" program, which at default includes the ADRO+Whisper input processing schemes, to be labeled as a "Car" program, it will be necessary to adjust the input processing as well. Because cars are noisy places,

it is probably most appropriate to disable Whisper (because Whisper can be detrimental to speech recognition in noise) and enable ASC and SNR-NR if it has received regulatory approval for commercial use in the clinician's home country.

### Input Processing

When selecting programs for the recipient in CustomSound 4.1, the clinician must consider the input preprocessing characteristics (i.e., SmartSound iQ processing), which include ASC, ADRO, Beam (multimicrophone directionality with adaptive positioning of the directional null), Zoom (fixed directionality), SNR-NR (channel-specific noise reduction), SCAN (fully automatic, adaptive directionality), and WNR (wind noise reduction) that will be provided to the recipient. The SCAN algorithm within SmartSound iQ is the cochlear implant industry's first input processing algorithm to perform acoustic scene analysis and automatically select the appropriate input processing with the goal of optimizing performance for the acoustic characteristics of a given listening situation. Based on previous research, optimal speech recognition for a variety of different speech sounds across a variety of listening environments can be achieved without the need for manual adjustment of processor controls by the combined use of ASC+ADRO in conjunction with SNR-NR, SCAN, and WNR (Wolfe et al., 2009, 2012, submitted). Access to low-level inputs is optimized by use of ADRO (James et al., 2002; Müller-Deile, Kiefer, Wyss, Nicolai, Battmer, 2008), and performance in noisy environments is optimized by use of SCAN, ASC+ADRO, and SNR-NR (Wolfe et al., 2009, 2012; Wolfe, Schafer, et al., 2013). Therefore, the authors of this text recommend simultaneous use of the following SmartSound iQ input processing schemes for adult recipients: ASC+ADRO, SNR-NR, SCAN, and WNR. The various input processing settings are selected in the "Write to Processor" screen where new programs are loaded into the sound processor (to be discussed later in this chapter).

These different input processing schemes (e.g., Smart Sound iQ input processing) are described in detail below. It should be noted that several of these input processing schemes (SCAN, SNR-NR, WNR) are not available for commercial use in the United States at the time of this writing. Nonetheless, all of the input processing schemes that may eventually be available with the Nucleus 6 sound processor will be described. Most likely, the SNR-NR and SCAN input processing schemes will be activated in most recipients' primary program (i.e., SNR-NR and SCAN will be a part of the "Everyday" program) once those preprocessing schemes are approved by the FDA. In fact, the primary default program in the version of Custom Sound used by the rest of the world includes SCAN with SNR-NR, WNR, and so forth.

### SmartSound iQ Preprocessing

As mentioned earlier in this chapter, the Nucleus system offers several Smart Sound preprocessing strategies that are designed to enhance listening abilities in a variety of situations. Also, the effectiveness of these preprocessing strategies is well supported by research (Gifford & Revit, 2010; James et al., 2002; Müller-Deile, Kiefer, Wyss, Nicolai, Battmer, 2008; Wolfe et al., 2009, 2012, 2014). As stated previously, all of the different Smart Sound iQ preprocessing options may be selected in the "Program Settings" menu, which may be accessed by clicking on the small square in the lower right-hand corner of each program box in the "Write to Processor" module (see Figure 5–29). As shown, two radio buttons exist to allow the clinician to enable or disable SCAN. If SCAN is disabled, the clinician may choose from one of three microphone modes including Standard, Zoom, or Beam. ADRO and ASC may be selected in the "Audibility" window of the "Program Settings" menu. Finally, SNR-NR and WNR may be selected by enabling "Background" and "Wind," respectively, in the "Noise Reduction" window of the "Program Settings" menu. Each of the aforementioned preprocessing strategies is described in detail below.

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**Autosensitivity Control and Autosensitivity Control Breakpoint.** As mentioned previously, ASC is a special type of input preprocessing exclusive to the Cochlear Ltd. SmartSound programming platform. ASC essentially functions as a slow-acting AGC in noisy conditions. When the

ambient noise level at the sound processor microphone exceeds 57 dB SPL, the ASC algorithm decreases the sensitivity of the sound processor microphone in an attempt to maximize the distance between the peaks of the speech signal and the ongoing competing noise.

As previously discussed, in quiet environments, ASC is not operational. However, ASC continuously analyzes the intensity of the competing noise during pauses in speech, and when the noise exceeds the default ASC breakpoint setting of 57 dB SPL, the digital ASC algorithm is engaged. The reduction in microphone sensitivity increases the input level that is subjected to high-level compression and prevents the peaks of speech from being compressed and embedded within the background noise. The reader is referred to Figure 5–15 for an illustration of how the sensitivity setting influences the positioning of IIDR window. Previous research suggests that use of ASC significantly improves speech recognition in noise relative to performance with fixed sensitivity settings (Wolfe et al., 2009, 2012).

The ASC breakpoint may be increased by 3 dB from a default of 57 dB SPL to 60 dB SPL if the recipient complains that too much attenuation is provided in noisy environments. The ASC Breakpoint is adjusted in the "Program Settings" menu which may be accessed by clicking on the small square (circled icon in Figure 5–31) in the lower right-hand corner of each program box in the Write to Processor module. The ASC Breakpoint may be adjusted to 60 dB by simply selecting the "Less (60dB)" option under the "Soften loud sounds (ASC)" option in the "Program Settings" menu. In our experience, the ASC breakpoint rarely needs to be adjusted for new recipients, but future research is needed to determine how it might be optimized for each individual. However, existing recipients who have not previously used ASC may report that ASC causes speech to be too soft in noisy environments. In this case, the Breakpoint should be increased to 60 dB SPL.

**Adaptive Dynamic Range Optimization (ADRO).** The objective of ADRO is to improve audibility for low-level sounds, maintain comfort for high-level inputs, and optimize the wide range of speech and other desirable acoustic inputs into the user's narrow electrical dynamic range. ADRO adaptively and constantly adjusts the gain in each of the channels of the sound processor based on an analysis conducted across three features of the input signal: the average input level, the level of the background noise, and the level of the loudest sounds. To maximize speech recognition, gains are increased in channels that possess a favorable SNR

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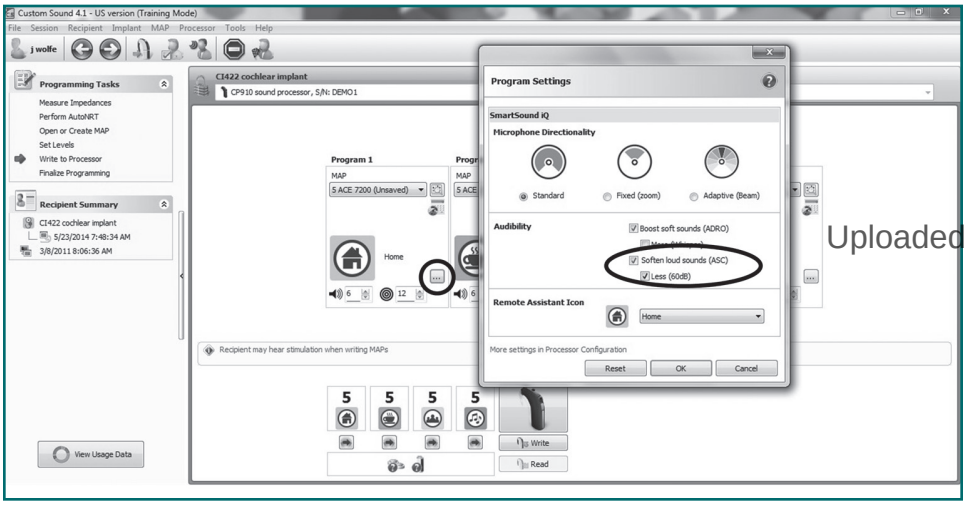



**FIGURE 5–31.** Adjust the ASC breakpoint in the "Program Settings" menu (*large circle*) by clicking on the small square (*small circle*). *Note.* Provided courtesy of Cochlear™ Ltd.

or include low-level speech. Conversely, to normalize loudness and prevent discomfort, gains are decreased in channels with high-intensity sounds.

Clinical studies evaluating speech recognition in noise with ADRO have found that it allows for an improvement in the recognition of low-level speech, but the benefits of ADRO for speech recognition in noise have been mixed. Dawson et al. (2004) studied speech recognition for a group of 15 children using ADRO with the Nucleus® SPrint™ body-worn sound processor. Speech recognition in quiet and in noise was significantly better with use of the ADRO program as compared to the children's standard program, which did not include preprocessing. Additionally, everyday sounds were comfortable with ADRO (Dawson et al., 2004). In contrast, James and colleagues (2002) evaluated speech recognition in quiet and in noise for a group of nine adults using ADRO with the SPrint sound processor and found that ADRO improved recognition of low-level speech in quiet relative to the standard program, but provided no significant improvement in speech recognition in noise. James et al. (2002) did report a significant improvement in perceptual sound quality with ADRO compared to the conventional program.

ADRO and ASC typically are used in conjunction with one another. ASC does not affect speech recognition in quiet as it automatically disengages in quiet settings, and it significantly improves speech recognition in noise (Wolfe et al., 2009). As a result, ASC should be used in conjunction with ADRO in order to optimize the recognition of speech in noise as well as the recognition of low-level speech in quiet.

**Beam.** Beam preprocessing, as used in the Focus program in the SmartSound iQ and SmartSound 2 (Nucleus 5) input processing schemes, utilizes the outputs of both processor microphones to create an adaptively varying directional microphone response. In noisy environments, the nulls of the Focus directional pattern are positioned to provide maximum attenuation for the locations in the rear hemisphere at which the most intense competing noise is arriving. However, this adaptive attenuation is disengaged when speech is the primary input present from the rear hemisphere (i.e., a positive SNR exists at the rear hemisphere). The focus mode significantly improves speech recognition in noise when speech is presented in front of the listener and the noise is from the side and back (Spriet et al., 2007). Focus should be used by adults in conjunction with ADRO and ASC, but it is most appropriate for use in noisy environments where the primary talker will be in close proximity (within 3 to 6 feet) in front of the user. In addition, the recipient must be counseled to face the talker and maintain a close proximity (i.e., 3–4 feet). Full-time use of Focus may limit access to speech and other important sounds that arrive from behind and from the sides of a recipient. Beam is available for use in the Nucleus 6, Nucleus 5, and Nucleus Freedom processors. It is implemented in the Nucleus 6 and Nucleus 5 sound processors by summing the outputs of two calibrated omnidirectional microphones. It should be noted that the Beam algorithm is implemented across all channels in order to provide a frequency-specific directional response (i.e., theoretically, the polar plots of two different channels may vary dependent on differences that exist in the signals present at those channels. The output of the single-cartridge directional microphone of the Freedom sound processor is combined with an omnidirectional microphone to achieve the adaptive directional processing of Focus.

**Zoom.** Zoom preprocessing utilizes the two omnidirectional microphones to create a fixed, hyper-cadioid polar plot pattern with a maximum null at 120 degrees (in the horizontal plane) when worn on the right ear or 240 degrees (in the horizontal plane) when worn on the left ear (see Figure 5–32 for an illustration of the polar plots associated with Beam, Zoom, and the Standard directional modes). Wolfe et al. (2012) reported that a large group of adult recipients received almost a 6 dB improvement in the SNR required for 50% sentence recognition with the Zoom preprocessing relative to their performance with the conventional microphone mode. Cochlear reports that Zoom provides optimal attenuation of diffuse noise. It should be noted that Zoom does not function adaptively, but instead, it implements a fixed directional response. As a result, this setting is unlikely to be appropriate for full-time use, because it is likely that the recipient will at least

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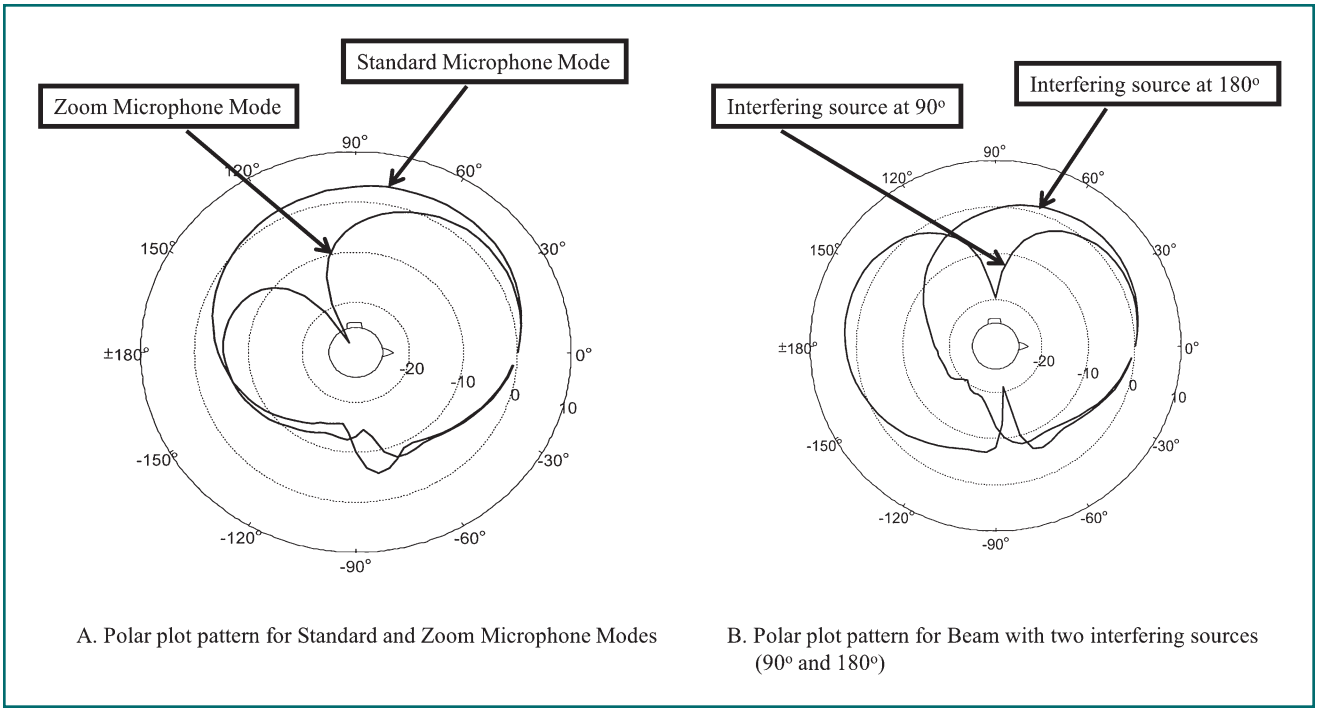



**FIGURE 5–32.** Illustration of the polar plots associated with Beam, Zoom, and the Standard directional modes. These responses were measured on a B&K HATS mannequin. *Note.* Provided courtesy of Cochlear Ltd.

occasionally need to hear sounds that arrive from the rear hemisphere.

The standard microphone mode of the Nucleus 6, Nucleus 5, and Nucleus Freedom sound processor is intended for use in quiet environments. It possesses a slightly positive directivity index (DI) as it seeks to mimic the directional response of the unaided ear Figure 5–32. The manufacturer's objective was to avoid the use of an omnidirectional response for full-time use, because the classic omnidirectional response when worn on the head is most sensitive to sounds arriving from the side of the user. As a result, the conventional omnidirectional response has a negative DI that may be detrimental for communication in face-to-face listening situations.

**SCAN.** SCAN preprocessing is new to the Nucleus 6 sound processor and is the first signal processing algorithm available in a commercial cochlear implant sound processor to perform scene analysis and automatically switch from the conventional microphone response to an adaptive beam-forming mode. The SCAN scene analysis attempts to determine whether the acoustic environment is primarily comprised of speech in quiet, speech in noise, noise, quiet (low-level or no sound present), wind, or music. Based on the results of the scene analysis, SCAN automatically selects the most appropriate preprocessing schemes for the given environment. If speech in noise is the primary input, then the processor automatically switches to the adaptive beam-forming mode. The activation threshold for activation to switch from the conventional microphone mode to a directional mode is 50 dB SPL. When noise is the only signal present, the sound processor will switch from the conventional microphone mode to the Zoom mode, which has a fixed, hypercardioid polar plot pattern. If speech and noise are both present, then SCAN will adaptively transition to beam-forming mode. At the time of this writing, there were no peer-reviewed publications reporting the potential benefits of SCAN. However, Wolfe et al. (2014) found an 18 percentage point improvement in speech recognition in noise with SCAN compared to performance without SCAN processing. Furthermore, word recognition in

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quiet was similar between the SCAN and conventional programs. Based on this study and others that are also submitted for publication (Mauger et al., 2014), the author's preliminary recommendation is to select SCAN for adult recipients and school-age children.

However, the authors are not as clear on the use of SCAN for infants and young children. There are essentially no published studies examining the potential benefits and limitations of adaptive input processing schemes for young children. Similarly, there are just a few published studies examining digital noise reduction and automatic, adaptive directionality for young children with hearing aids. In general, these studies have suggested that digital noise reduction in hearing aids does not degrade speech recognition in noise and may actually improve sound quality, comfort, and speech recognition, particularly when the competing noise is steady-state in nature (Pittman, 2011; Pittman & Hiipakka, 2013; Stelmachowicz et al., 2010). In contrast, the evidence for the use of directional amplification with children is mixed. Some researchers have suggested that the benefits of an automatic, adaptive directional system outweigh any limitations (Ching et al, 2009; Dillon et al., 2012; King, 2010). Specifically, Dillon (2012) noted that directional amplification may improve speech understanding in noise and pointed out that young children frequently orient toward the signal of interest. Indeed, Ching and colleagues (2009) evaluated young children's tendency to orient toward the signal of interest and estimated simulated directional benefit based on the children's orientation toward speech present in their environment. Ching et al. (2009) concluded that the net detriment of directional use in real-world settings is likely to be negligible at worst.

Additionally, Dillon noted that directional amplification is not highly directional in real-world use because the automatic gain control present in most hearing aids partially nullifies the attenuation provided to sounds arriving from the rear hemisphere, and reverberation present in most realistic environments limits the directivity of directional microphone. Finally, Dillon (2012) noted that modern automatic, adaptive hearing aids are unlikely to provide significant attenuation for rearward arriving sounds when the primary sound arriving from behind is speech. Consistent with the aforementioned information, the Australian Hearing's Protocol for Pediatric Amplification recommends providing automatic, adaptive directional hearing aids for infants and toddlers with hearing loss (King, 2010).

Although directional hearing aids can improve speech recognition in noise, there is an understandable reticence toward recommending directional technology with children. The primary concern stems from the potential for directional hearing aids to limit a child's access to important speech and environmental sounds arriving from behind or to the side of the child. Incidental listening refers to a listener's ability to attend to a message that is not directly intended to the listener. For instance, incidental listening occurs while a child is sitting in the corner of the room playing with a toy while also passively listening to a conversation his parents are having across the room. Estimates suggest that incidental listening accounts for 60 to 90% of what children learn about language (Ching et al., 2009; Cole & Flexer, 2007). If directional amplification limited access to incidental sounds arriving from behind a child, the child's model for language development may be compromised. Also, directional amplification may potentially hinder audibility for warning sounds arriving from behind (e.g., a car horn). Indeed, there are research studies that indicate that directional amplification can be detrimental to a child's speech understanding ability when the signal of interest arrives from behind the child (Ching et al., 2009; Ricketts, Galster, & Tharpe, 2007). Rickets and colleagues (2007) evaluated the speech recognition in noise of 26 children who ranged in age from 10 to 17 years old and used hearing aids capable of switching between omnidirectional and directional mode. The evaluation was completed in a classroom environment, and the noise was presented from four loudspeakers located in the corners of the classroom. Speech recognition in noise was assessed while the signal of interest arrived from the front of and also from behind the child. Compared to performance in the omnidirectional mode, children received a 2-dB improvement in the SNR with directional amplification when the signal of interest arrived from the front. However, when the signal of interest arrived

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from behind the children, the SNR required for 50% correct performance was 2-dB poorer in the directional mode. Ricketts et al (2007) concluded that children should only use directional amplification when the signal of interest is located in the front hemisphere and that hearing aids should be equipped with the ability to switch between directional and omnidirectional modes. Consistent with the concern that directional amplification may hinder access to important speech and environmental sounds arriving from behind a child, the Ontario Infant Hearing Program's protocol for the provision of amplification in children indicates that audiologists refrain from fitting directional amplification on infants and toddlers (Bagatto, Scollie, Hyde, & Seewald, 2010).

The authors of this text routinely provide automatic adaptive technology to school-aged children with hearing loss. These children are typically capable of describing their listening experience and are able to voice displeasure with limitations in the signal they receive. In contrast, infants and toddlers are typically unable to provide feedback pertaining to the signal they receive. As previously discussed, the language development of infants and toddlers is largely dependent upon their ability to access sounds that arrive from all directions. Furthermore, infants and toddlers often fail to orient toward the speech signals present in their environment. As a result, a directional processor may hinder their full-time access to speech arriving from all directions in their environments. For now, the authors of this text prefer to disable automatic adaptive directional programs (e.g., SCAN) for infants and toddlers and select the standard microphone mode for these recipients. Additional research is required to explore the benefits and limitations of real-world use of automatic adaptive directional systems for infants and toddlers with cochlear implants.

**SNR-NR.** Signal-to-Noise Ratio-Noise Reduction, or SNR-NR, is a digital preprocessing algorithm that analyzes the SNR of each channel and automatically reduces the gain of channels comprised primarily of noise. The primary goals of SNR-NR are to improve speech understanding and listening comfort in environments with steady-state noise, such as automobile or industrial noise. Dawson, Mauger, and Hersbach (2011) evaluated an early prototype of the SNR-NR algorithm with 13 experienced cochlear implant recipients and found about a 2-dB improvement in the SNR necessary for 50% sentence recognition compared to performance with conventional ACE signal coding. Mauger, Dawson, and Hersbach (2012) also evaluated a prototype of the SNR-NR algorithm, and found that SNR-NR provided a 27 percentage point improvement in sentence recognition the presence of steady-state competing noise and a 7 percentage point improvement in sentence recognition in the presence of babble noise relative to performance with conventional ACE signal coding. The aforementioned improvements are quite impressive for a single-microphone noise reduction technology. The SNR-NR feature is only available in the Nucleus 6 sound processor.

**WNR—Wind Noise Reduction.** Wind noise reduction (WNR) automatically switches to a polar plot pattern more closely resembling an omnidirectional response because use of a directional microphone in wind enhances the level of the wind noise. Additionally, WNR reduces the gain for channels that are dominated by wind noise. Although no peer-reviewed publications examining the WNR existed at the time of this writing, Cochlear has reported on an in-house study with 13 experienced adult recipients and found a 22 percentage point improvement in sentence recognition in the presence of wind noise with the use of WNR relative to the WNR-disabled condition.

**Whisper.** The goal of Whisper is to enhance audibility for low-level signals or sounds that originate from a distance. At default settings, the Nucleus 6 sound processor provides audibility for inputs down to 25 dB SPL. In addition, ADRO enhances access to these low-level sounds. However, Whisper may also be used to further enhance the salience of low-level sounds. The primary function of Whisper is to place low-to-moderate level sounds (e.g., ≤52 dB SPL) into a higher position in the EDR relative to typical settings. This may be helpful for stimuli with a wide range of inputs, such as music or the television, but it has been shown to potentially impair performance in noise (Cochlear, 2002; McDermott, Henshall, &

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McKay, 2002). Therefore, full-time use of Whisper is not recommended. Nevertheless, Whisper most likely should be enabled when listening to music, speech, or the television in quiet environments. Whisper and ADRO are enabled in the "Music" program within the Custom Sound software.

### Selecting Smart Sound iQ Options for Various Programs

The aforementioned Smart Sound iQ input processing schemes may be used simultaneously with one another. As stated previously, the clinician selects which Smart Sound iQ preprocessing schemes are available in each program slot in the "Write to Processor" section of Custom Sound. Once MAP levels are relatively stable, the authors typically provide one or two programs for adult recipients. Most adults receive an "Everyday" or Automatic (Auto) program and a program for specific situations (e.g., restaurants, work, car). For adults who are very eager to adjust the controls of their implant to optimize their listening abilities, the authors will provide up to four user programs. In fact, it may be worthwhile to allow adult recipients to use all four programs during the first few weeks of implant use to determine which are the most useful. Programs that are not deemed to be helpful may be removed.

At the time of this writing, SCAN was unavailable in the USA, so the first program typically possesses ASC+ADRO along with the conventional microphone mode and is labeled as the "Everyday" program. Many users are also provided with a program for background noise. The authors typically select the "Focus" program, which includes the adaptive Beam directional algorithm along with ASC+ADRO. Additional programs may be included as needed.

Once SCAN and SNR-NR are commercially available for use in the United States, then the first program will be labeled as "Auto" for Nucleus 6 and will include SCAN, ASC+ADRO, SNR-NR, and WNR. For many users, the Auto program will likely meet most, if not all, listening needs. However, additional programs may be offered as needed (e.g., the provision of a program with ADRO+Whisper only for listening to music or low-level speech in quiet).

The Nucleus 5 and Nucleus Freedom sound processors are not equipped with SCAN, SNR-NR, and WNR, and as a result, those processors will typically possess an "Everyday" program with ASC+ADRO, and for adult recipients, a noise program with Beam (or possibly Zoom in the case of the Nucleus 5) and ASC+ADRO. Additional programs may be added as needed.

Additionally, within the "Program Settings" menu, the clinician may access additional settings that affect how the sound processor is configured (Figure 5–33). The "Processor Configuration" menu may also be accessed by selecting the image of the processor in the lower, middle section of the "Write to Processor" module (see Figure 5–33). The top one-third of the "Processor Configuration" menu allows the clinician to set defaults for several parameters that are adjustable by the recipient using the CR230 remote assistant. For instance, the clinician may lock the buttons on the sound processor in order to prevent the recipient from inadvertently changing the program, telecoil, or volume settings. Also, the default telecoil and accessory-mixing ratios may be adjusted within this menu. The telecoil-mixing ratio determines the relative strength of signals from the telecoil and the sound processor microphone. The processor's built-in telecoil receives signals via electromagnetic induction and may be activated manually or in automatic mode for hearing on telecoil compatible telephones or in conjunction with induction loop hearing-assistance technology (e.g., FM system with neckloop).

In Custom Sound 4.1, the default Telecoil-Mixing ratio is 3:1 for adults and 1:1 for children, but the user may adjust the ratio with the CR210 and CR230 wireless remote assistants to allow for more or less emphasis from the sound processor microphone. A 3:1 mixing ratio provides 10 dB of attenuation to the signal from the processor microphone, whereas a 2:1 mixing ratio provides 6 dB of attenuation to the processor microphone. A 1:1 mixing ratio offers no attenuation to the signal from the sound processor microphone. However, for the Nucleus 6, Nucleus 5, and Nucleus Freedom sound processors, the T-SPL value is raised from 25 to 35 dB SPL. As a result, audibility for low-level sounds is reduced when the telecoil is activated.

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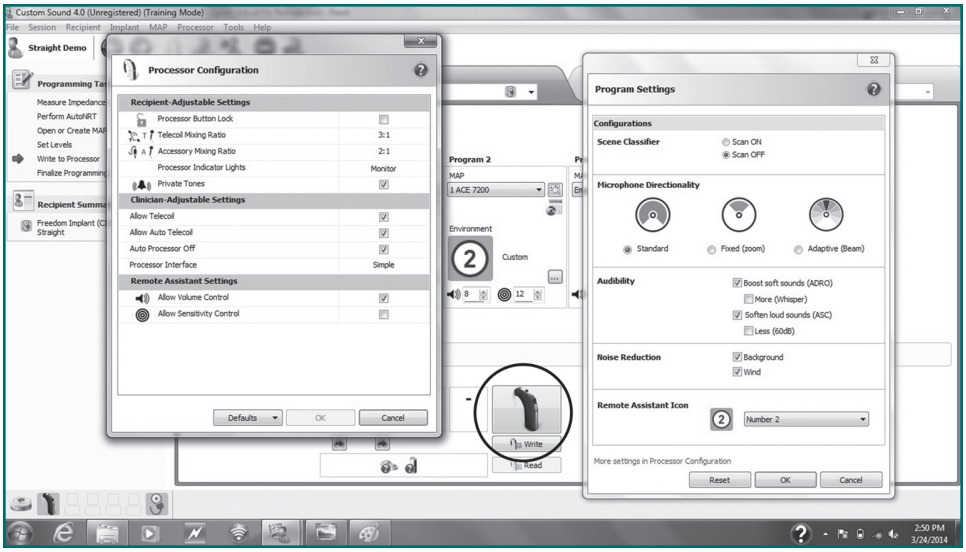



**FIGURE 5–33.** Additional settings that affect the processor configuration within the "Program Settings" menu. *Note.* Provided courtesy of Cochlear™ Ltd.

The accessory-mixing ratio functions exactly the same as the telecoil mixing ratio, but it influences the relative strength of signals from the sound processor microphone and the direct audio input source. Also, with the Freedom processor and all preceding processors, adjustments to the microphone sensitivity parameter through manual adjustments via ASC affect the strength of the signal from the sound processor microphone, but not the signal from the accessory source. However, with the Nucleus 5 and Nucleus 6 sound processors, an adjustment to the sensitivity setting affects the strength of the signal from both the sound processor microphone and the accessory source. As a result, the clinician must recommend a higher mixing ratio rather than a sensitivity adjustment to achieve an "accessory advantage."

The default accessory mixing ratio is 2:1 for adults and 1:1 for children (i.e., 6 dB reduction to processor microphone), but there are several other ratios that are discussed in Chapter 9. In addition, it should be noted that when an accessory source (e.g., FM receiver, MP3 player) is plugged into the Nucleus 5 and Nucleus Freedom processors, the floor of the IIDR is automatically attenuated by 10 dB (i.e., the T-SPL is raised from 25 to 35 dB SPL), even when set to 1:1. However, this elevation of the T-SPL does not occur when an accessory in use with the Nucleus 6 processor when set to a 1:1 mixing ratio.

It should be noted that both the telecoil- and accessory-mixing ratios may be adjusted by the recipient by using the CR230 remote assistant in the advanced mode. More information is provided on mixing ratios in Chapter 9, which covers hearing assistance technologies. For the Nucleus 6 and Nucleus 5 sound processors, the mixing ratio that is set in the "Processor Configuration" menu is applied to all programs within the processor, but the recipient may adjust the mixing ratio using the remote assistant. For Freedom processors, the mixing ratio is not a recipient-adjustable parameter, and it must be set by the clinician for each individual program. As a result, if the recipient desires different mixing ratios, then the clinician must provide multiple programs, each with a different mixing ratio.

Furthermore, the clinician may alter the operation of the sound processor's LED light, which indicates processor functionality or a fault condition. There are three choices: (1) Junior: Indicator lights are displayed (flickers green) for microphone and DAI inputs exceeding 25 dB SPL, and the indicator light is displayed (flashes yellow) to indicate a fault condition and when Auto Processor Off has powered down the implant;

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(2) Monitor: Indicator lights are displayed for auxiliary input when the sound processor is not on the head to indicate the presence of the auxiliary signal, and the indicator light flashes yellow to indicate a fault; and (3) Adult: Indicator light flashes yellow to indicate a fault when the processor is off the head. The private tone indicator may also be enabled/disabled within the "Processor Configuration" menu.

Furthermore, the clinician may alter several settings that cannot be altered by the recipient with the CR230 remote assistant. For example, the clinician may determine whether the telecoil is enabled/disabled. The authors of this book recommend disabling the telecoil for infants, young children, and all other recipients who do not possess the cognitive or motor abilities to manually activate the telecoil when appropriately needed. If the telecoil is enabled within this menu, the recipient may activate it by pressing the top button of the Nucleus 6 or Nucleus 5 sound processor, pressing the button on the left side of the CR230 remote assistant, or pressing the telecoil button of the CR210 remote control. As a result, the telecoil may be inadvertently activated by recipients who do not understand its use or do not have good manual dexterity.

The Auto Telecoil may also be enabled/disabled within this section. As mentioned previously, the Nucleus 6 and Nucleus 5 sound processors have an automatic telecoil that engages when a fluctuating electromagnetic signal with desirable speech-like characteristics is present from a nearby telephone or induction loop system. When the Auto Telecoil is enabled in the Custom Sound 4.1 software, the user may enable or disable the automatic telecoil feature with the CR230 remote assistant; however, when this feature is disabled in the software, it is unavailable to the user even via the remote control. In this case, the user may manually engage and disengage the telecoil with the CR210 or CR230 remote assistant and with a button located on the upper spine of the processor. The Auto Telecoil feature of the Nucleus 5 (CP810) sound processor may also be enabled/disabled with the CR110 remote assistant. The Freedom sound processor possesses a manual telecoil, but does not have Auto Telecoil capability; therefore, the indicators selected by the clinician are fixed for each program.

The "Auto Processor Off" parameter controls a feature that powers down the processor if a link is not achieved between the sound processor and recipient's implant over the course of two minutes. "Auto Processor Off" may be enabled/disabled within this menu. The authors of this book recommend enabling the feature for adults but disabling it for children.

The "Processor Interface" option allows the clinician to configure the sound processor in "Advanced" or "Simple" mode. In the simple mode, the user must access volume (to be discussed in the next section of this chapter) and/or sensitivity setting adjustments via the CR210 or CR230 wireless remote assistant. Additionally, in the Simple mode, a short push of the bottom button on the processor changes the recipient's program, whereas a long push of the bottom button powers down the processor. A short push of the top button enables/disables the telecoil.

In the Advanced mode, the recipient may adjust volume or sensitivity with the buttons on the processor by selecting the Advanced (volume) or Advanced (sensitivity) mode, respectively, in the Processor Configuration menu. It should be noted that the Advanced Mode is rarely provided to recipients as most recipients do very well when using the simple mode, and many recipients find the advanced mode to be somewhat complicated and cumbersome to use.

If the processor is programmed in "Advanced" mode, then the recipient may change either the volume or sensitivity, depending upon whether the clinician selected Advanced (volume) or Advanced (sensitivity) setting by a short press of the buttons on the processor. Furthermore, the recipient must provide a long press of the bottom button of the processor in order to change the program and a long press of the top button to activate the telecoil.

In the "Remote Assistant Settings" section of the "Processor Configuration" menu, the up/down buttons of the CR210 and CR230 remote assistants may be configured to operate as a volume or a sensitivity control. In most cases, the authors of this book recommend programming

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the up/down buttons of the remote assistant to function as a volume control with the sensitivity setting fixed at the default of "12." Recipients can typically satisfy the need for a change in loudness by adjusting the volume control, and research has shown that use of ASC is effective in automatically adjusting the sensitivity setting (Wolfe et al., 2009, 2012).

Finally, the clinician may choose to select from Adult or Pediatric default configuration settings (Figure 5–34). This option is especially important for use with infants and young children. The pediatric defaults implement 1:1 mixing ratios and enable the "junior" LED mode. Also, the Auto Telecoil and Auto Processor Off feature are disabled.

Once the desired programs are selected to be loaded onto the recipient's sound processor and the processor is configured as desired, the clinician simply selects the "Write" button to program the processor with the selected programs and settings. It is necessary for the processor and transmitting coil to be in place on the recipient's head so that the RF power level required to operate the processor may be measured and programmed into the processor. Once this measure is completed, the requisite RF power is saved in the recipient's file for that programming session and will be used when the same programs are loaded to processors at later dates (assuming the clinician opens to file for the session date at which the desired programs were saved to the sound processor). RF power level is explained in detail below.

### Power Level and Battery Life

The power level determines the strength of the RF signal necessary to deliver the requisite information from the external transmitting coil to the internal device without intermittencies or interruption. The power level is automatically estimated in Custom Sound 4.1 when new programs are loaded onto the sound processor (Figure 5–35). Power level is determined by several factors, such as stimulation levels and impedance, but the primary determinant is the thickness of the recipient's skin flap. The power level should be measured at every programming session. An inaccurate power level will result in signal intermittencies or an unnecessary waste of power with reduced batter life. The Custom Sound software measurement of the automatic power level estimate is sufficient for the vast majority of recipients. However, if the recipient complains that the signal "cuts out" in noisy environments, the clinician may want to set manually the RF power level to 10% higher than the estimated value.

A power level of 100% corresponds to the maximum voltage the processor can provide based on the power limits (i.e., voltage range) of the battery.

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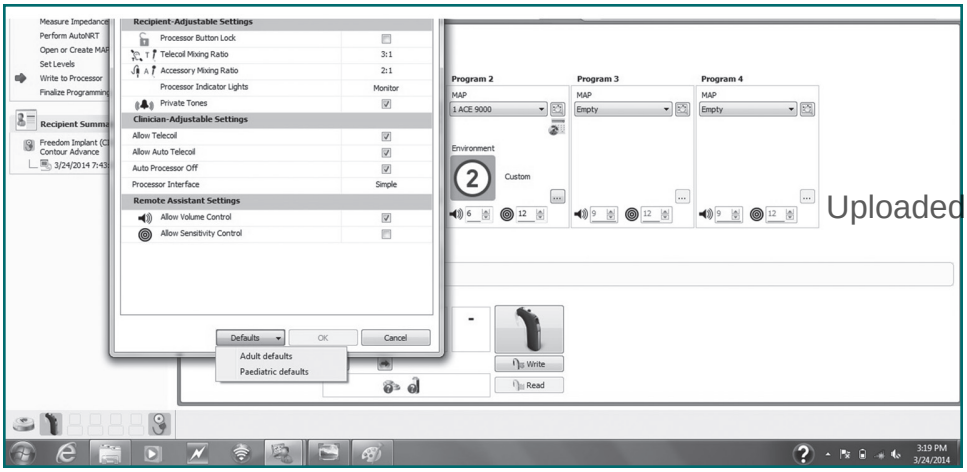



**FIGURE 5–34.** Selection of adult or pediatric default configuration settings. *Note.* Provided courtesy of Cochlear™ Ltd.

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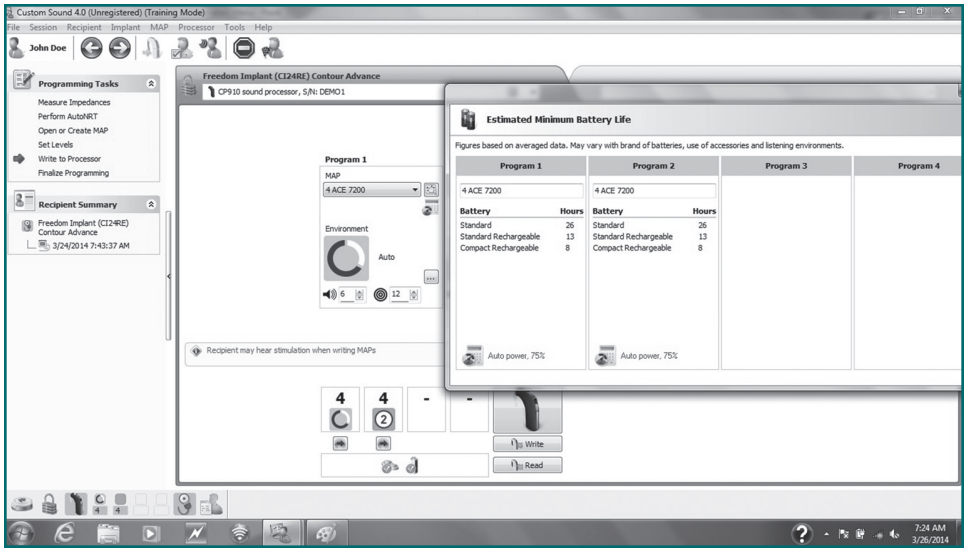



**FIGURE 5–35.** Estimation of power level. *Note.* Provided courtesy of Cochlear™ Ltd.

Power level is determined by the amount of voltage necessary to deliver the RF signal across the skin flap and maintain a consistent, intermittent-free signal at the electrode with the highest power requirements (i.e., highest C levels and electrode impedance) while the recipient communicates in a high-level noise environment. Each sound processor and battery option possesses a theoretical power level maximum limit. For instance, with the Freedom sound processor, a power level of 56% is the maximum RF power level that can be supported by the mini, two-zinc-air controller. The maximum RF power level supported by the three-zinc-air battery controller is 75%, and the body-worn, two-AAA-alkaline battery controller provides the highest power-level capacity of 100%. The Nucleus 6 and Nucleus 5 sound processors and cable/coils are more efficient than the Freedom sound processor and cable/coil, and as a result, power level is calculated differently. Because the maximum power level is determined on a processor-specific basis, it is not appropriate to compare power level required for one processor to the power level required for another model of processor.

With that said, the maximum power level supported by the two-battery module of the Nucleus 6 sound processor is 77%, whereas the rechargeable batteries provide a 100% power level. As a result, recipient with thick skin flaps and high power needs (i.e., high stimulation levels) may be more inclined to experience a consistently intact signal without intermittencies and perform better with rechargeable batteries compared to the performance with two #675 zinc-air batteries. For the Nucleus 5 sound processor, the maximum power level supported by the two zinc-air battery module is 75%, whereas the rechargeable batteries can support power levels up to 100%. It should be noted that the rechargeable batteries can generally support higher power levels without signal intermittency, but excessively high power levels are generally associated with reduced battery life.

Given the power constraints of the different processors, explicit instructions are provided as to which battery options are suitable for use by a recipient. In most cases, all battery options are suitable, but in some rare cases, there may be some battery configurations that may not provide sufficient power, particularly toward the end of the battery life. A good example of such a case would be the use of a two zinc-air for a Nucleus 6 recipient with a power level exceeding 77%. Because children may not be able to report potential sound quality issues, they should use battery options that will guarantee suitable power.

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It is important to note that it is occasionally difficult to accurately estimate power level for recipients with very thin skin flaps. In some cases, the telemetry may be interrupted when recipients with thin skin flaps operate their processors in daily use. In this event, the recipient may complain of signal intermittencies and/or the LED light and datalogging may indicate that a consistent lock is not obtained. If these issues occur, the clinician may adhere a layer of mole skin to the underneath side of the coil or use a small plastic ring (spacer) provided by Cochlear to attach to the underneath side of the coil. Either solution artificially increases the width of the skin flap and allows for accurate provision of power level.

On the opposite end of the spectrum, it may be difficult to provide sufficient power to maintain a continuous lock and telemetry for recipients with unusually thick skin flaps. For recipients with thick skin flaps, it may be necessary to shave the hair over the skin flap to facilitate a connection. Typically, the pressure from the coil on the skin flap will result in "thinning" of the skin flap over time. Thick skin flaps are typically associated with high power levels, which will result in lower battery life. Also, it is more favorable to use rechargeable batteries, which possess a higher power level capacity than two zinc-air batteries. Finally, for rare circumstances in which the aforementioned solutions have been attempted and the recipient continues to experience signal intermittencies secondary to a thick skin flap, it may be necessary to try the Freedom sound processor, which possesses a larger transmitting coil capable of generating a higher RF level. The Freedom may also be used with a body-worn battery controller to provide a higher voltage capacity.

Finally, the RF power level may be changed to a manual mode in the "Parameters" menu ("Power" tab) of the "Set Levels" screen (Figure 5–36). This adjustment is rarely needed. If a recipient complains of intermittencies, the clinician can adjust the RF power level to an alternative setting in an attempt to eliminate the intermittency.

After power level is measured, the Custom Sound software also provides an estimate of the life (in hours) that a recipient should expect to receive from different battery configurations (Figure 5–37). Battery life is influenced by skin flap thickness, stimulation levels, and electrode impedances. As one might expect, lower battery life is achieved for recipients with thicker skin flaps and higher stimulation levels and electrode impedances. It should be noted that the battery life measure is a general estimate of the life one should expect from a particular type of battery. Battery

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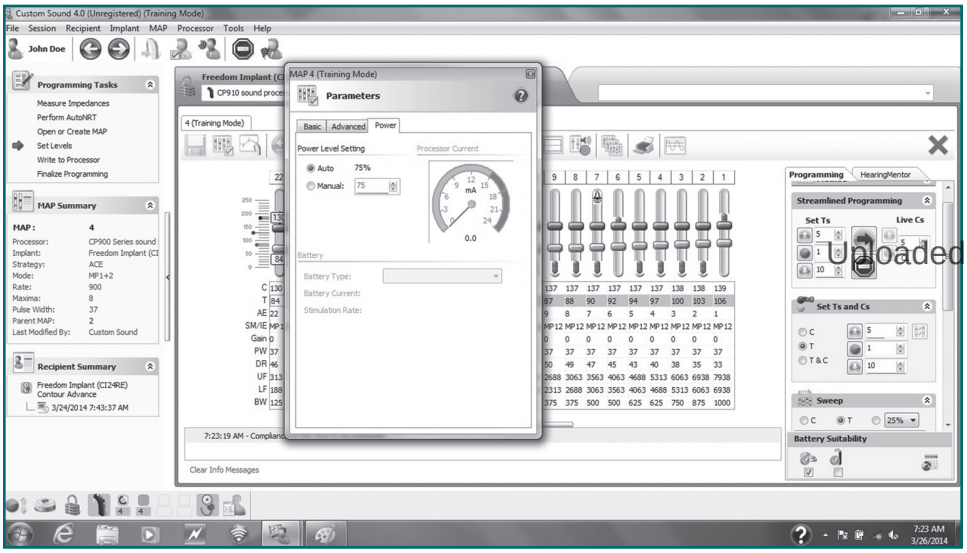



**FIGURE 5–36.** The RF power level may be changed to a manual mode in the "Parameters" menu. *Note.* Provided courtesy of Cochlear™ Ltd.

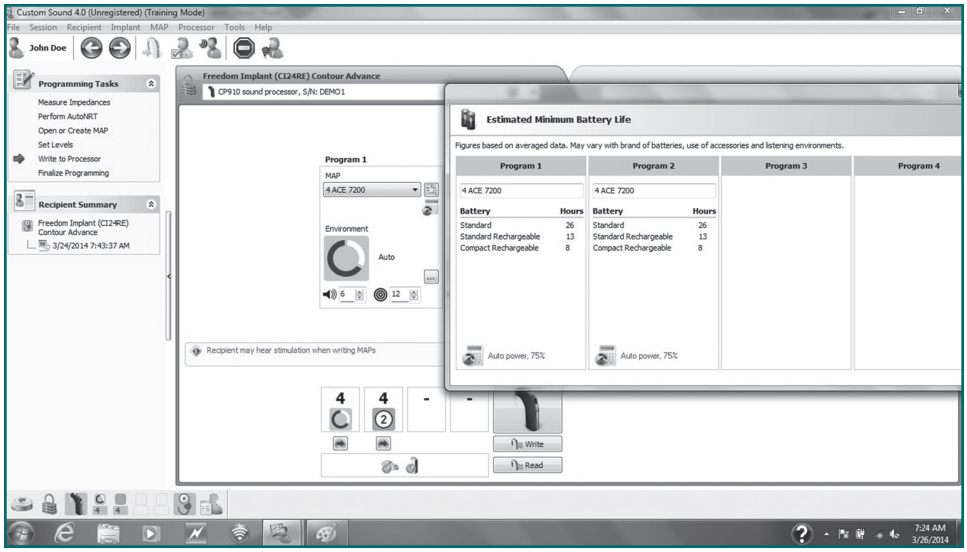



**FIGURE 5–37.** Estimated battery life. *Note.* Provided courtesy of Cochlear™ Ltd.

life will be shorter if the recipient uses a personal digital RF/FM receiver or wireless streaming (i.e., GN ReSound wireless accessories). Battery life will also be shorter if the recipient continually communicates in high-level noise environments.

### Bilateral Mapping

The Custom Sound 4.1 platform allows for simultaneous mapping of recipients with bilateral cochlear implants via two programming pod interfaces connected to the computer. The "Bilateral Balancing" tab allows for the activation of the bilateral sound processors in live speech mode (Figure 5–38). Global adjustments to T and C levels will ensure loudness balance between the two ears for speech and environmental sounds. Most users require a slight individualized reduction in C levels when listening in the bilateral mode compared to the unilateral conditions, which is likely related to binaural summation. The frequency allocation table for each processor may also be adjusted within the "Bilateral Balance" platform.

### Neural Response Telemetry Measurement and AutoNRT

Neural Response Telemetry (NRT) is the feature within Custom Sound ("Perform AutoNRT") that allows the clinician to measure the recipient's electrically evoked compound action potential (eCAP). The AutoNRT platform allows for the automatic measurement and estimation of NRT thresholds. A comprehensive discussion of the eCAP response and NRT in particular is beyond the scope of this book. The clinician is directed to Michelle Hughes' excellent book entitled *Objective Measures in Cochlear Implants* for a thorough description on NRT.

The default AutoNRT settings call for measurement of the NRT response on five electrodes (22, 16, 11, 6, and 1). The clinician may choose to measure the NRT response on all electrodes by selecting the "All" option under the "Number of Electrodes" that are tested (Figure 5–39). The clinician may also choose the starting level and the step size (both in Current Level—CL) used for assessment. When the NRT response cannot be successfully measured in the AutoNRT platform, then the clinician can attempt to adjust measurement parameters in the Custom Sound EP application, which is a separate program that provides comprehensive control of stimulation parameters pertaining to objective measures. It should be noted that it is definitely the rare exception when the eCAP cannot be recorded from an experienced recipient when using in the AutoNRT platform. Finally, the reader is referred

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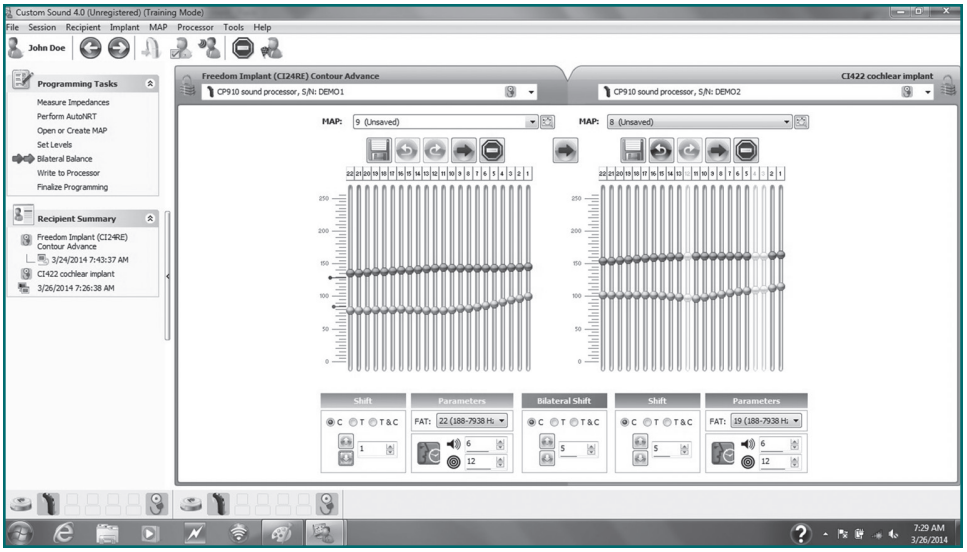
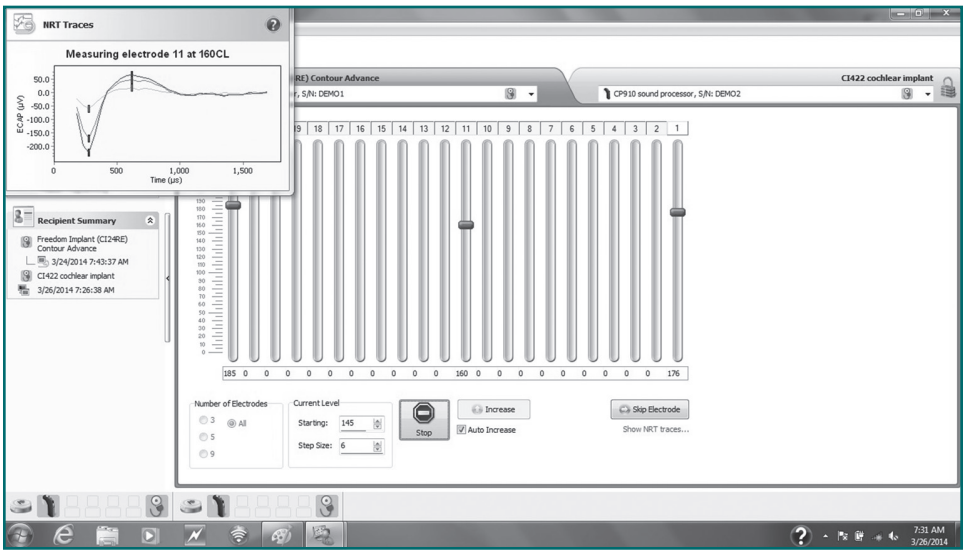



**FIGURE 5–38.** "Bilateral Balancing" tab for activating bilateral sound processors in live speech mode. *Note.* Provided courtesy of Cochlear™ Ltd.




STUDENTS-HUB.com **FIGURE 5–39.** Neural response telemetry (NRT) measurement. *Note.* Provided courtesy of Cochlear™ Ltd. Uploaded By: tuka abuzneid

to Chapter 7 for additional discussion on the use of the eCAP response in programming of cochlear implants.

### *View Usage Data*

The View usage Data platform allows the clinician to review a datalogging report, which provides comprehensive information pertaining to the total use and usage patterns for the recipient since the previous programming session. The clinician may review the number of hours the processor is used per day, the number of hours in which link is obtained/lost, the frequency in which each programs are used, and acoustic characteristics of the environments the user encounters (Figure 5–40).

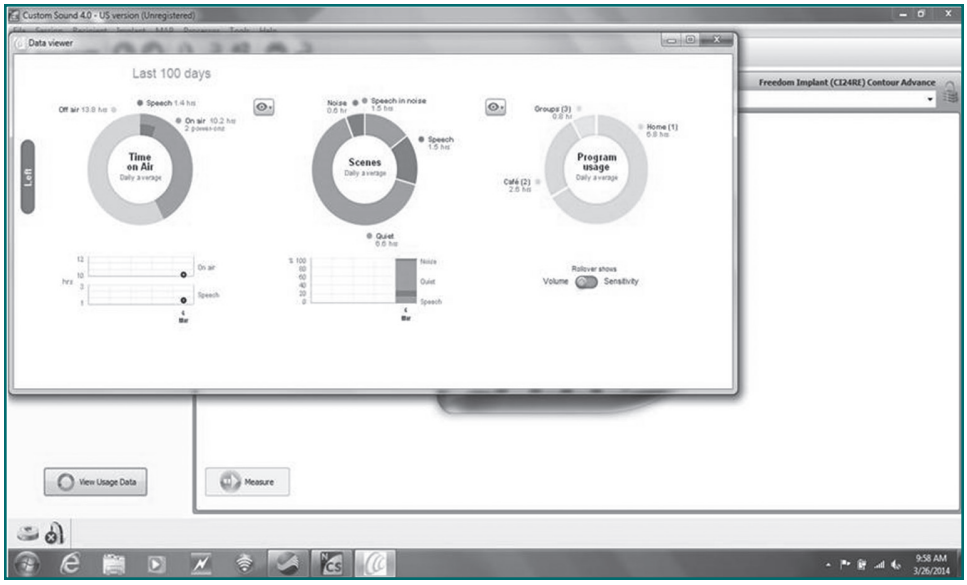



**FIGURE 5–40.** Access the view usage data platform to review a datalogging report. *Note.* Provided courtesy of Cochlear™ Ltd.

This datalogging features is the first of its kind in the cochlear implant industry.

### *Finalize Programming*

The "Finalize Programming" platform is simply a mechanism for the clinician to record notes for a programming session and to generate both a clinic report (with MAP/Program parameters) as well as a report for the recipient to take home.

## PROGRAMMING PREVIOUS GENERATIONS OF NUCLEUS IMPLANTS

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Cochlear Ltd. has made a conscientious effort to offer the latest sound processor technology to its earliest recipients (Nucleus 22 users). However, due to limitations of the signal processor within the Nucleus 22 implant, as of now, it is only compatible for use with the Nucleus Freedom sound processor and its predecessors. There are plans to make the Nucleus 6 sound processor available for Nucleus 22 recipients by the end of 2015.

In addition, the same programming software can be used across all generations of implants, which greatly simplifies programming for clinicians. The basic principles described in the previous section will enable a clinician to program most Nucleus implants with a few exceptions.

Programming of the Nucleus 5 sound processor is essentially identical to the process of programming the Nucleus 6 sound processor described previously. The primary difference is that the Nucleus 5 sound processor is not capable of using SCAN, SNR-NR, and WNR input processing. As a result, the default "Everyday" program utilizes ASC+ADRO. The "Noise" program features Zoom+ASC+ADRO, while the "Focus" program features Beam+ASC+ADRO. Finally, the "Music" program uses ADRO+Whisper. Additionally, the Nucleus 5 sound processor does not possess datalogging, so that feature is obviously unavailable in Custom Sound. Finally, the wireless streaming accessories are unavailable with the Nucleus 5 sound processor.

The aforementioned programming differences between the Nucleus 5 and 6 sound processors also apply to the Freedom sound processor. Also, Zoom is unavailable in the Freedom sound processor as is the Auto telecoil. Additionally, the

Freedom sound processor does not have a remote assistant, so the telecoil- and accessory-mixing ratios must be made to individual programs that are loaded onto the Freedom sound processor.

There are also a handful of differences pertaining to the programming of Nucleus 22 implants as well. First, Nucleus 22 internal devices must use the SPEAK signal coding strategy, which typically uses bipolar (BP) electrode coupling (although it rarely occurs, the clinician may also select common ground coupling) (Chapter 2). The current default electrode coupling mode is BP+3 where the reference electrode is four electrodes apical to the stimulating electrode. However, when the SPEAK was first introduced, the default mode was BP+1, and many users continue to use that today. As a result, with BP+3 coupling, the four most apical electrodes are disabled in the user's program. If adequate loudness growth cannot be achieved with this configuration, a wider bipolar coupling mode (e.g., BP+4) may be used. Finally, a "variable" electrode coupling mode in SPEAK allows the clinician to vary the bipolar coupling mode (i.e., BP+1, BP+2, BP+3) from one electrode to the next. In the variable mode, the clinician should ensure that the expected tonotopic organization exists by sweeping at C level from apex to base.

Second, the Nucleus 22 device is not capable of making telemetry measures (e.g., electrode impedance assessment, NRT, etc.). As a result, the RF power required to provide a signal with consistent integrity across a variety of environments will need to be measured. Also, the clinician should perform a Skin Flap measure to estimate its thickness (i.e., density). This information is used to estimate and optimize the RF power used in the recipient's program. For the ESPrit 3G device, the clinician also must perform a Power Optimization Test to estimate the necessary RF power level.

Finally, stimulation levels in the Nucleus 22 device may be measured in Stimulus Level units or Current Level units, the same scale used for contemporary devices. When using Current Level units, the pulse width is set at a fixed value. However, with Stimulus Level units, the pulse amplitude and the pulse width change in a complex manner to in attempt to minimize power and also to minimize time by providing the shortest possible pulse width. As a result, Stimulus Level units are preferred, but Current Level units may be used if manual determination of the pulse width is potentially advantageous.

## KEY CONCEPTS

- The provision of appropriate C levels is one of the most important requirements for creating an optimal program for Cochlear recipients.
- T levels should be measured for Nucleus recipients and should be set to the minimal stimulation level that is consistently audible to the user.
- Analysis of electrode impedance results and NRT thresholds provides the clinician with valuable information pertaining the the function of the Nucleus cochlear implant system as well as the physiology of the recipient.
- Use of advanced signal processing features, such as Autosensitivity Control, ADRO, Scan, dual-microphone directional technology (e.g., Beam), SNR-NR, WNR, and Whisper are likely to improve recipient performance in difficult listening situations. ASC and ADRO should be considered for use in most listening situations.
- Use of the Nucleus wireless accessories, such as the Unite Mic, the Phone Clip+, and the TV Accessory will likely optimize user performance in several challenging situations.

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# 6

# Programming MED-EL Cochlear Implants

*Jace Wolfe, Erin C. Schafer, and Sara Neumann*

The primary programming platform for current MED-EL sound processors is the MAESTRO software platform, which is used for the more recent OPUS 1, OPUS 2, and RONDO sound processors. Recipients with the COMBI 40+, PULSAR CI$^{100}$, SONATA TI$^{100}$ and CONCERT internal devices can all be programmed in the MAESTRO software as long as they use the OPUS 1, OPUS 2, or RONDO sound processors.

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## PATIENT INFORMATION ENTRY

Within the MAESTRO software, along the left hand side of the screen, there are multiple tabs. This is referred to as the task column. The clinician must select the first tab to enter or review patient information. If the patient is an established user who has transferred from another clinic, the clinician must simply hook up the patient's sound processor to the Diagnostic Interface Box (DIB) II box and enter the appropriate software (MAESTRO for OPUS and RONDO processors and CI.STUDIO+ for TEMPO+ and CISPRO+ processors). A pop-up will automatically appear and auto-populate the patient's information into the programming software platform.

For a new patient, the clinician must enter the standard information including name, date of birth, as well as gender, and then she or he must select the tab with the implanted ear (Figure 6–1). The notifications pane in the lower third of the screen will prompt you to input requisite information and those messages will disappear as the required information is entered. Next, there are several pieces of information that must be entered regarding the patient's internal and external devices. See Figure 6–2 for the relevant information needed to start programming the MED-EL cochlear implant sound processor.

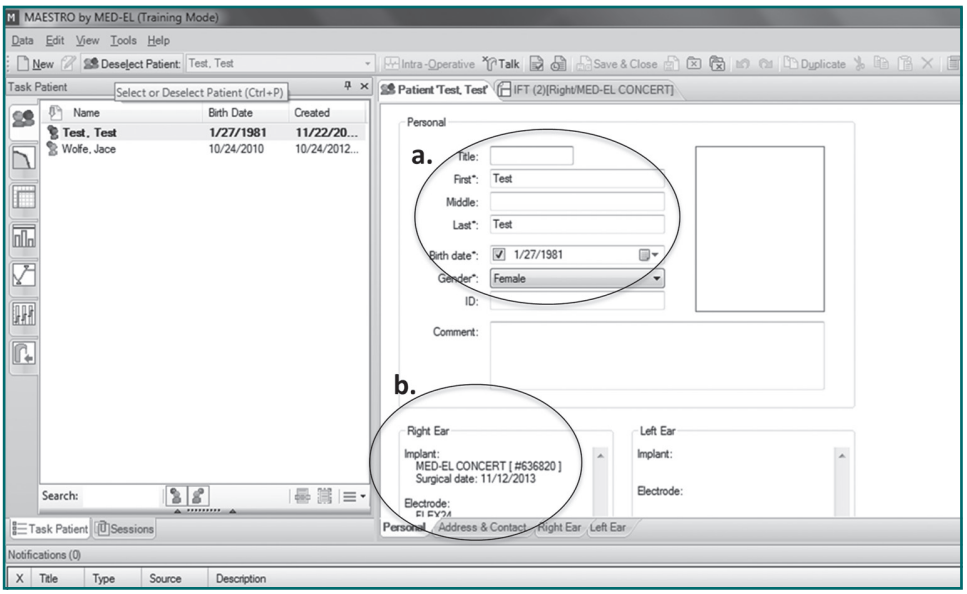
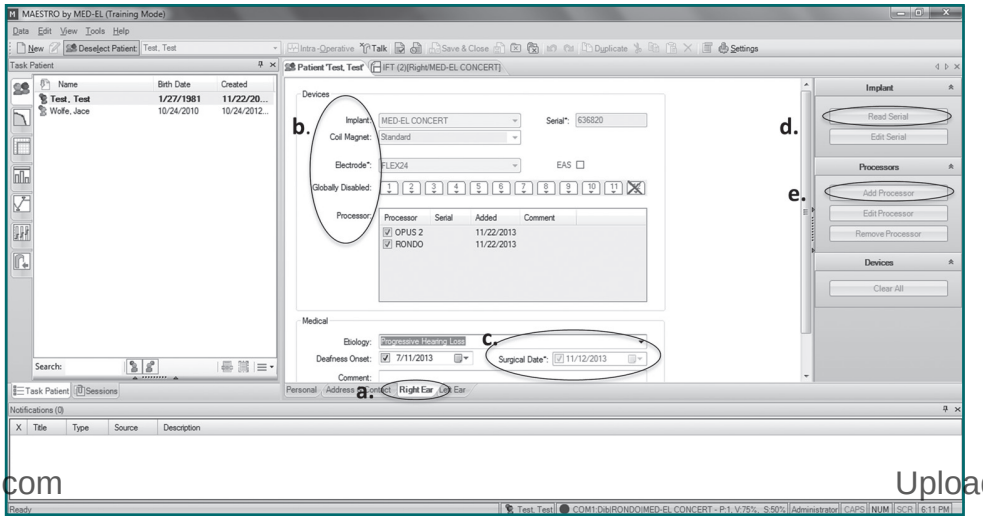



**FIGURE 6–1.** Entering new patient information into "personal" tab including demographic information (**a**) and implant information (**b**). *Note.* Image courtesy of MED-EL.




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**FIGURE 6–2.** Information necessary for programming the processor including **a.** which ear will be added; **b.** the internal device, coil magnet strength, electrode array, globally disabled electrodes (if extracochlear or if they are globally disabled at any point), and processor(s); **c.** the surgical date; **d.** read serial number button will determine the serial number for the internal device; **e.** for additional processors, select the add processors (this would be applicable if a patient was upgrading as well as a new activation). *Note.* Image courtesy of MED-EL.

## CONNECTING THE SOUND PROCESSOR AND CONDUCTING THE ELECTRODE IMPEDANCE MEASUREMENT

The first step in programming MED-EL cochlear implants is to connect the DIB II coil to complete Impedance Field Telemetry (IFT) measurements, which are used to provide an objective evaluation of the integrity of the implant electrode contacts and leads as well as the status of the electrode contact/tissue interface. Figures 6–3 and 6–4 show examples of normal and abnormal IFT measurements, respectively. Impedances are measured at the 12 intracochlear electrode contacts relative to

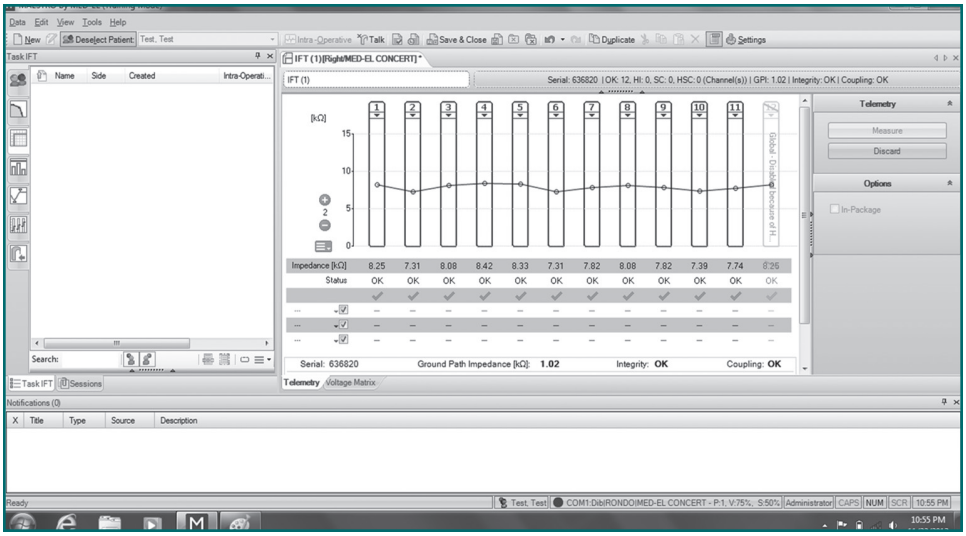



**FIGURE 6–3.** Normal Impedance Field Telemetry (IFT) measurement. *Note.* Image courtesy of MED-EL.

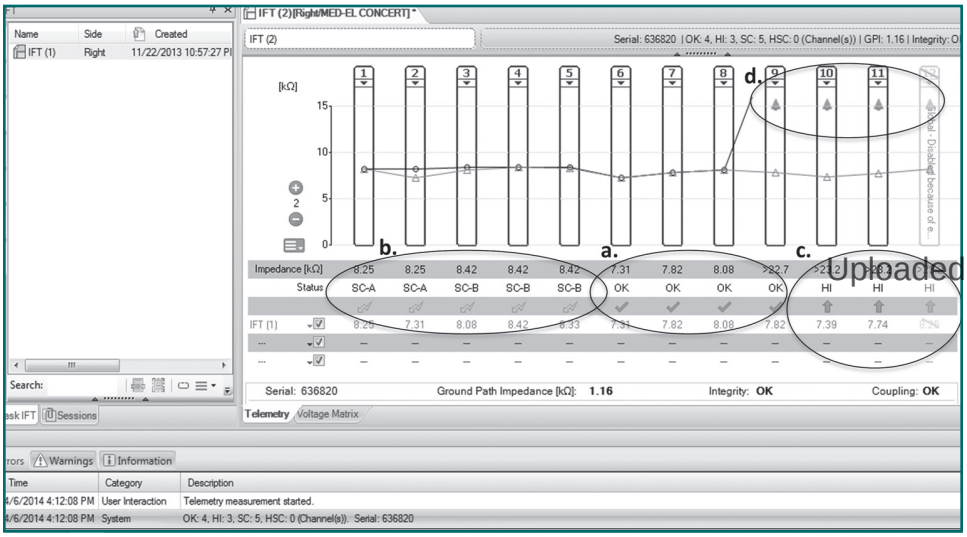



**FIGURE 6–4.** Abnormal Impedance Field Telemetry (IFT) measurement with electrodes that are/have **a.** Normal electrodes—"OK"; **b.** Short Circuit—"SC"; **c.** Open circuit (> or "HI"); **d.** "HI" impedance symbol. *Note.* Image courtesy of MED-EL.

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the reference electrode. The IFT task also allows for measurement of voltage distribution across all intracochlear electrodes when one active electrode is stimulated. These IFT measurements ensure the device is functioning properly and may guide or influence programming choices. Once the clinician has selected a patient, the IFT tab (the second tab in the task bar on the left side of the MAESTRO display) will automatically open.

Impedance results of the 12 intracochlear electrode contacts and one reference electrode contact are displayed in kiloohms in the IFT table. The impedance of the reference electrode is affected by the surrounding tissue, the current conduction through the extracochlear medium, and the mechanical status of the electrode. The electrode impedance of the intracochlear electrode contacts is influenced by the status and integrity of the electrode leads and contacts as well as the characteristics of the surrounding cochlear fluids and tissues (i.e., fibrous tissue growth, perilymph, macrophages, proteins).

The impedance value of intracochlear electrode contacts will typically fall within 5 to 15 kiloohms, while the typical ground path impedance is approximately 1 kiloohm. The overall status of each electrode is provided under the impedance value and includes one of the following five categories. First, the *"OK"* status suggests that the voltage is within normal limits and suitable for stimulation (see Figure 6–4a). Second, the *"HI"* status indicates that the voltage exceeds the normal limits of the system (open circuit) and cannot provide sufficient current flow (see Figure 6–4c–d). These channels should be deactivated. The software may also indicate an electrical impedance that is abnormal by noting a "greater than" (≥) sign next to the impedance value. When the "greater than" (>) sign is seen next to a number, it indicates that the impedance is too high to be accurately measured. In that case, the clinician should check for adequate loudness growth when measuring MCLs (see Figure 6–4c–d). If loudness growth is considered to be okay, then the clinician may leave that electrode active; however, if a full complement of channels are available for mapping, most clinicians elect to deactivate any "greater than" channels. If loudness growth is insufficient, then the clinician should consider disabling the electrode contact. It should be noted that, although it is within normal limits, the impedance of the ground electrode of COMBI and PULSAR implants is typically higher than the impedance of the ground electrode of SONATA and CONCERT implants. This is a relatively common finding that does not signify a problem with the COMBI or PULSAR implants but rather is due to the design of the ground electrode for those devices.

Third, the "*SC-x*" status refers to a short circuit and indicates that the signal also is traveling to another intracochlear electrode along with the electrode contact that is intended to be stimulated. The letter (e.g., "x") following the "SC-" allows the clinician to identify which channels are shorted together in the event that more than one short circuit is present (Figure 6–4b). For instance, the screenshot in Figure 6–4 indicates that electrodes 1 and 2 are each labeled at "SC-a," whereas electrodes 3, 4, and 5 are labeled as "SC-b." These findings indicate that electrodes 1 and 2 are shorted together, and electrodes 3, 4, and 5 are shorted together. Fourth, the "*SC?*" status indicates the rare situation where the software was unable to confirm the status of a possible shorted electrode and is intended to cue the clinician to check the voltage table manually for a possible short.

Electrode shorts can be manually identified by reviewing the "Voltage Distribution Profile" (VDP) table, which is described in the next section. If the voltage table confirms the presence of a short circuit, both electrode contacts should typically be disabled. However, if additional shorts are identified, the clinician should consider contacting a MED-EL clinical representative to report that abnormality and to seek further direction.

Fifth, the "HSC-x" status suggests abnormally high electrode impedance in addition to electrical coupling to another electrode. The result of this test is generally intended to identify possible short circuits when the implant is tested in its sterile packaging, prior to implantation. It identifies possible short circuits even in the presence of very high impedance on the channels (as would be expected when the electrode and ground contacts are not contacting fluid or tissue). Finally, a "HSC?" status is shown when the reference electrode impedance is not measured correctly (although this is a rare finding when measured in

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vivo). Again, the VDP table will aid the clinician with electrode management and review current spread on any given electrode.

In addition to the impedance value and status, the electrode impedance results table also displays the serial number of the internal CONCERT, SONATA TI[100], and PULSAR CI[100] devices and the total number of functional, shorted, and open electrodes. Furthermore, the table describes the "integrity of the internal device electronics" and the quality of the coupling between the DIB II coil and the internal device. On the Integrity test, an "OK" status indicates normal functioning of the internal device electronics. If a "--" status is measured along with normal electrode impedance results, the clinician should contact the MED-EL representative for further assessment of the internal device.

The quality of coupling may be described as either "OK," "Weak," "Faint," or "Poor." If the software indicates an "OK" status, it has determined that there is successful communication with the internal device. "Weak" or "faint" classifications means only partial data can be obtained. If there is insufficient coupling between the DIB and the implant, then the clinician should reposition the DIB coil over the recipient's implant and/or attempt to improve the implant-to-coil distance to facilitate better adherence of the coil and the desired measurement. In the case of "poor" coupling, the clinician should confirm that the DIB, cable, and coil are adequately connected and functioning properly. If the coupling problem continues, the clinician should contact the MED-EL representative.

The VDP table shows voltages measured across all electrodes when stimulating each electrode contact individually and provides a picture of how current fields spread in the cochlea. In the typical case, the voltage distributed to the stimulated electrode contact should be substantially greater than that at any of the other intracochlear electrodes. For example, if electrode 4 is the stimulated electrode contact and the voltage distributed to that electrode is 2.35 volts, then the voltage distributed across the other electrodes should be significantly lower than 2.35 volts (i.e., 0.3 volts may be measured across electrode 1). In practice, a small amount of voltage should be distributed across adjacent electrode contacts, but the values should be less than 50% of the voltage distributed across the stimulated contact. If voltages distributed across any other channel and the stimulated electrode are similar, it is likely that these two channels are shorted together. Figures 6–5 and 6–6 display a voltage distribution matrix/profile for both normal and abnormal IFT measurements, respectively. If the clinician has concerns or questions pertaining to the recipient's IFT measures or interpretation of IFT results, then he or she should send the recipient's file to the a MED-EL representative for further review.

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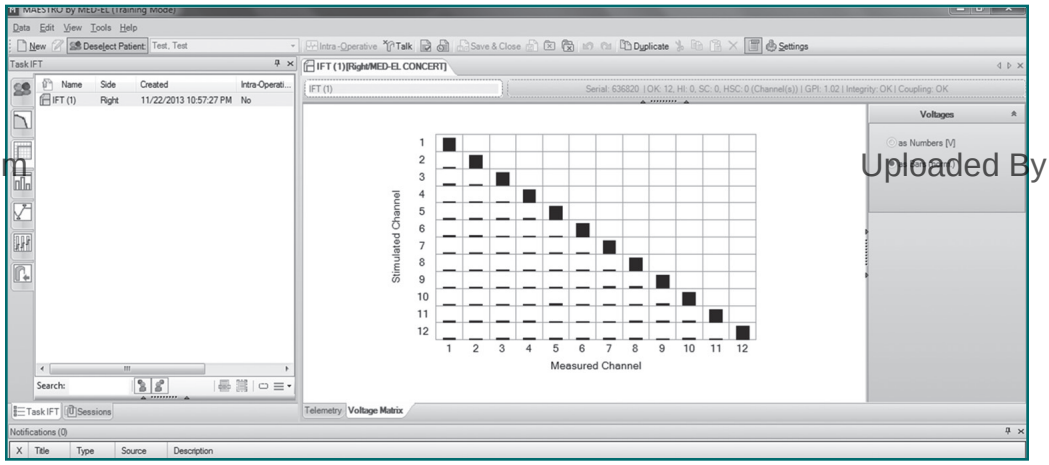



**FIGURE 6–5.** Voltage distribution matrix/profile for a normal Impedance Field Telemetry (IFT) measurement. *Note.* Image courtesy of MED-EL.

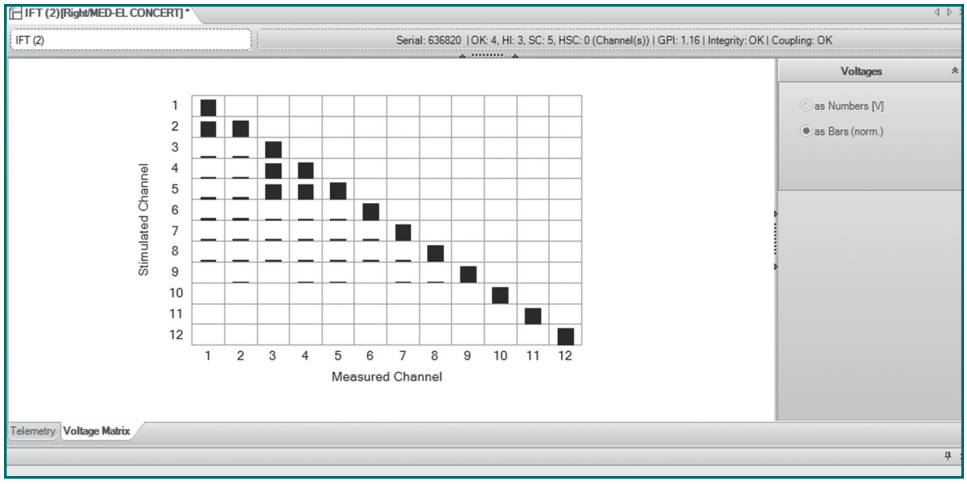



**FIGURE 6–6.** Voltage distribution matrix/profile for an abnormal Impedance Field Telemetry (IFT) measurement. *Note.* Image courtesy of MED-EL.

## CREATING A PROGRAM IN MAESTRO 4.0

### Selecting a Signal Coding Strategy

The next step in the MAESTRO programming platform is to select a signal coding strategy. The clinician must connect to the recipient's sound processor in order to create a MAP for MED-EL recipients. Then, the clinician must access the "Strategy" tab to select the signal coding strategy that the recipient will use (Figure 6–7). As described in Chapter 2, there are four MED-EL signal coding strategies from which the clinician may select: Fine Structure Processing (FSP), FS4, FS4-p, and one envelope coding strategy known as High Definition CIS (HDCIS). It should be noted that the FS signal coding strategies are not indicated for use with prelingual children in the USA. However, they are commercially approved for use with children in most other parts of the world, and many clinicians in the USA use these strategies with children with success (i.e., off-label use).

Previous research suggests that the FSP strategies yield better performance and sound quality relative to the older HDCIS strategy. FS4 and FS4-p provide fine structure processing over a larger number of apical channels (and thus a wider frequency range) when compared to the traditional FSP strategy. Therefore, the authors recommend initial use of one of the newer generation of the FSP strategies. If the recipient is not performing well with the FSP strategies after several weeks of use, the HDCIS strategy may be used. It may be wise to forgo use of parallel stimulation (e.g., FS4-p) for persons with significant cochlear abnormalities (e.g., common cavity, ossification), especially if the recipient is unable to provide feedback about the signal she or he receives from the implant. Parallel stimulation may result in undesirable channel interaction for persons with significant cochlear abnormalities. It should be noted, however, that the FS4-p strategy does possess a novel algorithm referred to as Channel Interaction Compensation, which determines the current amplitudes for parallel stimulation so that the compound current amplitude is equal to the current amplitude that would occur with sequential stimulation.

In the FS strategies, pulses on channels that are noted to be "fine structure" will be grouped together into bursts called Channel Specific Sampling Sequences (CSSS). The onset time of each burst varies with the zero-crossing of the input signal such that the bursts are timed to deliver information that reflects the temporal fluctuations in the input signal. The auditory nerve is more likely to be able to process fine structure cues in response to biphasic electrical stimulation when the signal is delivered at a relatively low frequency

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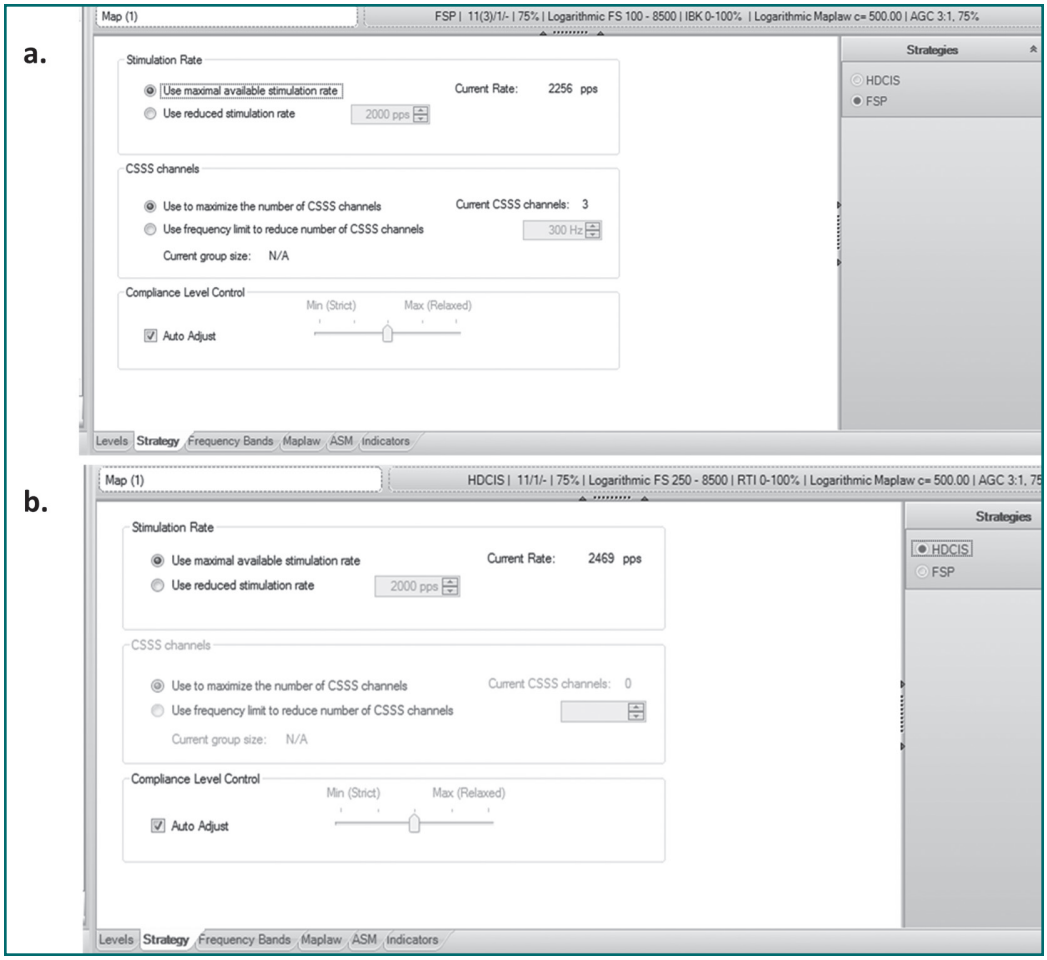



**FIGURE 6–7.** Selecting the "Strategy" tab for FSP (**a**) and HDCIS (**b**). *Note.* Image courtesy of MED-EL.

(i.e., definitely 400 Hz and below and possibly as high as 1000 Hz) (Clark, 1968; Moeller, 2001). Therefore, fine structure information is delivered to the most apical channels. For the traditional FSP strategy, fine structure is usually delivered to 1 to 3 channels. With the traditional FSP strategy, the actual number of fine structure channels that will be derived in any given program is dependent on the pulse width required to achieve a sufficient charge value and the recipient's overall stimulation rate. It should be noted that the range for FSP extends to approximately 350 Hz, whereas the range for FS4 extends to approximately 1000 Hz.

In the Signal Coding Strategy tab of MAESTRO (see Figure 6–7), the clinician may choose the frequency range for CSSS and the corner frequency above which CSSS will not be implemented for the traditional FSP strategy; although, MED-EL recommends the default, which seeks to maximize the number of CSSS channels. Again, the FS4 and FS4-p strategies incorporate CSSS stimulation on apical channels up to 1000 Hz in an attempt to maximize the frequency range over which fine structure is captured and delivered, and as a result, the clinician should typically select the newer FS4 strategies for use with most recipients. The typical range of CSSS stimulation for the traditional FSP strategy extends to approximately 350 Hz. The channels with FS/CSSS stimulation for all of the FS strategies are denoted by a music note icon next to the channel number in the programming screen. FS4-P differs from FS4 in that FS4-P stimulates two channels out of the available fine structure channels instead of one in each

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frame of stimulation. The stimulated channel(s) are selected based on which frequency filters logged a zero-crossing during the last sample. During FS4-P, MED-EL's Channel Interaction Compensation algorithm is in effect, which minimizes the negative effects of channel interaction so that the channels stimulated in parallel in order to preserve the independent information on each channel (rather than aggregating to form a "virtual" channel).

In this tab where the FS parameters are manipulated, the clinician may also set the maximum stimulation rate, which like other strategies, is partially determined by the pulse width on individual channels. The primary unit of stimulation in the MAESTRO software is charge-based, so that the software increases the charge delivered to a particular channel by either increasing current amplitude or pulse width, duration, or both of these parameters in response to the clinician's request for more stimulation during programming. The goals are to (1) use the narrowest pulse width possible to allow for fast stimulation rates, and (2) maximize power efficiency. In the process, the MAESTRO system takes into account electrode impedance as well as the limits of the current source.

At the time of this writing, the current version of MAESTRO programming software (version M4.1) does not allow the clinician to adjust the stimulation rate. Instead, the MAESTRO platform attempts to provide the highest stimulation rate possible without exceeding voltage compliance limits. If the level of stimulation requires the use of a wider pulse width than what is called for at the default settings, which is common for most recipients, the stimulation rate will automatically decrease. Higher stimulation rates are possible with deactivation of one or more of the 12 channels. The clinician should strive to maintain a stimulation rate of at least 800 pps. Refer to Chapter 2 for more information on stimulation rate and Chapter 3 for more information on programming considerations related to stimulation rate.

The clinician may also manage voltage compliance within the Signal Coding Strategy menu tab. The Compliance Level Control is an adjustable slider bar that determines how voltage compliance limits are measured. In short, the parameter determines how vigilant the system is at avoiding voltage compliance problems. These may be determined by (1) measured values (i.e., electrode impedance), (2) strict voltage compliance estimation, or (3) theoretical maximum limits using relaxed voltage compliance estimation.

The Compliance Level Control default, which is typically used by most clinicians for most recipients, is Auto Adjust. The Auto Adjust setting seeks to strike a balance between avoiding voltage compliance issues, maximizing stimulation rate, and setting the pulse width to allow for FSP coding in as many apical channels as possible. If the user is not experiencing adequate loudness growth with default settings, manual adjustments to the Compliance Level Control may improve performance.

It should be noted that even the oldest internal devices can be programmed using the newest strategies such as FS4, FS4-p, and FSP, with one notable exception. The FS4-p strategy is available only in MED-EL implants with the i100 signal processing chip (e.g., PULSAR, SONATA, and CONCERT)

## Setting Stimulation Levels

In the MAESTRO platform, stimulation levels are adjusted in charge units (qu), but clinical units (cu) can also be displayed. Charge units are defined as the product of the amplitude of the stimulus current and the pulse width divided by 1000 [i.e., charge unit = (amplitude of current × pulse width/1000). This unit of stimulation represents constant charge. The system automatically adjusts the amplitude of the stimulus current and/or the pulse width to provide the requested loudness while avoiding voltage compliance limits and attempting to maximize stimulation rate. The range of stimulation levels are approximately 0 to 282.2 qu.

Software measurement tools for adjusting the THR and MCL are shown in Figure 6–8. It is best to measure stimulation levels on all possible channels; however, an automatic interpolation feature is available in the MAESTRO software for both threshold and MCL measurements (see Figure 6–8). The electrical threshold is referred to as "THR" and is defined as the highest level at which

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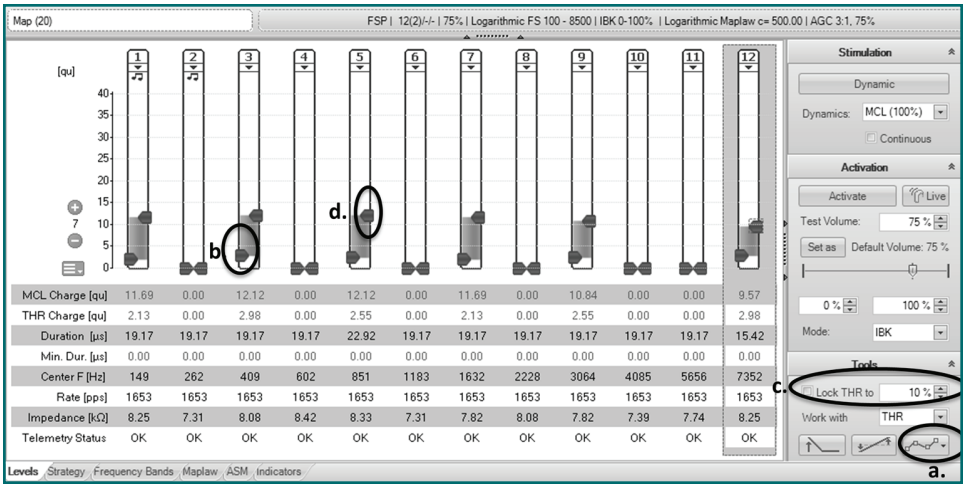



**FIGURE 6–8.** Threshold (THR) and most comfortable level (MCL) measurement tools: **a.** With the interpolate feature under tools, the clinician can select whether to work with THR, MCL, or THR + MCL; **b.** Handle on left side of each channel is used to manipulate THR levels; clinician can use track pad or up/down arrow keys to change stimulation level. Press the space bar to stimulate/present sound to measure THR; hold the space bar for continuous stimulation; **c.** To lock THR, mark the check box and adjust the percent of MCL (MED-EL's recommended default is 10% for adults and 0 for children; **d.** Use right handle to adjust MCL (manipulated the same way as the THR adjustment). *Note.* Image courtesy of MED-EL.

a response is not obtained. The handle on the left hand side of each channel is used to manipulate THR (see Figure 6–8). As a result, the clinician may obtain threshold via typical audiological procedures, and if the recipient complains of a persistent humming/frying/static noise, then the programming clinician would most likely need to provide a global decrease in the THR values until the recipient reports that the low-level sound is eliminated. For young children who may not be able to reliably report on the presence of a continuous low-level noise, then the clinician should consider globally decreasing THR values by two or three steps (approximately 10%) below the measured value. As mentioned earlier, the threat of a constant, audible low-level noise is a concern for recipients using CIS-based signal coding strategies. The signal coding strategies used with the MED-EL system stimulate all activated channels with every frame of data (i.e., CIS-based); if a channel does not have any information in it during a frame of stimulation, then that channel is stimulated at the THR level. As a result, when THR levels are set at a level that is consistently audible to the recipient, it is likely that a continuous sound (e.g., buzzing, humming, frying, static) will be detected.

However, it should be reiterated that MED-EL indicates that THR values should be estimated on the basis of the MCL settings (MED-EL, n.d.). Specifically, THR may be set at 5 to 10% of MCL. Additionally, it is suggested that THR be set at 0% for channels with FS/CSSS in order to reliably prevent audibility for low-level noise secondary to electrical stimulation. Previous research has shown similar performance is achieved when using a THR set to 10% of the upper-stimulation level versus a measured THR (Spahr & Dorman, 2005). The "lock THR to" option is selected on the right hand side of the Levels screen and will automatically set thresholds to a level determined by the clinician (10% of MCL, 0, etc.) (Figure 6–8c). At default settings, THR is measured using 300 msec bursts of electrical pulses, but it is common practice to create programs with estimated THR levels. When estimated THR levels are used, the stimulation levels for low-level inputs are based on the measured upper-stimulation levels.

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As previously mentioned, adequate audibility for soft sounds may be determined by measuring sound-field detection thresholds in the audiometric test booth. If sound-field thresholds are elevated (>30 dB HL), the clinician could consider increasing MCL values, selecting a steeper maplaw function, or measuring THR levels. After this programming change is made, sound-field measurements should be repeated to confirm adequate access to low-level sounds. Additional parameters to improve audibility of low-level sounds will be discussed in the following sections. Finally, if the patient notes a humming sound, especially in quiet environments, it is an indication that at least one threshold value is set to an audible level. The clinician needs to identify the channel that has an audible threshold and lower the THR level to a point where it is not audible to the patient. This can be achieved by sweeping through each channel at the THR level to determine what channel's THR is set to an audible level. Of course, the THR value of the offending channel would then be reduced. Alternative approaches involve globally decreasing the THR values until the low-level sound ceases or setting THR values to 0 to 10%.

The upper level of stimulation, MCL, refers to the maximum level of stimulation that is perceived as very loud, but not uncomfortable or painful with user volume set to 100%. It is important to note that when the volume setting is adjusted to 100%, a 100 dB SPL input will receive MCL stimulation. Inputs between 25 and 100 dB SPL are mapped in between THR and MCL based on a logarithmic function referred to as maplaw (to be discussed later). As a result, when conversation (at an average conversational level of 60–65 dB SPL) is occurring in live speech mode during the programming process, it is likely that the stimulation the recipient receives will be well below the MCL value.

The MCL level can be manipulated using the handle on the right hand side of each channel (see Figure 6–8d). When measured with a psychophysical approach, the recipient completes loudness scaling while 50 msec bursts of electrical biphasic pulses are delivered to a single electrode contact. MED-EL recommends beginning the MCL measurement on an electrode located in the middle of the array (electrode 5 through 8), then measuring on an electrode at the apical end of the array (electrode 1 through 4), and then an electrode at the basal end (8 through 12). Then, MCLs may be measured on the intermediate channels. MED-EL also recommends that the clinician remeasure the MCL on the first channel or two that were initially measured in order to ensure the reliability of those measurements (MED-EL, n.d.). It should be noted that most recipients of MED-EL implants have MCL values between 5 to 25 charge units. MCL values exceeding 40 charge units are uncommon.

Alternatively, MCLs may be set by globally increasing all MCL values in live speech mode (i.e., "flat MAP") to the recipient's MCL value. Then, channel-specific MCL values may be made based on the recipient's report or via loudness balancing. MED-EL recommends that loudness balancing be completed at 80% of the MCL value. Loudness balancing at 80% of the MCL should result in equal loudness across the array for input levels that approximate typical speech signals encountered to realistic listening situations.

Loudness balancing for two channels may be completed by using the computer mouse to highlight the MCL of two neighboring channels while pressing the "shift" key. Pressing the keyboard "space bar" will then present stimulation at the selected percentage of MCL. Sweeping may be completed by holding down the "shift" key and highlighting all of the channels across the array. Pressing the "space bar" will sweep stimulation across all selected channels at the indicated percentage of the MCL. Pressing down the "control" key will allow the clinician to highlight channels that are not adjacent to one another. Pressing the "space bar" will, then, allow for stimulation of the selected channels. Refer to Chapter 3 for additional information on determination of upper-stimulation levels.

Stimulation is delivered by clicking on the MCL slider of a channel and then pressing the "Dynamic" icon/radio button (with "Dynamic" set to 100%) in the upper right hand corner of the "Levels" menu screen. Additionally, the space bar may be used to deliver stimulation. The clinician may use the "Page up/down" keys to change stimulation levels by 15% of the MCL. Furthermore, the "arrow up/down" keys and the "+ / –"

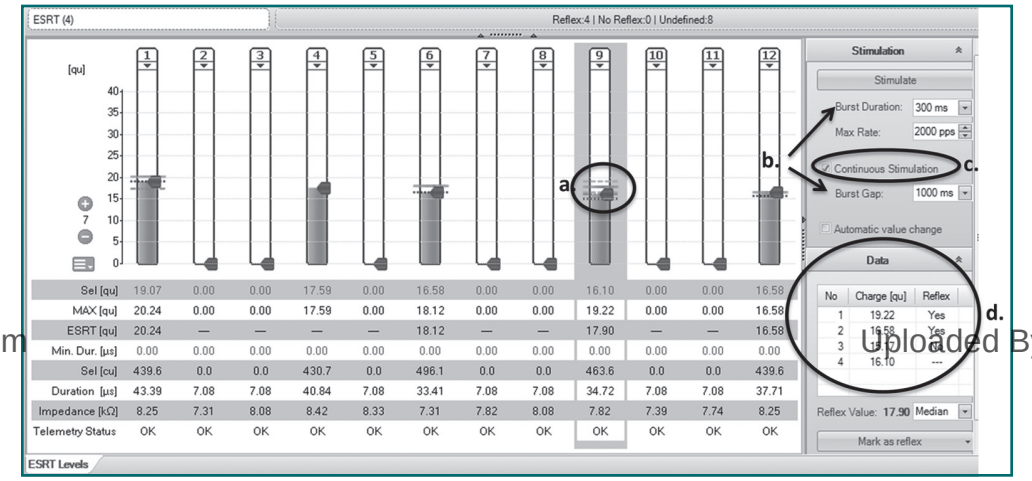
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keys may be used to change stimulation levels by 3 and 1%, respectively. The MAESTRO software provides an alert message anytime that the clinician increases the stimulus level by more than 25% between stimulus presentations. The clinician must dismiss this message prior to delivering stimulation.

Based on the results of research (Brickley et al., 2005), MED-EL strongly advocates the use of the electrically evoked stapedius reflex threshold (ESRT) as a guide for setting MCL for patients who are unable to respond behaviorally to setting an MCL level. ESRT can be obtained in the normal fitting screen, but a special platform in MAESTRO provides a method to track and better manage ESRT measurements (Figure 6–9 shows the eSRT platform features). In the ESRT platform, the stimulus burst has a longer duration (i.e., 300 msec at default) than the stimulus used to determine MCL. The longer duration of the stimulus will make the stapedial reflex more salient for measurement with the acoustic immittance system. MED-EL recommends setting the burst duration to 300 ms and the gap to 1,000 ms to allow the auditory system to recover between presentations (MED-EL ESRT Guide). Within the ESRT platform, the clinician can mark the response as a reflex using the following key strokes: mark as reflex (R), mark as no reflex (N), or mark as undefined (M).

The measured ESRT can be imported into the recipient's program in the Levels screen in the ESRT platform (Figure 6–10 is an overview of options to import ESRTs into the patient's map). The ESRT has a strong correlation to MCL (Brickley et al., 2005) and, therefore, the clinician should attempt to measure it in all recipients without tympanostomy tubes or obvious middle ear dysfunction. In general, most recipients perform quite well when MCL values are set to the same levels that elicited to ESRT. The clinician should be suspicious of MCL values that are 10% to 15% higher than the ESRT. This may indicate that the MCL values are excessive. Additional information on ESRT measurements is provided in Chapter 3.

After ESRT are measured, the authors of this book suggest that the clinician conduct loudness balancing at 80% of the MCL across the electrode array (MED-EL, n.d.). Loudness balancing after ESRT measurement will ensure equal loudness across the frequency range and also will typically result in a flatter MCL profile across the array, which usually supports satisfactory performance.




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**FIGURE 6–9.** ESRT platform features: **a.** ESRT measurements: solid line is marked as reflex, dotted line is marked as no reflex, and dashed line is marked as undefined; **b.** Recommended fitting parameters with burst duration set to 300 ms and burst gap set to 1,000 ms; **c.** Check box to maintain continuous stimulation once space bar is pressed, and press space bar again to stop stimulation; **d.** Data indicating levels at which a reflex was present, not present, or undetermined. *Note.* Image courtesy of MED-EL.

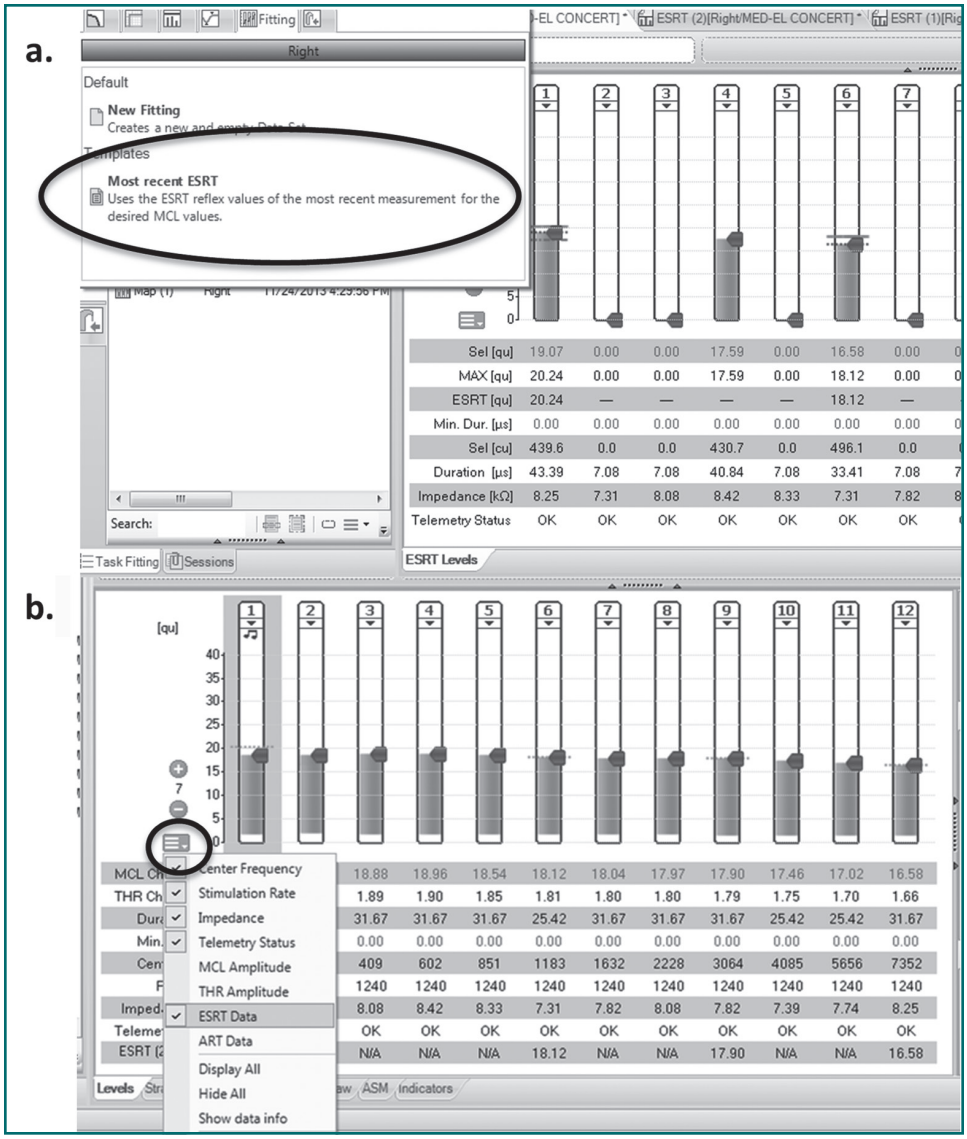



**FIGURE 6–10.** Options to import ESRTs into the patient's map include select a new map based on ESRTs by selecting "New" and choosing Templates: "Most recent ESRT" (**a**) and import ESRT to existing map by selecting drop-down (see circled icon) and checking ESRT data (**b**). *Note.* Image courtesy of MED-EL.

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Furthermore, the clinician should consider conducting sweeping (at 100% of MCL) at the end of the session to ensure that all MCLs do not exceed the user's discomfort threshold and do not elicit any other nonauditory side effects.

Finally, when activating in live speech mode, the clinician may choose the percentage of the electrical dynamic range where stimulation will be presented. When activating the device, the clinician should select "Activate," which turns the processor microphones on (deactivate turns the processor microphones off). When the clinician selects "Live" it allows the processor microphones to remain active while changes are being made. This facilitates additional adjustments to the programming based on patient feedback.

MED-EL recommends activating in live speech mode with the volume control set to a reduced setting (e.g., 60% or lower) that is certain to avoid recipient discomfort. Then, the user's test

volume should be increased to 100%. If the recipient expresses signs of discomfort with the user test volume set to 100%, then the clinician should globally decrease MCL values until optimal comfort and clarity for speech are achieved at a volume setting of 100%.

Within the software, there are additional tools that can be used to streamline the process, including pivot and shift and tilt (Figure 6–11). These tools are generally used to quickly address the complaints of adult recipients (e.g., too tinny, too hollow, not sharp enough). It should be noted that these tools provide a coarse adjustment to the recipient's program, and the use of ESRT and loudness balancing tends to be more successful in addressing these types of complaints.

## Additional Programming Parameters

### *Frequency Bands*

The Frequency Bands screen allows the clinician to control how input sound frequencies are allocated across the active channels. As stated in Chapter 1, MED-EL has the longest commercially available electrode array in the CI industry, which provides access to more apical sites of the cochlea. As a result, the MAESTRO software has a low cut-off of 70 Hz, which is quite lower than the low end of the frequency range of the implant systems of competing manufacturers. The overall frequency range spans from 70 to 8500 Hz. The lower end of the frequency range may be adjusted in 10 Hz

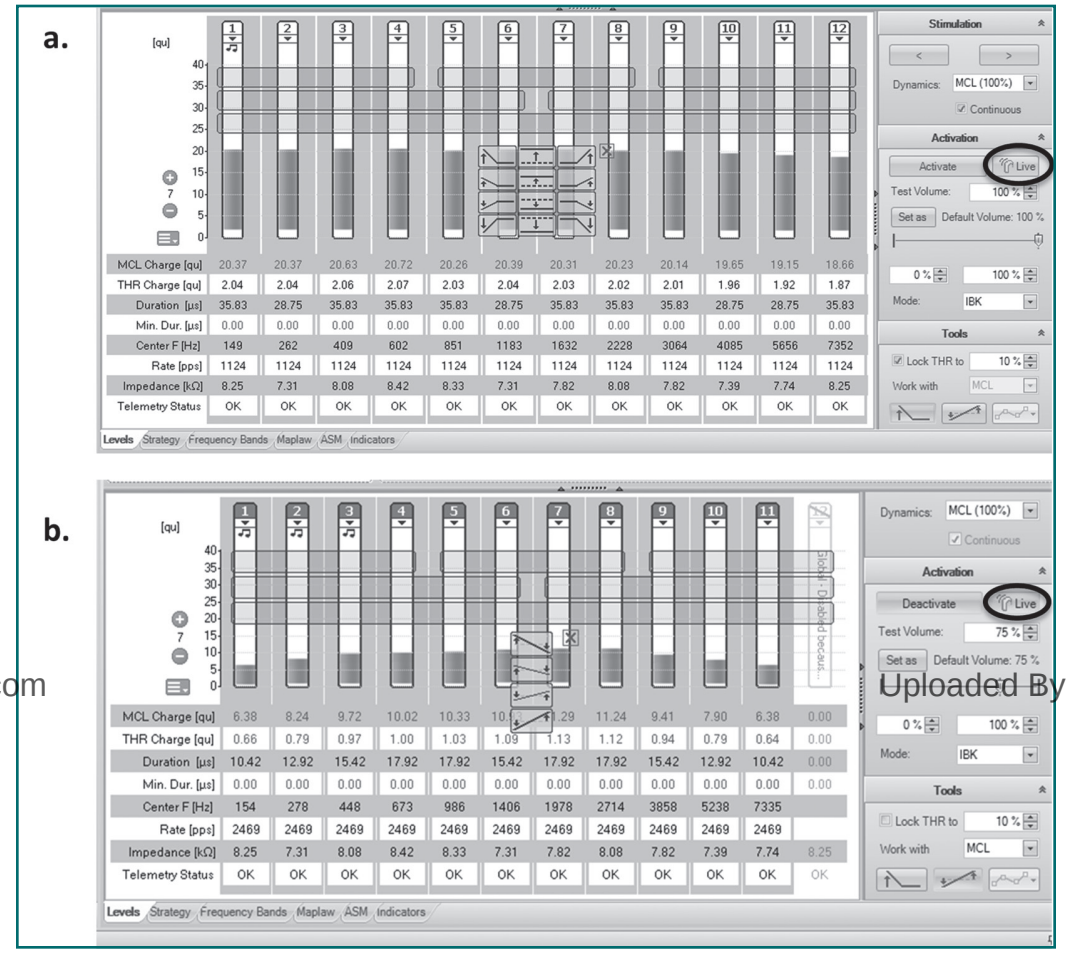



**FIGURE 6–11.** Tools for adjusting most comfortable level (MC) and threshold (THR): **a.** Shift and Tilt Function: If "Live" is selected, the processor microphones will remain active while changes are made and **b.** Pivot function. *Note.* Image courtesy of MED-EL.

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steps from 70 to 350 Hz (default = 100 Hz for FS strategies), whereas the upper end of the range can be adjusted in 500 Hz steps from 3500 to 8500 Hz (default = 8500 Hz). Setting the upper end of the frequency range slightly lower (e.g., 7000 Hz) may be beneficial because it will create channels with narrower bandwidths in the speech frequencies. However, the upper range of 8500 Hz will serve most recipients well.

Frequencies within the defined input range are reallocated to existing channels using one of four approaches. The default and recommended allocation for the majority of participants is "Logarithmic FS" in which the input frequencies are assigned to logarithmically spaced bands. The second option, "LinLog," separates the lower frequencies into linear bands and the higher frequencies into logarithmic bands. This mode results in narrower low-frequency bands, which will likely provide better spectral resolution in the low frequencies. Improved spectral resolution may be helpful for discriminating common speech and environmental sounds. The "Tonotopic" approach attempts to mimic the tonotopic organization observed in the normal cochlea. Theoretically, this mode will produce the most normal frequency percept. Finally, "Linear Increasing" divides the input frequencies into linear bands with increased bandwidth from the apical to basal direction. This strategy results in wider bandwidths for low-frequency channels relative to "Logarithmic FS," but the highest-frequency channels are narrower. This mode may be helpful if the recipient exhibits confusion patterns with high-frequency sounds (e.g., /f/ vs. /s/ vs. /h/).

The Frequency Band screen allows the clinician to reorder how channels are allocated to electrode contacts, which is necessary (and should be performed first) when programming a split electrode (Figure 6–12).

It should also be noted that the clinician may change the order in which the channels are allocated to the 12 stimulating electrode contact sites. This is accomplished by using the computer mouse to select a channel and then dragging it to a desired location. It is very unlikely that the clinician will have to reorder the channel to electrode assignment.

### Maplaw

The maplaw screen allows the clinician to control the logarithmic function used to map acoustic inputs into the recipient's electrical dynamic range. The default choice in the maplaw menu is "Logarithmic maplaw," which maps lower-level inputs to a higher point in the electrical dynamic range. The other logarithmic function, "S-shaped maplaw," provides less amplification for low-level sounds and a steep increase in amplification for moderate and high-level inputs. The "S-shaped

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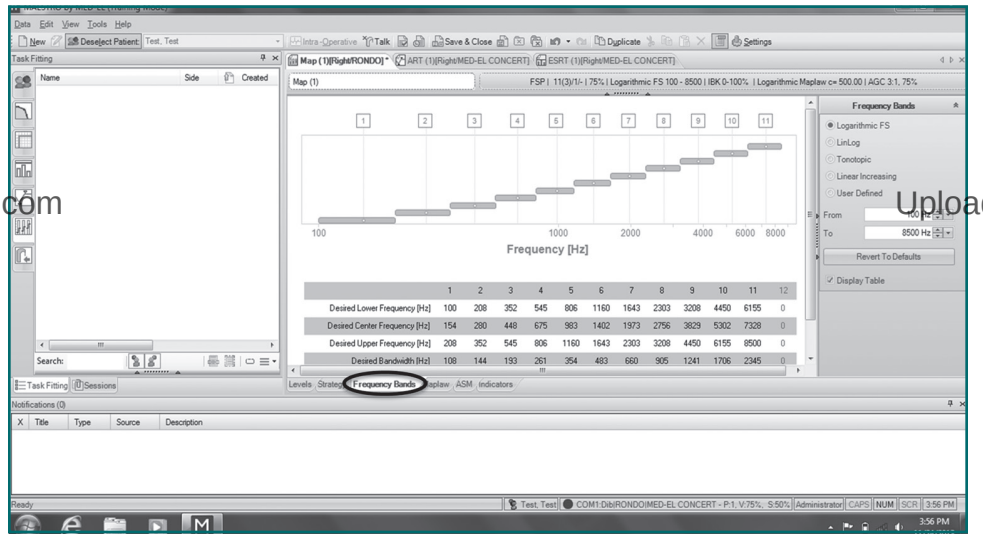


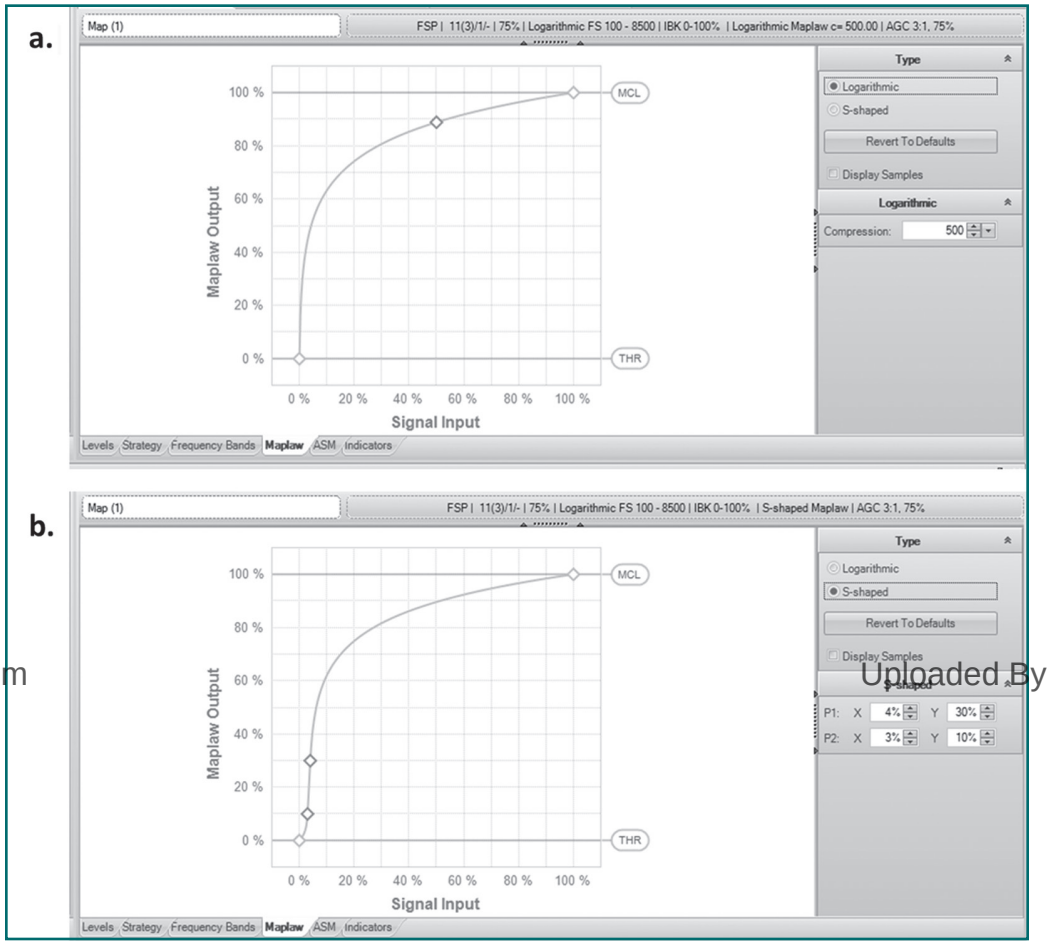

**FIGURE 6–12.** Frequency bands tab in the map parameters. *Note.* Image courtesy of MED-EL.

maplaw" can be helpful in controlling constant background noise and can be used with patients who need to manage a specific noisy situation (i.e., sitting near a ventilation fan at work) or for adult recipients who are newly activated and are annoyed by the newfound audibility for low-level sounds.

The clinician also may adjust the default compression coefficient (i.e., 500) used to define the logarithmic function. Higher coefficients map softer inputs to a higher point in the electrical dynamic range. The authors recommend using the compression coefficient of 500 for recipients who use measured THR or THR set to 10% and a compression coefficient of 1,000 for recipients with THR set to 0%. Often, at initial stimulation for patients who have hearing experience (adults), a maplaw of 1,000 will be perceived as too "crisp" or "sharp." In this case, the clinician may need to wait until the patient has a few days of experience with the system before attempting to use a higher maplaw. Long-term, improved audibility of low level sounds should improve performance. Figure 6–13 shows a logarithmic and S-shaped maplaw.

### Volume Mode

The two volume options on the sound processor include the IBK (Innsbruck) and RTI (Research Triangle Institute) modes. With the IBK default mode, volume control settings correspond to stimulation at a percentage of the electrical dynamic range 0 to 100%. Two parameters can be adjusted




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**FIGURE 6–13.** Comparison of maplaws: **a.** Logarithmic-use compression coefficient of 500 until the patient has had time to adjust; then increase to 1,000 to improve access to low-level sounds; **b.** S-shaped-use in situations where there is a constant background noise (i.e., ventilation system). *Note.* Image courtesy of MED-EL.

in the IBK mode, "minimum" and "maximum" volume control effects, which determine the overall range of the volume control. The minimum and maximum parameters denote the level that will be presented when the volume is turned all the way down or up, respectively. The upper range of the volume control determines the maximum stimulation that the recipient will receive (re. MCL setting of the program). The upper range should be set to 100%, which will be appropriate for the recipient if the clinician has ensured during that programming session that MCL values are comfortable for the recipient.

Setting the lower end of the volume range to 0% will allow the recipient to decrease the stimulation level to THR. For most patients, the range should be set from a minimum of 80% to a maximum of 100%. This will allow the user to achieve a reasonable decrease in stimulation if needed, while also providing a manageable volume control range. When the lower end of the volume range is set to 80% (with an upper end of 100%), then the user must decrease the volume control of the FineTuner by 16 steps to go from the maximum to minimum of the volume range. Setting the lower end to 0% (with an upper end of 100%) will require the recipient to decrease the FineTuner volume control by 64 steps (i.e., presses or clicks). A narrower range can be used to deactivate the volume control, which may be helpful for children and adults who cannot reliably adjust their volume controls. In addition, the volume buttons on the FineTuner remote control can be deactivated.

The RTI volume control mode is rarely used by clinicians. At the minimum position, no stimulation is provided. As the volume control is increased, the minimum and maximum stimulation levels are increased toward THR and MCL, respectively. One disadvantage of the RTI mode is that some inputs may be inaudible when lower volume settings are used.

### Automatic Sound Management (ASM)

The ASM screen allows the clinician to adjust input processing characteristics of the system, such as automatic gain control (AGC). The MED-EL system is designed to use front-end processing (ASM) to provide a wide input dynamic range (IDR) of 75 dB by way of AGC, much like an adaptive sound window of 55 dB that roves based on the input to the sound processor. This effectively maps inputs from as soft as 25 dB to as loud as 100 dB SPL. Within the ASM menu, the two parameters that may be adjusted include compression ratio and microphone sensitivity.

The compression ratio of the AGC system may be disabled (although it would typically be undesirable to do so) or set to one of four ratios (i.e., 2:1, 2.5:1, 3:1, or 3.5:1). The default ratio is 3:1. A lower ratio may be selected if the user reports that loud sounds are too soft (e.g., music). At lower compression ratio settings (e.g., 2:1, 2.5:1), the difference between loud and soft sounds is increased, which may be useful in high-level noise environments because the background noise may be reduced. Higher compression ratios (e.g., 3.5:1) may address reports that loud sounds are uncomfortable (although this complaint also should prompt the clinician to re-evaluate and possibly decrease MCL levels). At a 3.5:1 compression ratio, the difference between low-level and high-level sounds will likely become less noticeable for the recipient. Background noise will be perceived as being louder.

The microphone sensitivity controls the gain of the microphone of the sound processor. The sensitivity parameter may be adjusted over a range from 0 to 100%. Higher sensitivity settings will provide better audibility of very soft sounds that are close by. However, moderate-level sounds may be too loud, and performance in noise may deteriorate. In contrast, lower sensitivity settings may improve performance and comfort in noise, but soft sounds will not be audible. The default setting of 75% is appropriate for most recipients. The patient can control sensitivity with the FineTuner remote control. (Figure 6–14 shows a view of the ASM tab.)

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### Indicators

The Indicators screen controls warning signals (e.g., low battery beep) and confirmation beeps (e.g., switching programs). The clinician can disable all alerts and control the loudness of enabled alerts according to user preferences (Figure 6–15 shows options on the Indicators page).

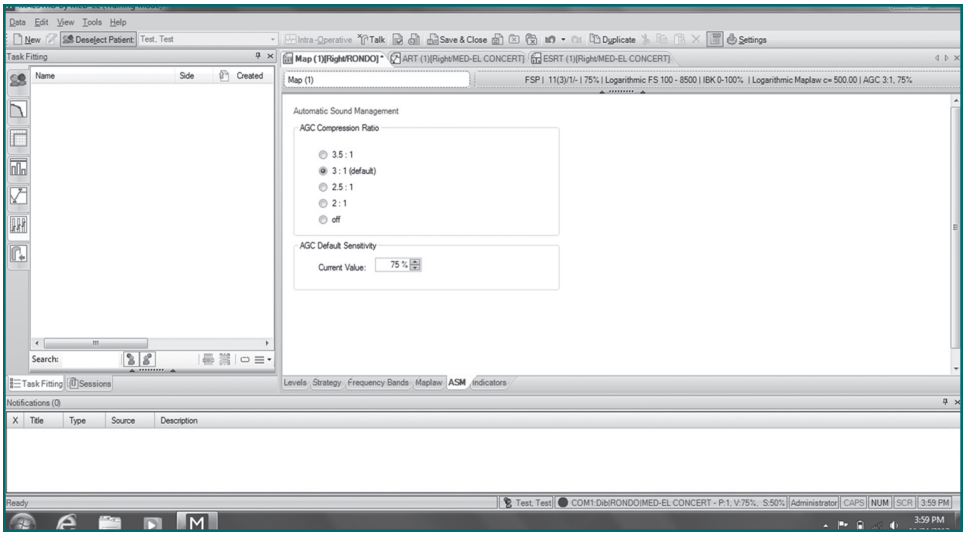



**FIGURE 6–14.** Automatic Sound Management settings. *Note.* Default AGC ratio is 3:1 and default sensitivity is 50%. *Note.* Image courtesy of MED-EL.

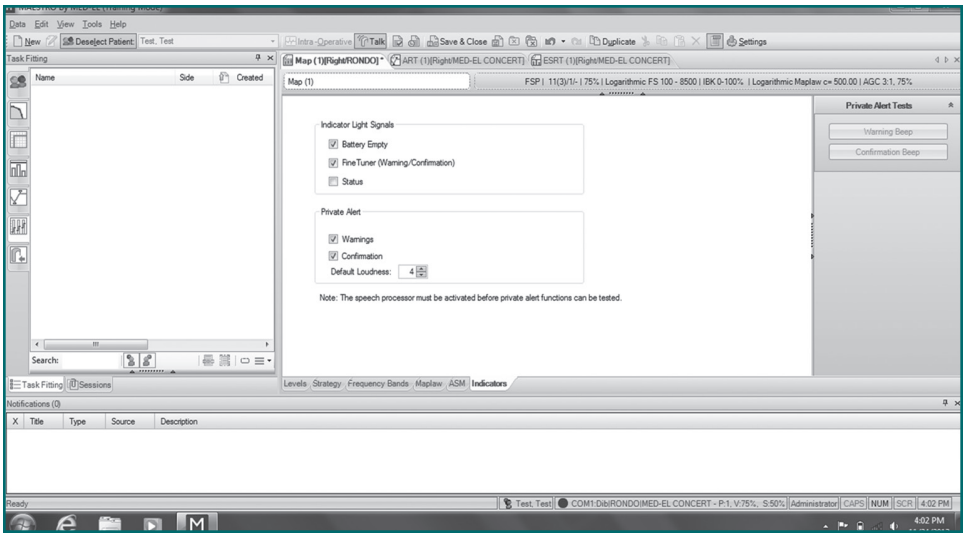



**FIGURE 6–15.** Indicators page. *Note.* The audiologist can enable or disable indicator light signals and private (audio) alerts and can also demonstrate these private alerts. *Note.* Image courtesy of MED-EL.

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## Finalizing Programming

The final steps in saving the programs and configuring both the processor and FineTuner can be completed in the last tab: Configuration. The clinician can load the desired programs in up to four individual slots and then select "program processor" (Figure 6–16). Under the FineTuner tab, the clinician can also activate and deactivate certain features. The authors recommend deactivating the "sensitivity" settings until the patient has had more experience with using the cochlear implant for some time. In the early stages of implant use, it is generally preferable for the recipient to adjust

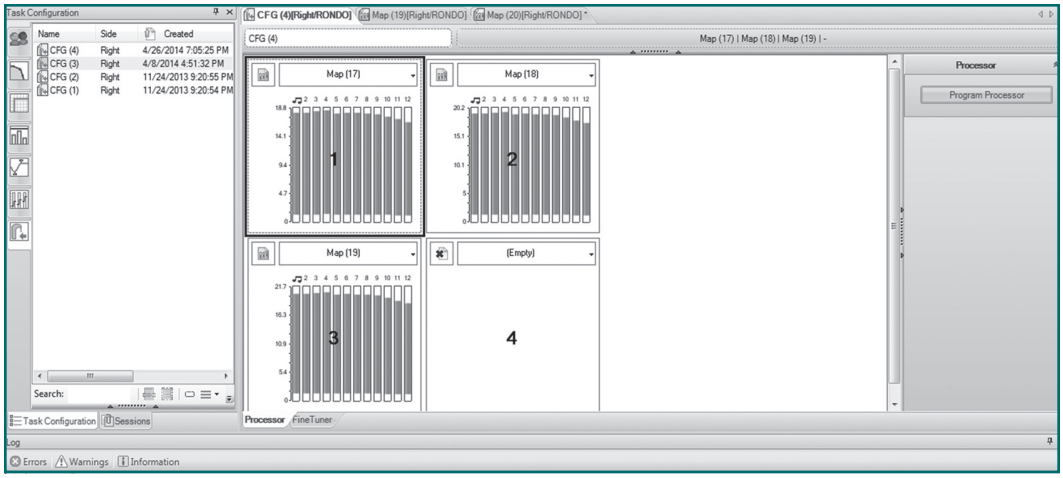



**FIGURE 6–16.** Processor configuration screen: place selected programs into each slot and select "program processor." *Note.* Image courtesy of MED-EL.

the volume control setting to achieve a desired loudness for speech and other environmental sounds. The telecoil may also be deactivated at first until the patient acclimates (Figure 6–17).

## MED-EL SPECIFIC TROUBLESHOOTING

In situations where patients are reporting difficulty with their equipment, there are several options to complete troubleshooting to determine the problem source. The Speech Processor Test Device (SPTD) can be used to check for shorts in the cable. The clinician should move the cable around while saying a long vowel (i.e. ahhhhhh) to check for shorts. If there does not seem to be any obvious short, but the patient is still experiencing intermittencies, try simply replacing the cable first. Replacing the coil should be the second troubleshooting step. In the case of poor sound quality or decreased sound quality, the Microphone Test Device (MTD) can be used. It requires disconnecting the battery module (whether it be the OPUS 2 or the RONDO) and attaching to the appropriate cord that connects to the MTD. The clinician can use earbuds to listen to the microphone input to determine the integrity of the microphones.

The processors also have light indicators that may help with troubleshooting. The company provides a troubleshooting guide that includes various error patterns that may be emitted by the indicator lights (Table 6–1). If you get an indicator that you do not recognize, turn the processor off and back on. If the abnormal indicators persist, consult the troubleshooting guide or contact the company for assistance.

## ADDITIONAL CONSIDERATIONS

If the clinician would like to share a recipient's MAPping information with another clinic or a MED-EL representative, or if programs need to be remotely loaded onto a recipient's replacement processor prior to being sent to the patient, the clinician may access the "Data" drop-down menu and select "export" to check the information that must be exported. The file should be saved as a packed data file (.mpd) so it can be opened in MAESTRO, and the clinician must also send MAP numbers and configuration via email to expedite remote loading. If statistical analysis is the purpose of the export, the file should be saved as an XML file.

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## AUDITORY RESPONSE TELEMETRY (ART)

The Auditory Response Telemetry (ART) platform is used to record the electrically evoked compound action potential (ECAP) of the auditory nerve in

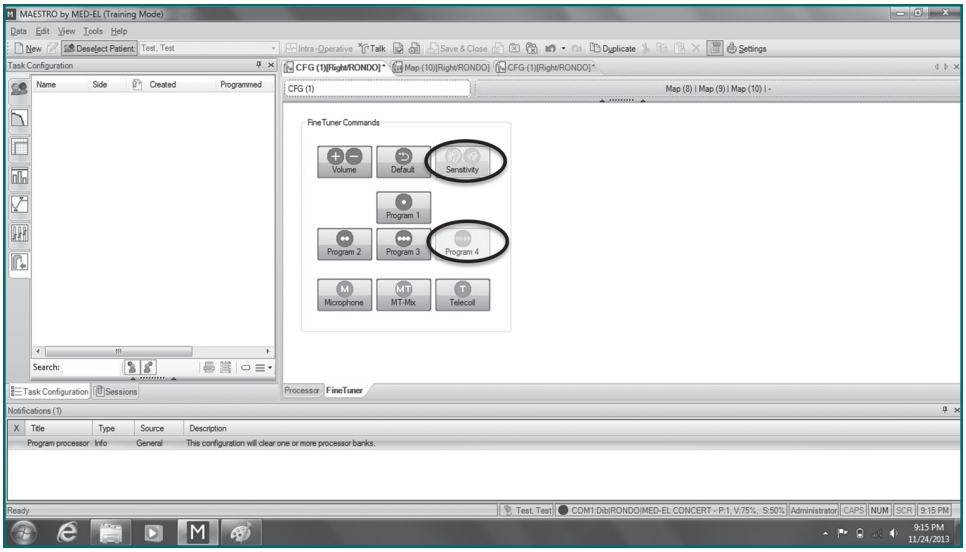



**FIGURE 6–17.** Fine Tuner Configuration: note that Sensitivity is deactivated and Program 4 is also shaded because there is no program loaded into that program slot. *Note.* Image courtesy of MED-EL.

**Table 6–1.** Error and Warning for the OPUS 2 and RONDO Processors

| *Error Pattern* | *Cause of Error/Warning* | *Necessary Action* |
|---|---|---|
| 1 blink every 2 seconds | Electronic issue or disturbance | Turn processor off and back on |
| 2 blinks every 2 seconds | Selected program not programmed or program failure | Choose another program |
| 4 blinks every 2 seconds (quick blinks in pairs) | Electronic issue or program failure | Turn processor off and back on |
| 3 quick blinks every 2 seconds | Electronic issue or disturbance | |
| Continuous quick blinks | Battery empty | Change batteries |
| Continuous light | Maximum or minimum volume or sensitivity reached | Cease button pushing on FineTuner |

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response to electrical stimulation from the MED-EL cochlear implant system. The reader is referred to Michelle Hughes's book entitled "Objective Measures in Cochlear Implants" (another title in the Plural Publishing, Inc. Core Clinical Concepts in Audiology Series) for an excellent review on the ECAP measurement. A couple of clinically relevant aspects of ART are discussed here.

First, to conduct ART measurements, the clinician must access the "ART" icon in the MED-EL MAESTRO software. Also, the DIB coil must be used to administer the ART assessment. Then, the clinician must define the maximum and minimum stimulation levels that encompass an amplitude growth series that may be used to determine the ECAP threshold (Figure 6–18). The clinician may manually determine the ECAP threshold in ART via visual analysis (similar to ABR assessment) or may estimate the ECAP threshold via the zero-crossing of the *x*-axis calculated from a best-fit line calculated from the measures obtained in the amplitude growth series (Figure 6–19).

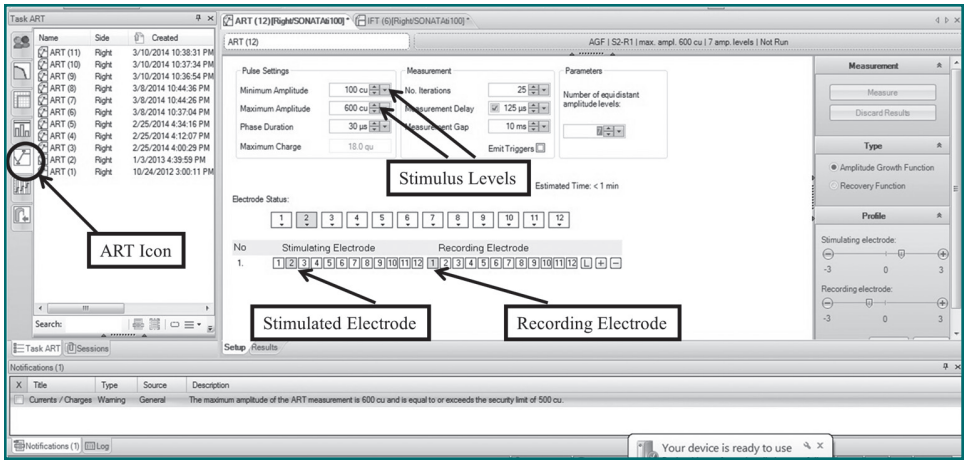



**FIGURE 6–18.** ART measurement parameters including stimulus levels, stimulating electrode, and recording electrode. *Note.* Image courtesy of MED-EL.

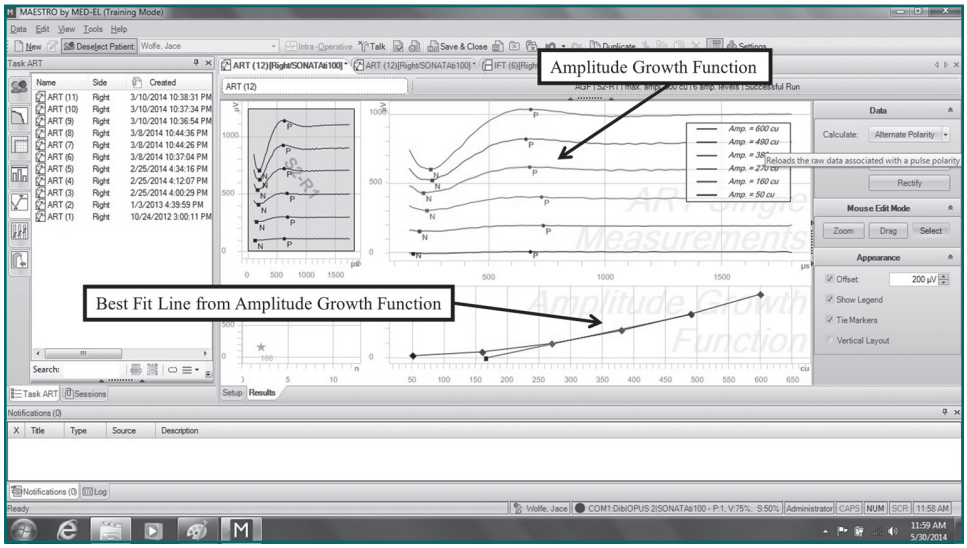



**FIGURE 6–19.** The ECAP threshold may be estimated through a visual analysis of the ECAP threshold or from a best-fit line calculated from the measures obtained in the amplitude growth series. *Note.* Image courtesy of MED-EL.

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Second, as mentioned in Chapters 3 and 7, the ECAP threshold has, at best, a modest correlation to behavioral stimulation levels (e.g., THR and MCL values). As a result, the clinician should exercise caution in trying to predict behavioral stimulation levels for recipients on the basis of the ECAP (e.g., ART) measurements. The recipient is better served when the clinician relies on behavioral measures and the ESRT to determine stimulation levels. As stated elsewhere in this book, the ECAP (e.g., ART) threshold does indicate a stimulation level that should be audible to the recipient, which in and of itself may be an important piece of information that assists the clinician in the programming process (i.e., determining a level that may be used for conditioning, etc.). Also, the ECAP/ART threshold typically falls within the THR and MCL for most recipients. However, as mentioned in the general discussion of ECAP in Chapters 3 and 7, the ECAP measured in all implant systems

may exceed a recipient's upper-stimulation level. One should anticipate that the ART threshold may exceed MCL on some electrodes for a minority of recipients (e.g., approximately 30% of recipients). Clinicians should be concerned and apply further scrutiny to the recipients' programs that have MCL values that are markedly higher or lower than the ART threshold.

## KEY CONCEPTS

- The provision of appropriate MCL levels is one of the most important requirements for creating an optimal program for MED-EL recipients.
- THR is typically estimated from MCL measures for most MED-EL recipients. Research has suggested that MED-EL recipients' access to low-level sounds is sufficient when MCL levels and maplaw (e.g., 1,000) settings are appropriate.
- The ESRT has been shown to possess an excellent correlation to MCL for MED-EL recipients.
- Fine Structure Processing is designed to convey the fine structure of speech and environmental sounds and may improve performance across a variety of listening situations.
- The MED-EL OPUS 2 processor may be used with contemporary digital RF/RM receivers. Additionally, the OPUS 2 sound processor is equipped with a built-in telecoil, which allows the recipient to use a vast array of hearing assistance technologies (e.g., induction neck-loop, induction room loops, telephones, etc). These hearing assistance technologies should be considered to optimize performance in challenging listening situations.

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# 7

# Clinical Considerations: Putting All of the Pieces Together

The previous chapters in this text introduced the basic components of programming a cochlear implant. The objective of this chapter is to assimilate the information into a practical guide for creating successful programs for cochlear implant recipients. General information also is provided on recipient counseling, habilitation/rehabilitation, and interdisciplinary teaming.

## COLLABORATION WITH THE COCHLEAR IMPLANT SURGEON/ MEDICAL EVALUATION

Prior to the initial activation of the sound processor, the clinician should inquire about preoperative and postoperative imaging results and the medical evaluation from the cochlear implant surgeon. Preimplant imaging studies, such as Computed Tomography (CT) or Magnetic Resonance Imaging (MRI), will reveal any abnormalities within the cochlea. For instance, ossification or fibrous tissue growth in the cochlea may occur for patients who are postmeningitic. Ossification may prevent full insertion of the electrode array into the cochlea or elevate the requisite stimulation levels. Additionally, these imaging studies will reveal the presence of a common cavity cochleae or Mondini's dysplasia. These conditions also may prevent a full insertion as well as be associated with higher stimulation levels, fluctuating stimulation needs, and nonauditory side effects (i.e., facial nerve stimulation).

An MRI also may show the status of the auditory nerve. Several publications suggest that some children with congenital deafness are born with aplastic auditory nerves (i.e., absent or severely deficient auditory nerves; Buchman et al., 2006; Rance, 2005), which certainly would result in diminished outcomes and success with a cochlear implant. However, in many of these cases, the aplasia was unilateral. Identification of unilateral aplasia would be critical in selecting the ideal ear for implantation. This information is obviously important to consider during the preimplant counseling process. Additionally, obtaining an MRI prior to surgery might be desirable as it is difficult to obtain an MRI of the head after cochlear implant surgery.

The clinician also should inquire about any postoperative x-rays that were conducted to evaluate the status of the electrode array insertion. This information is very important as it indicates how many stimulating electrode contacts are inserted into the cochlea and any extracochlear electrode

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contacts that should be disabled. Finally, the clinician should query the cochlear implant surgeon about any other surgical/postsurgical complications that might influence the programming of the cochlear implant. Recipients with complications may exhibit anxiety at the activation appointment and often will benefit from positive encouragement from the programming clinician.

At the initial activation and all remaining appointments, the clinician must perform a visual examination of the incision site, an inspection of the area where the transmitting coil will be placed, and an otoscopic examination to rule out the presence of purulent middle ear effusion. Most clinics activate the sound processor 2 to 4 weeks after surgery to allow the incision site to heal and for the resolution of otitis media with serous effusion in the middle ear space, which is not uncommon a few weeks post surgery. The middle ear effusion often requires 3 to 6 weeks to resolve. This is a particularly important factor for recipients who had a considerable amount of low-frequency hearing prior to surgery and for whom an attempt has been made to preserve that low-frequency hearing to allow for electroacoustic stimulation (EAS). For these recipients, it is imperative to measure air and bone conduction thresholds at all octave and interoctave frequencies beginning at 125 Hz in order to effectively determine how to provide acoustic amplification.

At the activation appointment, inspection of the skin is important to ensure that the magnet strength in the external transmitting coil is appropriate. Swelling present at the implant site in the day of activation may necessitate a stronger magnet to facilitate good coil retention. However, when the magnet is too strong, it may cut off blood flow around the coil and compromise the health of the skin underneath the coil. It is imperative that the clinician closely monitors the implant site and magnet strength while any swelling resolves in order to ensure that the magnet strength is appropriate. The programming clinician should consider a medical referral when there is evidence of trauma or inflammation at the incision site. Also, medical referral should be considered when there is persistent purulent otitis media or otitis media with serous effusion that persists beyond six weeks postactivation.

## PROGRAMMING SCHEDULE

### Two-Day Initial Activation Sessions

As previously mentioned, the optimal stimulation levels for a recipient normally change over the first few weeks or months of cochlear implant use. As a result, frequent programming appointments are necessary to ensure maximum performance and benefit. Frequent programming sessions throughout the first months of implant use are also beneficial because they provide the clinician with the opportunity to counsel the recipient and his or her family on proper care, use, and maintenance of the implants as well as strategies for optimizing the recipient's outcome. After the first year of implant use, fewer appointments are necessary.

For most recipients, sound processor activation should be divided into two consecutive days. As shown in Table 7–1, a 2-hour session should be scheduled for the first day and 1½-hour session for the second day. The second day allows the recipient to provide feedback to the clinician about early experiences, and the clinician can use this feedback to make adjustments and fine-tune the programming of the sound processor.

On day 1, after the physical examination, the clinician will determine the appropriate magnet strength to support adhesion of the external transmitting coil to the recipient's head. Young children and elderly women often exhibit thin skin flaps and require relatively weak magnets. Middle-aged men and recipients who are obese, however, usually have thick skin flaps and require strong magnets. For users with very thick skin flaps, it may be helpful to shave the hair directly underneath the transmitting coil. The authors of this book keep razors in the clinic to allow for shaving of the implant site if the presence of hair prevents coil retention or the delivery of a consistent signal. For very thin skin flaps, it may be helpful to place a layer of mole skin on the underneath side of the transmitting coil to provide cushion between the coil and skin. Cochlear also has produced coil spacers (a small plastic disc that clips to the underneath side of the coil) to alleviate

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**Table 7–1.** Overview of 2-Day Activation Appointments

| *Time Postsurgery* | *Duration* | *Basic Overview of Typical Procedures* |
|---|---|---|
| 2 to 4 weeks | 2 hours | • Visual examination of incision site and otoscopy<br>• Counsel caregiver to monitor incision site<br>• Basic information: early expectations, use, care, volume control, and batteries<br>• Determine processor magnet strength<br>• Measure impedance at electrode contacts<br>• Select signal coding strategy and base parameters<br>• Determine T levels, USL, volume control |
| 2 to 4 weeks | 1.5 hours | • Visual examination of incision site and otoscopy<br>• Additional information: proper use, care, habilitation/rehabilitation, and telephone use<br>• Demonstrate mastery of basic processor functions<br>• Adjust stimulation levels, measure ESRT* and ECAP<br>• Frequency-specific programming<br>• Adjust and fine-tune programming, restrict volume control |

*Note*. * = if recipient will tolerate procedure; *ECAP* = electrically evoked compound action potential; *ESRT* = electrically evoked stapedius reflex threshold; *T* = threshold levels; *USL* = upper-stimulation levels.

signal interruptions for Nucleus users who have thin skin flaps.

Determination of the correct magnet develops with experience; however, the manufacturers provide guidelines on typical magnet strength for children and adults. Although the clinician may be tempted to select a magnet strength that prevents the coil from ever falling off the recipient's head, it is normal for the coil to occasionally detach during rigorous physical activity (e.g., gymnastics, basketball, wrestling, or rough child's play). A magnet with an appropriate strength should be selected cautiously to avoid blood flow restriction to the skin over the internal device. This restriction could result in a skin flap breakdown and the possible need for internal device removal. If the clinician can hold the external coil several inches from the side of the recipient's head and the transmitting coil is aggressively drawn to the internal magnet, then the coil is likely too strong. However, if the transmitting coil falls off the user's head several times during programming, the magnet is obviously too weak. The reader is encouraged to refer to Chapters 3 through 6 for a more detailed discussion of the determination of appropriate magnet strength for implant recipients.

Regardless of the initial magnet strength, the clinician visually should examine the skin flap at every clinical visit. If any sign of irritation occurs or a very prominent impression of the coil can be seen in the user's skin, the magnet strength should be reduced. Adult users may be advised to reduce their own magnet strength if they feel tightness or discomfort around the coil site.

Caregivers of young recipients should be counseled to routinely monitor the status of the incision site. If the child's hair is cut substantially, the caregivers should monitor the skin flap closely for several days following initial activation. A change in skin flap thickness may alter the power level required to deliver the signal without intermittency. This may occur when the recipient has a haircut or when he or she gains or loses a substantial amount of weight. The Advanced Bionics 90K implants use bidirectional telemetry to allow for continuous, real-time measurement

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of the minimum power level needed to deliver the signal without intermittency, and as a result, changes in the skin flap are not an issue. For other recipients, it may be necessary to remeasure the requisite power level after a substantial change in skin flap thickness.

After the magnet is selected for the transmitting coil, the user's sound processor is connected to the programming computer via a device-specific interface. Contemporary cochlear implant programming software contains several icons that inform the clinician that the computer is connected to the programming interface and that the external sound processor and cochlear implant have been identified. These indicators are helpful in signifying that all equipment is engaged and prepared for the programming process. They are also helpful tools to use for the troubleshooting process to identify components that may be faulty (e.g., sound processor, coil, cable, cochlear implant). When the programming computer indicates successful communication with the recipient's internal device, the clinician will measure impedance of the stimulating and reference electrode contacts. The programmer must disable identified short or open electrode circuits. The audiologist must, then, select the appropriate signal coding strategy and initial parameters. Refer to Chapters 4 through 6 for direction on choosing signal coding strategies and various adjustable parameters for each implant system.

Next, the clinician will need to remind the patient about realistic expectations in the early stages of cochlear implant use. For an adult recipient, a suggested discussion is as follows: "Remember, when we turn your cochlear implant on today, you may not be entirely pleased with the sound you receive. In fact, you may leave here today thinking, 'I've made a big mistake.' Also, when we first turn it on, many people say that voices sound much different from what they remember with better hearing. Some people note that voices sound cartoonish (like a chipmunk), mechanical, or robotic. In some instances, recipients comment that speech sounds more like beeps or noise. Don't worry. As we've discussed before, it will take the hearing centers of your brain some time to adjust to this new signal. You should notice considerable improvement in sound quality and speech understanding over the next few weeks and months. It will just take time and experience to reach that point. So, the main point today is to not panic or get too upset if the initial sound you receive is less than ideal."

For the caregivers of young children, the conversation immediately prior to activation will be different and may go as follows: "Remember, the initial response a young child shows to sound from a cochlear implant varies widely from child to child. Some children may cry or become upset. It's unlikely that the sound is uncomfortably loud; they are simply unsettled by hearing sounds they've never experienced before. Other children may smile, wave, or point to their ear. However, in many cases, a child's first responses are subtle. For instance, they may stop playing with the toys, they may glance at you, me, or the programming clinician, they may change their sucking pattern on their pacifier, or they may slightly change their facial expression. Finally, on occasion, children may exhibit very limited responses. Regardless of your child's initial response, remember that it is not at all indicative of how they will eventually perform with the cochlear implant. Also, please remember that your child's auditory skill level is similar to that of a newborn; therefore, sound will have little meaning today. It will take time for the brain to associate meaning with signals from the cochlear implant. As a result, the early responses may be very inconsistent and less than what you might have hoped. Take heart in the fact that we fully expect your child to benefit from the implant with time and experience."

After the recipient's expectations are established, the programming audiologist will determine the signal coding parameters and appropriate stimulation levels for the recipient. This is likely the most important component of the implant programming process, and research has suggested that a substantial degree of variability exists in the methods employed to create cochlear implant recipients' programs (Vaerenberg et al., 2014). Refer to Chapter 3 for a detailed description of measuring threshold and upper-stimulation levels. During the first 2 days of activation, it may be difficult for the adult patient to respond

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to programming stimuli at threshold (i.e., T level) because of inexperience with electrical stimulation and constant tinnitus. For young, prelingually deafened children with limited to no experience with sound, initial thresholds are likely minimal response levels and are suprathreshold in nature. As a result, the actual T levels are likely to be slightly lower than the measured minimal response levels. It is often necessary to provide a small global decrease in T levels from the values measured to programming stimuli. The child's T level responses to programming stimuli may include changes in sucking patterns and/or facial expressions, a time-locked head turn to the stimulus, a glance toward the parent, clinician, or assistant, a change in play, smiling, pointing to his or her ear, crying, and so forth. Parents and assistants should help the clinician observe these subtle responses.

It should be noted that some manufacturers do not recommend or even discourage the measurement of T level (e.g., MED-EL). This recommendation is supported by research that shows adequate access to low-level sounds for many recipients when T levels are estimated, particularly when mapping functions (e.g., maplaw) are optimized (Boyd, 2006). The authors of this book measure minimal response levels for all children in order to establish the amount of stimulation that is necessary to provide audibility for the recipient. The measured T levels are included in the programs that are created for Nucleus and Advanced Bionics recipients. Based on the recommendation of the manufacturer, the authors are more inclined to base T levels on the upper-stimulation levels for MED-EL recipients. However, for some MED-EL recipients, measured T levels are included in the program. In such cases, the authors measure T levels and then globally decrease these measured values to a level at which the we have a high degree of confidence that the recipient will not experience a continuous low-level buzzing/humming noise secondary to consistently audible, low-level electrical stimulation. Ultimately, the authors' objective is to provide low-level stimulation that will result in sound-field detection thresholds that are no higher than 25 dB HL for children and 30 dB HL for adults. Fine tuning of T levels is conducted when sound-field detection thresholds do not meet these criteria.

For measurement of upper-stimulation levels, the authors strongly recommend that parents and caregivers watch their child for signs of discomfort. If a child becomes uncomfortable, the clinician should exercise caution with continued increases in stimulation to avoid overstimulation. Excessively loud stimulation is unpleasant for adults, but it may particularly complicate the early stages of implant use for young children. Specifically, if the child associates the cochlear implant with an unwanted or unpleasant percept, he or she may refuse to wear it. Therefore, when initially activating the implant in live speech mode, threshold and upper-stimulation levels should be set at a low level and increased slowly as the clinician or family member talks to and observes the reaction of the recipient. For young children, T levels are fixed at the initial-measurement level. However, after a response is obtained, increases to upper-stimulation levels should continue slowly while the child's response is observed.

The final upper-stimulation levels for children are based on numerous factors including typical stimulation levels of the cochlear implant system, behavioral responses during programming, electrically evoked stapedial reflex thresholds (ESRT), and observations made during live speech mode. For instance, for young recipients using the Nucleus 6 system, the authors of this book attempt to set the upper-stimulation levels at least 40 clinical units above the T levels. However, if a child is uncomfortable with these levels, a reduced dynamic range is used for the initial program(s). Remember, the primary goal of the activation appointment for a young child is to facilitate bonding with the implant by providing an audible and comfortable signal across the frequency range, not to create the perfect program. As a colleague of ours is fond of stating, "The process of developing spoken language and auditory skills for a young cochlear implant recipient is not a sprint, but rather a marathon." Given the importance of consistent audibility during the critical period of language development, it is necessary to move quickly toward the optimal program for a child. However, the entire "26.2 miles of the

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marathon" need not be completed during the first day. Ideally, the clinician will seek to achieve optimal stimulation levels by the 1-month post-activation appointment. Once again, a detailed description of the determination of T levels and upper-stimulation levels may be found in Chapter 3.

For adult recipients, it is important to establish expectations for how speech and environmental sounds likely will be perceived when the sound processor is initially activated in live speech mode. The following instructions may be provided to the recipient: "We are now ready to turn on your cochlear implant so you can hear speech and other environmental sounds. We will start at a very soft volume level, and you most likely will not hear anything at first. After that, we will very slowly increase the volume until you can hear. At first, you may hear a buzzing or ringing sound from the implant. Don't worry. This sound will stop after you wear the implant for a short time and your program settings are optimized."

When live speech first becomes audible, an adult may initially complain. To address this complaint, the clinician may temporarily decrease upper-stimulation levels; however, it is best to leave the stimulation on. In many cases, the slow increases to the upper-stimulation levels will result in better sound quality. The adult recipient should be asked to indicate when the global upper-stimulation level elicits the most comfortable listening level. In the authors' experience, the initial level selected by the recipient is not the most ideal level; therefore, it may be more desirable to continue slowly increasing stimulation in small step sizes and ask the adult to indicate when the signal is too loud. The clinician may tell the patient, "I am going to increase the 'volume' by a small amount to see if it makes your implant sound better. If it is too loud, then we will turn the 'volume' back down." From this point, the clinician then may decrease the upper-stimulation levels by small increments (e.g., 2–5 CL/CU or 1–3% in MED-EL) and confirm that the sound quality is satisfactory. In many cases, recipients request an increase in stimulation levels after acclimating to the sound from the cochlear implant for a day. In other cases, the user may report that certain sounds are too loud. If this occurs, the clinician may decrease upper-stimulation levels in the corresponding frequency range or recounsel the recipient about using the volume control. Furthermore, some users will report hearing some sounds (e.g., low-frequency sounds) but struggling to hear others (e.g., high-frequency sounds). In this case, the clinician can make frequency-specific programming changes to alleviate the difficulties. Finally, at the end of every programming session, the clinician may wish to ensure that the recipient can at least detect, if not discriminate, all six Ling sounds. This step ensures sufficient audibility for sounds throughout the speech frequency range (see Chapter 3).

During the first few weeks of cochlear implant use, it is wise to provide a full range for the volume control for the adult and child recipient. This allows them to initially place the transmitting coil on the head with a minimal amount of stimulation and then they may increase the stimulation to their desired level. A wider volume control range also provides the recipient with flexibility in stimulation levels to suit their needs as they acclimate to electrical hearing across a wide variety of environments. Access to a wide volume control range is especially important for the caregivers of young children. Specifically, the volume control can be set to the minimum position when the processor and coil is placed on the head and increased to the level deemed appropriate in the programming session within a few seconds. If the child reacts negatively to sound as the volume control is being increased, then the volume should be set temporarily to a level that is lower than what was prescribed at the programming session in order to facilitate comfort and full-time implant use. The caregiver should be counseled to attempt to increase the volume setting to the preferred setting throughout the first day or few days of implant use. For children, the volume control is always capped so that the upper-stimulation levels provided within a given program will not exceed what was considered appropriate in the programming session. For adults, the authors of this textbook generally provide a little "headroom" beyond the volume control setting that was deemed to be most comfortable during the programming session so the recipient may access more stimulation if needed. Typically, the clinician should provide about a 20 to 25% increase in headroom beyond the volume

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control setting used during the programming session. For example, the volume control of the Nucleus 6 sound processor ranges from 1 to 10 in one-unit steps. The clinician should consider setting upper-stimulation levels so that a most comfortable listening level is established at a volume control setting of 6 with the volume control range set to 50%. With these settings, increasing the volume control from the setting of 6 to the maximum setting of 10 will result in a 20% increase in stimulation for moderate to high-level sounds. To further clarify this concept, the reader is referred to Chapters 4 through 6 for a detailed discussion of the effect of the volume control setting for the contemporary processors offered by each implant manufacturer.

A common practice (particularly with pediatric recipients) during the early stages of cochlear implant use is to provide the recipient with multiple programs with successively greater stimulation levels (i.e., progressive programs/MAPs). This is especially helpful for people who exhibit significant tolerance problems during the first 2 days of activation. For instance, in the Nucleus 6 processor, four programs may be provided with a 3 to 5 clinical unit increase from one program to the next. In the Advanced Bionics and MED-EL systems, multiple programs with progressive global increases in upper-stimulation levels will also allow the recipient flexibility to increase total stimulation level. The recipient or family caregiver is encouraged to try to progress through each of the four programs by the next programming session. Again, if the recipient exhibits signs of distress at any program, the previous program should be used. Furthermore, if program #1 is too loud for the recipient, the volume control setting can be reduced from the recommended setting. Finally, if the clinician is satisfied with the stimulation levels that the user will tolerate at the initial activation session, it is not necessary to provide multiple programs. In fact, in most cases, the authors of this book find it to be simpler to provide one program with a wide volume control range which allows the recipient to access the desired amount of stimulation across a wide variety of environments. Situation-specific programs (e.g., noise, telephone, music) are not typically introduced until after at least 1 full week of cochlear implant use to allow the recipient and family to master the more basic functions of the cochlear implant sound processor.

To aid in the determination of appropriate stimulation levels during the 2-day activation process, the clinician may choose to complete objective measures of cochlear implant function including the ESRT and the electrically evoked compound action potential (ECAP). As previously mentioned, the ESRT is an excellent tool to establish appropriate upper-stimulation levels. However, due to the high-level stimulus necessary to elicit the ESRT, some recipients (i.e., young children) may not tolerate this testing during the first 2 days of activation. During the first several sessions, the acoustic immittance probe should be placed in the ear contralateral to the cochlear implant. If the user received simultaneous bilateral cochlear implants, the clinician should conduct otoscopy and tympanometry and place the immittance probe in the ear with the best admittance. If the ESRT is not obtained in the selected ear, it may be worthwhile to remeasure the ESRT in the opposite ear if it has reasonable admittance at the tympanic membrane.

The stimulus used for ECAP possesses a much lower stimulation rate than the stimuli used for programming and live speech. This lower rate results in a softer loudness percept at an equivalent level of stimulation (i.e., CL/clinical unit/charge unit) because of temporal summation. As a result, most recipients can tolerate the measurement of the ECAP on some (and typically all) channels during the first 2 days of use. The ECAP measurement system is referred to as Auditory Response Telemetry (ART), Neural Response Imaging (NRI), and Neural Response Telemetry (NRT) in the MED-EL, Advanced Bionics, and Cochlear Corporation systems, respectively. The clinical utility of the ECAP is discussed later in this chapter.

Finally, the clinician must counsel the recipient and his or her family about care, use, and maintenance of the cochlear implant equipment. Recipients often become overwhelmed if every single aspect of the external equipment is discussed during the first day (e.g., accessories, warranty, implications for MRI). For this reason, the authors recommend discussing and reviewing the

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basics of implant use during the first 2 days and more advanced topics at subsequent programming sessions. It is helpful to have the recipient or a family member demonstrate mastery of the most basic functions, such as powering the processor on and off, adjusting the volume control, and changing the batteries.

Additionally, it is useful to discuss a wear schedule with the recipient. In most cases, the authors of this textbook subscribe the philosophy of "eyes open, ears on." In other words, when a recipient is awake, she or he should be using the cochlear implant, even if the stimulation levels are lower than what was recommended by the clinician. Some clinicians may suggest that recipients should gradually build up to using the cochlear implant during all waking hours. The authors of this book do not agree with that opinion. When the implant is programmed appropriately, it should be used to enable audibility for speech and environmental sounds at every moment that the recipient is awake. Possible exceptions include when the recipient is experiencing considerable pain/discomfort or a nonauditory side effect that cannot be resolved with a change in the volume control setting. The recipient should be encouraged to contact the clinic if the implant cannot be worn during all waking hours.

Another exception to full-time is when there is a chance that a non-water-proof processor might be submerged in water. However, most manufacturers now provide solutions that enable sound processor use during activities that the processor may get wet. Implant use during all waking hours is especially imperative for pediatric recipients. Research has conclusively demonstrated that consistent audibility of intelligible speech is necessary for children to develop age-appropriate speech, language, and auditory abilities (Moog & Geers, 2003; Sharma, Dorman, & Spahr, 2002). Every moment that the implant is not used is a moment in which the child is deprived of stimulation that facilitates auditory brain development. The clinician has an important responsibility to stress the importance of full-time cochlear implant use to families across multiple programming sessions. Of course, the adjustment that an adult will make to his or her cochlear implant is also expedited when the implant is used during all waking hours.

Some recipients inquire about whether they should use the implant while sleeping. The general recommendation is for the implant to be removed while the recipient sleeps. This provides the implant site an opportunity to be free from the pressure of the coil, and it gives the recipient an opportunity to charge the sound processor batteries. Some adult recipients insist on using their implant at night because they have to be able to hear important sounds while sleeping (e.g., a new mother needs to hear her baby). In these cases, the recipient is encouraged to closely monitor the coil site to ensure that there are no signs of irritation or inflammation. Some parents also report that their children refuse to remove the implant before going to bed. In these cases, the parents are counseled to allow the child to fall asleep with the implant, and then it may be removed once the child is asleep.

During waking hours, it is also important to encourage the recipient to explore the sounds in her or his environment. Parents are encouraged to take their child on a "listening walk" in which they create and point out common sounds around the house (e.g., the telephone ringing, the dog barking, a musical instrument, water running). Adults are also encouraged to actively explore the sounds of their environment and to ask a friend or loved one to identify the source of sounds with which they are unfamiliar.

Table 7–2 provides a suggested schedule for discussing various topics throughout the first few weeks of cochlear implant use. Furthermore, the manufacturers provide excellent literature and videos describing the use, care, and maintenance of the cochlear implant along with contraindications for medical procedures. Finally, it is beneficial to provide some basic aural habilitation/rehabilitation during the first 2 days of activation. Specific suggestions are provided later in this chapter.

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## One-Week Postactivation Appointment

Following the 2-day activation appointment, children and adults should come back for a 1-week postactivation appointment to adjust programming parameters. This appointment and all subsequent appointments are scheduled for 60 to 90 minutes depending up on the needs of the recipi-

**Table 7–2.** Suggested Clinician Discussion Topics the First Few Programming Sessions

| Activation Appointment: Session 1 |
|---|
| **A. Basic Processor Operation**: (1) Powering processor on/off, (2) Changing and recharging batteries, (3) Typical battery life, (4) Adjusting volume control, (5) operation of remote control |
| **B. Wear Schedule**: (1) Importance of full-time use, (2) Recommended VC setting and program, (3) For children, instruct parent to decrease processor VC to minimum, place on child's head, then increase to recommended setting, (4) Progressive programs, if provided, (5) Importance of monitoring health of incision and coil sites |
| **C. Listening Strategies**: (1) Recommend age-appropriate listening strategies, (2) Creating ideal listening environment |
| **D. Avoid Processor Damage**: (1) Avoid excessive moisture and sources of static electricity |
| **E. Misc**: (1) Processor wearing/retention options, (2) Contact information; questions/problems |
| **Activation Appointment: Session 2** |
| **A. Discuss Experiences**: Counsel about reported experiences |
| **B. Basic Processor Operation:** Reiterate basic operation of processor-VC, battery life, etc. |
| **C. Basic Care, Use, and Maintenance**: (1) Use of dehumidifier; (2) Microphone care, (3) Integrity of various components/reducing strain on and replacing cables, (4) Warranty policies |
| **D. Basic Troubleshooting and Contraindications**: (1) MRI precautions, (2) Avoid monopolar cautery, (3) Avoid contact sports with risk of injury and scuba diving below 100 meters |
| **1-Week Session** |
| **A. Discuss Experiences**: Counsel about reported experiences |
| **B. Basic Processor Operation and Accessories**: (1) Inspect settings to ensure appropriate use, (2) Discuss basic operation as necessary, (3) For adults, discuss situation-specific programs, (4) Accessory cables-connecting to remote devices, (5) Monitoring earphones |
| **C. Bacterial Meningitis Vaccination:** Importance of vaccination with recommended schedule |
| **D. Miscellaneous**: (1) Rehabilitative/habilitative guidance, (2) Care of processor in extreme conditions (e.g., sports, exercise, working outside, toddlers) |
| **1-Month Session** |
| **A. Discuss Experiences**: Counsel about reported experiences |
| **B. Basic Processor Functions**: (1) Inspect settings to ensure appropriate use, (2) Discuss basic operation as necessary, (3) Importance of functional backup equipment loaded with current programs, bring to every subsequent appointment, (4) Protocol for replacing faulty equipment |
| **C. Removing Barriers**: (1) Hearing Assistance Technology, (2) Ensure adequate telephone abilities, (3) Introduce FM system if need additional assistance in challenging listening situations |
| **D. Future Sessions**: Discuss schedule for future programming appointments |

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ent. Bilateral cochlear implant recipient appointments are scheduled for 90 to 120 minutes. The week-long experience with the cochlear implant will (1) allow the recipient to gather experiences with sound from the implant so that they may provide useful information to the clinician to guide programming changes and (2) allow adults and older children to be more effective listeners when behavioral responses are required. Furthermore, recipients usually are able to tolerate higher

levels of stimulation and objective measures after they have used the implant for a short time (i.e., ESRT, ECAP).

At this appointment, the clinician should begin with otoscopy to rule out any external or middle ear pathology and an inspection of the skin flap and incision. After that, for adult recipients, frequency-specific, sound-field detection thresholds are measured to ensure adequate access to low-level sounds (i.e., ≤30 dB HL warble tones from 250 to 6000 Hz). When responses fall outside of the expected range, the clinician should make adjustments to improve access to low-level sounds. In most cases, open-set speech recognition is not measured at the 1-week appointment unless the patient is doing very well. Audiometric sound-field assessment typically is not performed for young children (<7 years) at the 1-week post-activation appointment given their limited attention spans. Instead, the clinician should focus on optimization of the program. When the physical examination is complete, electrode impedances should be evaluated. Impedance values likely will decrease substantially from the first day of activation and will begin to stabilize.

Next, the clinician should measure T levels on as many channels as possible to adequately characterize the user's electrical threshold profile across the electrode array. The anecdotal experience of the authors of this book suggests that measurement of T levels for children with behavioral observation audiometry (BOA) or visual reinforcement audiometry (VRA) may be more difficult at the 1-week checkup because the child has had experience with a variety of sound throughout the week, and as a result, the programming stimulus is no longer novel. Therefore, it may be challenging to keep young children engaged during this programming session.

When the T-level measurements are complete, the clinician should optimize upper-stimulation levels. This should first be completed through the use of behavioral measures with adults and older children (see Chapter 3 for a detailed discussion pertaining to the determination of upper-stimulation levels via the use of behavioral measures). The importance of loudness balancing and ensuring the upper-stimulation levels are set at an overall level that maximizes speech recognition and sound quality must be reiterated here because of the importance that each possesses in the outcome the recipient will experience.

For younger children (7 years old and below), the clinician should attempt to measure the ESRT to guide in the determination of upper-stimulation levels. When ESRT cannot be measured for young children, upper-stimulation levels should be based on a combination of behavioral measures, the clinician's observations of the child's reaction to stimulation, and the clinician's knowledge of what typically constitutes appropriate stimulation levels with the implant the child is using. Although the ECAP measurement is not strongly correlated to upper-stimulation levels, the clinician may use it as another piece of information to guide in determination of appropriate upper-stimulation levels (again, the reader is referred to Chapter 3 for a detailed discussion on the determination of stimulation levels for young children). Furthermore, the ESRT measure should also be attempted with adult recipients, even when the clinician believes the behavioral measures are valid because the ESRT serves as an excellent cross-check to confirm the appropriateness of upper-stimulation levels. Additionally, the ECAP should be measured for electrodes that were not completed at previous sessions, so the clinician may establish a record of the recipient's physiologic responses to electrical stimulation across the electrode array.

If an adult seems to understand the basic function of the sound processor and tolerates appropriate stimulation levels, the clinician may introduce situation-specific programs (e.g., telephone, listening in noise, music) (see Chapters 4–6 for examples of environmentally specific programs that users might find to be beneficial with each cochlear implant system). However, for young children who cannot switch settings, it is ideal to provide one program that is suitable across a variety of environments. If the caregiver reports that the child is still not tolerating appropriate stimulation levels, it may be necessary to continue providing multiple programs with increasing levels of stimulation. If the child is showing no aversion to appropriate stimulation levels and is able to initiate implant use at the recommended volume

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control setting, then the clinician should reduce the range over which the volume control operates. This will result in more precise control of volume adjustment for the child or parent.

If time permits, the clinician may demonstrate the use of accessory items (e.g., personal audio cables, monitoring earphones) and review and expand on the proper care of the system (i.e., use of the dehumidifier). In particular, the clinician should stress the importance of remote microphone technology (i.e., personal digital RF/FM system) in optimizing children's listening abilities and access to intelligible speech in real-world situations. The authors of this book firmly believe that these technologies should be used with even the youngest implant recipients. The reader is referred to Chapter 9 for additional discussion of the use of remote microphone technology with cochlear implants.

If at this point, appropriate stimulation levels still have not been determined for a child at the conclusion of the appointment, then the child should be scheduled for another programming session within a week. Otherwise, the user returns in 3 weeks for a 1-month, postactivation checkup. A summary of the procedures in typical 1- and 3-week appointments are provided in Table 7–3.

## One-Month Postactivation Programming Session

By the 1-month appointment, the recipient should begin to develop a strong bond with the cochlear implant. The recipient and family usually are able to provide substantial feedback about the recipient's progress and experiences with speech and other environmental sounds in order to guide the audiologist in programming changes. Most of the time, postlingually deafened adults will report significant improvements in their ability to understand speech. As a result, aided sound-field detection thresholds and speech recognition are evaluated at the beginning of this session. For users of bilateral implants, thresholds and speech recognition should be measured for each ear. These thresholds and scores contribute to necessary program changes. For example, if detection thresholds are elevated in the high-frequency range (e.g., >25 dB HL, 2000 to 6000 Hz), the clinician will focus on programming adjustments to improve access to high-frequency, low-level sounds.

At this point, it is imperative that children are using the cochlear implant during all waking hours. If the parent reports situations other than bath time or swimming in which the cochlear

**Table 7–3.** Overview of 1-Week and Optional 3-Week Postactivation Appointments

| *Time Postactivation* | *Duration* | *Basic Overview of Typical Procedures* |
|---|---|---|
| 1 week | 1 hour | • Visual examination of incision site, otoscopy, impedances<br>• Measure T levels, ESRT, and ECAP for remaining electrodes<br>• Adjustments to USL in live speech mode<br>• Child: provide one program<br>• Adult: aided sound-field thresholds; situation-specific programs<br>• Reduce volume control range<br>• Discuss care in greater detail and demonstrate accessories |
| 3-week interim | 1 hour | • Only necessary when T and USL levels not yet determined<br>• Visual examination of incision site and otoscopy<br>• Protocol similar to 1-week appointment |

*Note*. *ECAP* = electrically evoked compound action potential; *ESRT* = electrically evoked stapedius reflex threshold; *T* = threshold levels; *USL* = upper-stimulation levels.

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implant is not used, the clinician should counsel the family about the importance of full-time implant use. When available, the clinician should inspect the results of datalogging to ensure that full-time implant use is occurring. Children's sound-field detection thresholds should be evaluated at the beginning of the appointment and should be no poorer than 30 dB HL in the low frequencies and 25 dB HL in the mid to high frequencies. Programming changes are necessary if these goals are not met. Speech recognition may be evaluated for children older than 3 years of age who have previous hearing aid experience.

The remainder of the 1-month appointment involves the same measures conducted at previous appointments. Otoscopy and examination of the skin flap and incision site are followed by a visual inspection of the sound processor (i.e., signs of wear, tear, or damage). If the sound processor allows, the clinician should perform a biologic listening check with monitor earphones to evaluate the status of the sound processor microphone. Next, the clinician should perform impedance measures and determine T levels and upper-stimulation levels. Multiple T-level measurements at several channels will be necessary to determine profile changes across the electrode array. If T levels are relatively constant across the array, they will be measured on only a few electrodes and the remaining values will be interpolated. However, substantial T level variability warrants measurements on a greater number of channels. Next, upper-stimulation levels should be optimized using previously discussed procedures, and the ESRT should be measured. For adults, the volume control should allow for only small adjustments (e.g., –30%/+15% of the dynamic range). In children, the upper end of the volume control should not exceed the upper-stimulation levels set in the program, and the lower end of the volume control range is dependent upon the child's tolerance for stimulation. If no tolerance problems exist, the volume control should be disabled or should only allow for minimal decreases in volume (e.g., –20 percent of the dynamic range or less). However, if tolerance problems do exist, the volume control should allow for substantial decreases (i.e., 50 percent or greater of the dynamic range). If an adult subject is struggling substantially, it may be worthwhile to trial different signal coding strategies and/or stimulation rates in an effort to provide better sound quality and/or performance. The final component of the 1-month appointment is an introduction of situation-specific programs if not yet completed at the previous session.

## Remainder of Programming Schedule

After the 1-month appointment, programming schedules differ between adults and children but involve similar procedures and mirror those described for the 1-month session. A summary of the steps included in the 1- and 3-month postactivation appointments is shown in Table 7–4. For children younger than 7 years, 2- and 3-month postactivation appointments are scheduled to ensure optimal access to low-level sounds and appropriate program settings. Following the 3-month appointment, the child should return every 3 months for assessment of auditory skills, verification of cochlear implant function, and programming. In most cases, minimal programming changes are necessary at these appointments. Nonetheless, the appointments are very essential to ensure that the child's performance is satisfactory in the test booth and to evaluate the physical integrity of the implant site as well as the child's external and middle ear status. Furthermore, the clinician may ensure that the implant equipment is all functioning appropriately and that the implant is being used effectively during all waking hours. These appointments also provide the clinician with the opportunity to provide continual counsel to the family.

In addition, young children with cochlear implants should receive intensive audition-based therapy/habilitation to optimize auditory and spoken language development. The audiologist should collaborate with the Listening and Spoken Language Specialist (LSLS) or speech-language pathologist to ensure that the child's speech and language is developing appropriately. If concerns arise, more frequent audiologic evaluations and implant-programming sessions may be warranted. When the child reaches 7 years of age and has at least 2 years of cochlear implant experience, audiologic assessment and programming are conducted every 6 months.

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**Table 7–4.** Overview of 1-Month and 3-Month Postactivation Appointments

| *Time Postactivation* | *Duration* | *Basic Overview of Typical Procedures* |
|---|---|---|
| 1 mo | 1 hour | • Visual examination of incision site and otoscopy<br>• Feedback about progress and discuss duration of implant use<br>• Aided sound-field thresholds and speech recognition<br>• Visual exam and biologic listening check on processor<br>• Measure electrode impedances and T levels<br>• Optimize USL; measure ESRT<br>• Reduce volume control range or disable<br>• Provide situation-specific programs |
| 2 & 3 mo | 1 hour | • Visual examination of incision site and otoscopy<br>• Sound-field detection thresholds and speech recognition<br>• Similar procedures as 1-month session<br>• Child: Ensure access to soft sounds; determine if program levels and settings appropriate<br>• Adult: seen at 2 mo if struggling; otherwise, seen at 3 mo |

*Note*. *ESRT* = electrically evoked stapedius reflex threshold; *mo* = months; *T* = threshold levels; *USL* = upper-stimulation levels.

Following the 1-month postactivation appointment, adults typically return for a 3-month programming session. However, if an adult is struggling at the 1-month appointment, he or she should be seen for a 2-month postactivation appointment. Following the 3-month appointment, adults are seen every 3 months through the first year of implant use and on a biannual or annual basis thereafter. A summary of the appointments following the 3-month postactivation session is outlined in Table 7–5.

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## THE ROLE OF ELECTRICALLY EVOKED POTENTIALS IN PROGRAMMING

Several researchers have described the possible role of electrically evoked surface-measured potentials (i.e., measured from electrodes on the head: electrically evoked auditory brainstem response, electrically evoked middle latency response) in cochlear implant management (Brown, Abbas, Fryauf-Bertschy, Kelsay, & Gantz, 1994; Kileny, 2007; Sharma & Dorman, 2006), but for the most part, these measures are not used in routine clinical practice because, in short, these measures require additional, expensive equipment, considerable time, and clinical expertise to administer. Additionally, they require attention and cooperation on the part of the recipient, and there is no clear indication of how these measures may be used to improve the management of the recipient. Finally, the relationship between these measures and behavioral stimulation levels is generally considered to be modest at best.

There are, however, numerous clinicians who measure the electrically evoked compound action potential (ECAP) of the auditory nerve via specialized telemetry measures available in the implant hardware and software. A comprehensive discussion of the ECAP is beyond the scope of this book. Readers desiring more information on the ECAP are referred to another book in the Plural Publishing Inc. Cochlear Implant Core Clinical Concepts in Audiology Series entitled *Objective Measures in Cochlear Implants*, by Michelle L. Hughes. Dr.

**Table 7–5.** Appointment Schedule Following the 3-Month Postactivation Session

| *Time Postactivation* | *Duration* | *Basic Overview of Typical Procedures* |
|---|---|---|
| Every 3 months | 1 hour | • Visual examination of incision site and otoscopy<br>• Sound-field detection thresholds and speech recognition<br>• Visual exam and biologic listening check on processor<br>• Measure electrode impedances and ESRT<br>• Adjust T levels, USL, parameters, and programs<br>• Child: assess auditory skills; clinician collaborate with LSLS |
| Child: Every 6 months | 1 hour | • For child 7 years or older with 2 years of implant experience<br>• Visual examination of incision site and otoscopy<br>• Sound-field detection thresholds and speech recognition<br>• Procedures similar to 3-month session |
| Adult: Biannual or Annual | 1 hour | • For adult after 1 year of implant use; dependent on needs<br>• Visual examination of incision site and otoscopy<br>• Sound-field detection thresholds and speech recognition<br>• Perform procedures similar to 3-month session |

*Note*. *ESRT* = electrically evoked stapedius reflex threshold; *LSLS* = language and spoken language specialist; *T* = threshold levels; *USL* = upper-stimulation levels.

Hughes provides an excellent overview of the ECAP and every other objective measure that has been used to evaluate cochlear implant function. In this book, a short review of the ECAP is provided with a focus on the use of the ECAP as a guide to create programs for cochlear implant recipients.

The ECAP is measured by delivering biphasic electrical pulses to an intracochlear electrode contact in order to stimulate the auditory nerve and then by using a nearby intracochlear electrode contact to record the neural response. It is a fairly simple measure because no additional equipment is required (i.e., the measurement may be performed with the recipient's implant, the manufacturer's programming software, and either the recipient's sound processor or a special cable and coil provided by the manufacturer). Because the ECAP response originates from and is measured within the bony cochlea, it is essentially not susceptible to myogenic and other forms of electrical artifact that commonly complicate the measurement of evoked potentials. As a result, the ECAP may be measured while the recipient is awake and active. The authors of this book recommend that the clinician should consistently measure the ECAP in both pediatric and adult recipients because the measure is advantageous for several reasons including: (1) it indicates a stimulus level that definitely should be audible to the recipient, (2) it confirms the responsiveness of the auditory nerve to electrical stimulation, (3), it confirms device function, (4) it serves as an objective baseline of physiologic function to which subsequent measurements can be compared, and (5) although most researchers have suggested that the ECAP possesses a weak to moderate correlation to T- and upper-stimulation levels, it certainly may be used as a tool to guide the clinician in determining stimulation levels for recipients who cannot provide reliable feedback about the loudness of the signals they receive from their implant.

Early work examining the ECAP as a tool to determine T- and upper-stimulation levels suggested a weak relationship between the ECAP and behavioral stimulation levels (Brown et al., 2000; Franck & Norton, 2001). In an attempt to improve upon the potential of using the ECAP

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to predict cochlear implant stimulation levels, several researchers developed protocols, and the ECAP was combined with measured behavioral response levels at one or at most a few electrodes in an attempt to determine T- and upper-stimulation levels across the electrode array. For example, Hughes and colleagues (2000) proposed a method in which the ECAP threshold is measured on all active intracochlear electrode contacts. Next, behavioral T- and upper-stimulation levels were measured for one electrode contact (typically in the middle of the array), and an offset (i.e., difference value) was determined between the ECAP and the measured T- and upper-stimulation levels. Then, the offset measured at that electrode was used to estimate T levels (by subtracting the difference between the T and ECAP threshold levels at the measured electrode and then subtracting that offset value from the ECAP threshold on remaining electrode contacts) and upper-stimulation levels by adding the difference between the upper-stimulation and ECAP-threshold levels at the measured electrode and then adding that offset value to the ECAP threshold on remaining electrode contacts (see Figures 7–1, 7–2, and 7–3 for a visual example of Hughes method). Although

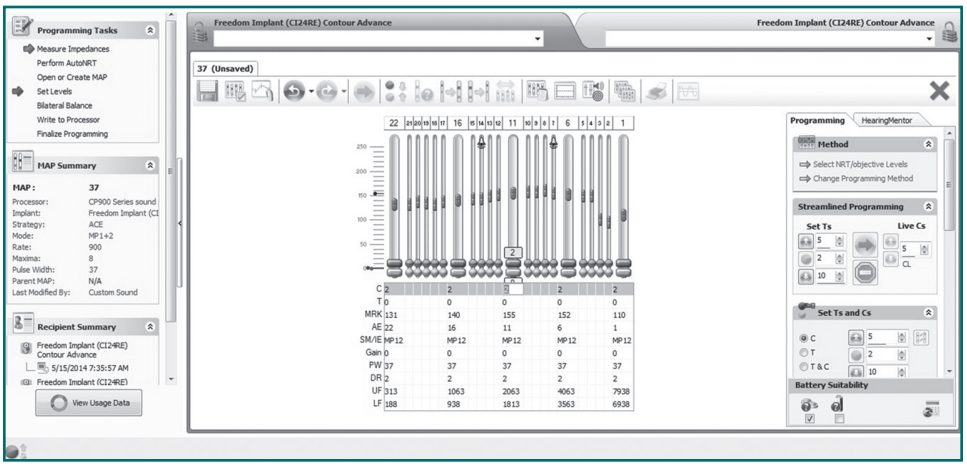
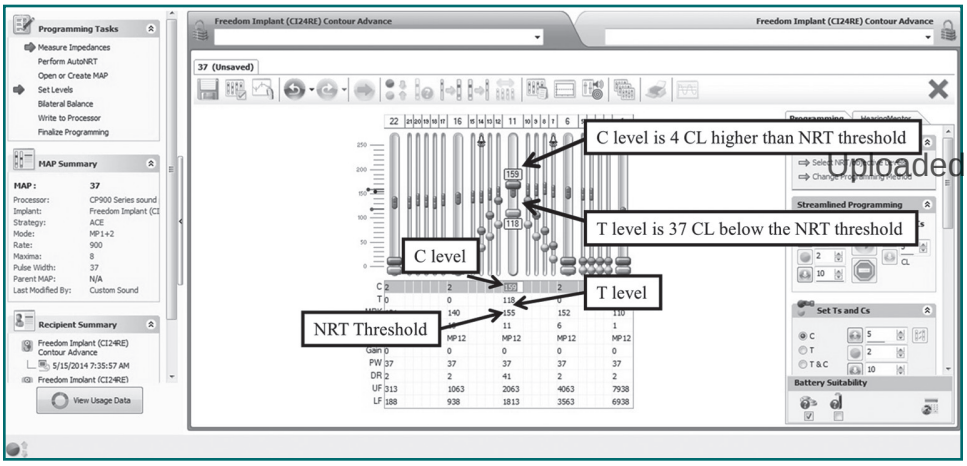



**FIGURE 7–1.** NRT thresholds measured across the electrode array. *Note*. Image courtesy of Cochlear Ltd.




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**FIGURE 7–2.** T and C levels that were measured behaviorally on one channel (channel 11). *Note*. Image courtesy of Cochlear Ltd.

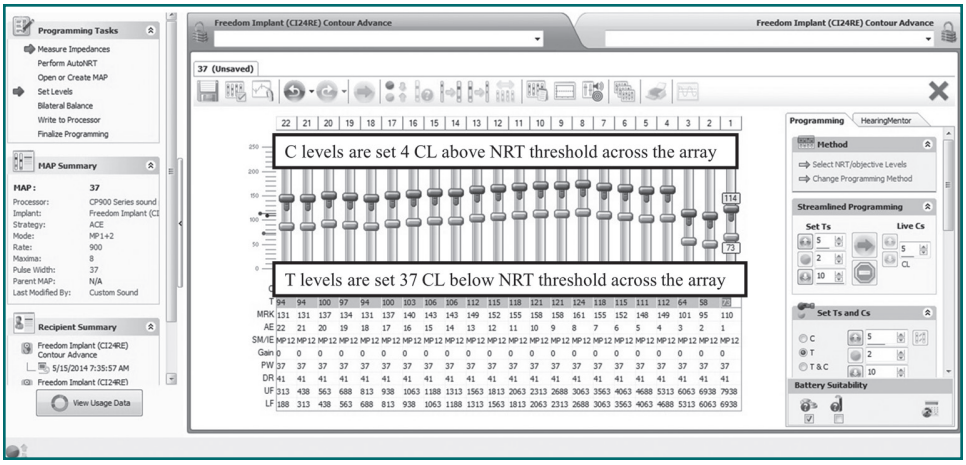



**FIGURE 7–3.** T and C levels were set across the array based on the offset of −37 and +4 CL, respectively. *Note*. Image courtesy of Cochlear Ltd.

the addition of measured behavioral levels on one electrode contact resulted in some improvement in the ability to use the ECAP threshold to estimate T- and upper-stimulation levels, research evaluating this method has suggested that the correlation between ECAP and behavioral stimulation levels was still weak to moderate at best (Franck, 2002; Hughes et al., 2000).

Smoorenburg, Willeboer, and van Dijk (2002) proposed another method in which behavioral measures were combined with the ECAP threshold in an attempt to estimate T- and upper-stimulation levels across the electrode array. These researchers also proposed that the clinician measure the ECAP threshold for every electrode contact within the array. Next, both T- and upper-stimulation levels were set to the same level as the ECAP threshold across the electrode array (i.e., match the ECAP threshold profile). Then, the T- and upper-stimulation levels were globally decreased to a level that was inaudible to the recipient. The clinician then activated this program in live speech mode, and the T- and upper-stimulation levels were globally increased while the clinician looked for a behavioral response to the stimulation. T levels were set to the point where a behavioral response to live speech was initially observed, and upper-stimulation levels were globally increased to a level that should provide adequate stimulation without discomfort (i.e., ideally at a level that offers optimal clarity for speech and that is most comfortably loud to the recipient). Of course, even with this approach, identifying optimal upper-stimulation levels is very subjective and difficult when working with recipients who are unable to provide reliable verbal feedback about the loudness of the signals they receive. Again, these researchers reported weak to moderate correlations between the ECAP threshold and behavioral stimulation levels with the use of this method (Smoorenburg et al., 2002).

In short, the majority of studies examining the ECAP as a tool to create cochlear implant programs levels have suggested that it quite limited in the ability to predict behavioral stimulation levels. It is difficult to pinpoint the exact cause of the weak relationship typically reported between ECAP and T/upper-stimulation levels, but three possibilities exist. First, there are potentially substantial inter- and intrarecipient differences in the geometrical orientation between the stimulating electrode and recording contacts across the electrode array. The geometric orientation between stimulating and recording electrodes and the auditory neural elements most likely have a big impact on the response recorded via a near-field approach, such as ECAP with cochlear implants. For instance, a recording electrode facing perpendicular to (directly toward) the auditory nerve most likely captures a more robust response than a recording electrode facing at a 45-degree angle to the neural elements. Second, subject differences

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in neural refractory properties may influence the relationship of the ECAP response measured to a slow rate (e.g., 30–80 pps) compared to a behavioral response measured to a pulse train with a stimulation rate of several hundred to several thousand pps. Differences in intersubject neural refractoriness of auditory nerve fibers may contribute to the significant intersubject variability observed between ECAP threshold and behavioral stimulation levels. These differences may be especially prominent in CI programs/MAPs using faster stimulation rates (e.g., several thousand pps). Third, there may be differences among electrode arrays and the impact they may have on the stimulus and recording characteristics of the ECAP. Given the manner in which these factors vary across recipients, a single correction factor is not feasible for estimating program levels from the ECAP.

Botros and Psarros (2010a, 2010b) recently proposed another reason for the relatively poor relationship reported between ECAP threshold and behavioral stimulation levels. These researchers contended that a limitation of these aforementioned studies was that they treated each electrode within the electrode array as single, independent data point. Instead, Botros and Psarros suggested that the overall profile (i.e., shape) of the T- and upper-stimulation levels across the array should influence how these levels are set. Specifically, they noted that the profiles of the T and upper-stimulation levels often take the same shape as the ECAP threshold profile, but the former are typically flatter than the latter. In particular, this is especially true for the profile of the upper-stimulation levels. In short, the profile of T levels often takes the same shape as the ECAP-threshold profile, and while the upper-stimulation level profile also mimics the ECAP-threshold profile, it tends to be flatter.

Botros and Psarros (2010a, 2010b) proposed that the flatter profiles at higher stimulation levels were due to a couple of factors. First, channel interaction is more prevalent at higher stimulation levels, which results in a greater similarity in the way the auditory nerve responds from one channel (e.g., electrode contact) to the next. Hence, the psychophysical responses are more similar at higher levels where channel interaction is more substantial. Second, more neurons are recruited by the electrical stimulation at higher levels, and the loudness growth functions are steeper. As a result, each increment change in stimulus intensity results in a larger change in loudness, which leads to fewer level differences between adjacent electrodes. Again, this results in a flatter profile for higher stimulation levels (which especially affects the upper-stimulation levels).

Botros and Psarros (2010a, 2010b) evaluated their hypothesis with Nucleus recipients. Using data collected from 15 Nucleus recipients, they developed a scaling factor to account for the flattening of the profile of stimulation levels associated with a given loudness percept that is particularly observed at higher stimulation levels. Then, they used this scaling factor to estimate the profile T- and upper-stimulation levels for 29 Nucleus recipients on the basis of the ECAP threshold and compared this estimation to estimates that were based solely on the ECAP threshold, on population estimates, and a flat profile. For each of the methods used to estimate T- and upper-stimulation levels, the profiles developed from the respective approaches were globally increased so that T levels were set to a level that was just noticeable, and upper-stimulation levels were set to a level that were loud but comfortable. They reported that the scaled ECAP estimates provided a significantly better estimate than any of the other methods used to estimate behavioral levels. Additionally, the majority of the subjects preferred programs created by the scaled ECAP approach relative to the other methods used to estimate behavioral stimulation levels. Botros, Banna, and Maruthurkkara (2013) further noted that the T- and upper-stimulation levels may be further optimized by tilting the profile (i.e., increasing/decreasing the emphasis of the low (bass) or high (treble) frequencies based on the recipient's feedback pertaining to the sound quality and clarity of speech and other environmental sounds. The interested read is referred to the Botros and Psarros (2010a, 2010b) articles for more information on conducting this procedure.

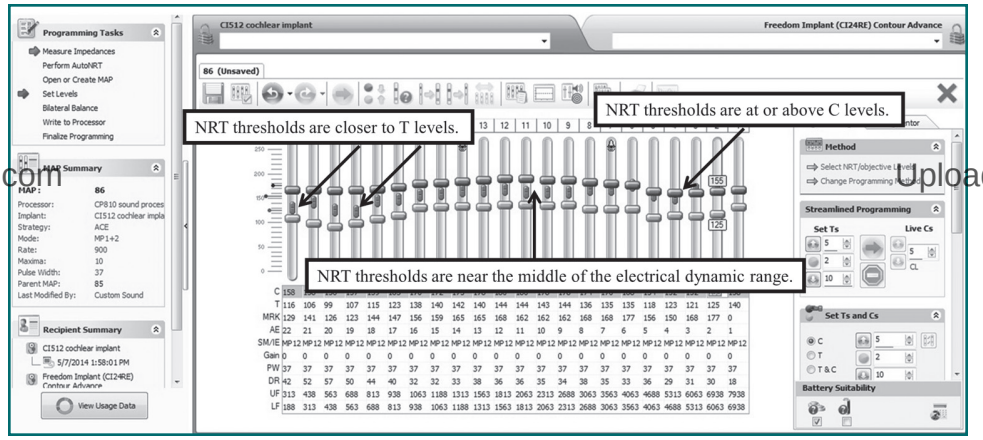
Although the Botros and Psarros (2010a, 2010b) approach of using scaled ECAP thresholds to estimate stimulation levels appears to hold promise, its clinical application is somewhat

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limited because of several factors. First, at the time of this writing, the scaling model that they used to shape the behavioral stimulation levels is not readily available. Second, even if the scaling model were readily available for clinical use, the clinician must still be able to obtain a behavioral threshold to speech or environmental sounds to set the global T-level profile and also determine the optimal level at which to set the global upper-stimulation level profile. Both of these tasks may prove to be elusive when creating cochlear implant programs for young and difficult-to-test recipients, who need objective measures the most. Third, the relationship between the ECAP and behavioral-stimulation levels is variable across recipients and may even be variable across the electrode array within an individual recipient. Subjects may have ECAP thresholds that fall near T levels, near upper-stimulation levels, or in the middle of their electrical dynamic range. Although the scaled ECAP approach may serve to account for some of this variability, it certainly does not capture it to the full extent.

For example, Figures 7–4 and 7–5 show NRT thresholds superimposed over program levels for reliable adults with excellent performance. As shown in Figure 7–4, the ECAP threshold falls near upper-stimulation levels at some segments of the array, within the middle of the EDR at other segments of the array, and above upper-stimulation levels for other channels. If the clinician was using any of the aforementioned ECAP-based approaches in isolation to set behavioral levels, then suboptimal stimulation levels would be provided for this recipient. The next two examples shown in Figure 7–5 show how the NRT threshold can approach the upper-stimulation levels (i.e., C levels) for one person and T levels for another. Given the variable relationship of ECAP to measured stimulation needs, the clinician should exercise caution when using the ECAP threshold to estimate T- and upper-stimulation levels. Behavioral and/or ESRT measures serve as better indicators of ideal program levels, and it is simply inappropriate to create cochlear implant programs from ECAP thresholds when reliable behavioral (or ESRT in the case of upper-stimulation levels) measures can be obtained. In short, the clinician should always strive to obtain valid behavioral levels to determine T- and upper-stimulation levels and also acquire the ESRT to corroborate upper-stimulation levels.

With that said, there are certainly cases in which the clinician will be unable to obtain behavioral and ESRT measures, and as a result will have to rely on the ECAP threshold. Examples of these situations include young children who exhibit inconsistent responses to auditory stimuli during the early stages of implant use as well as recipients with multiple disabilities that complicate the




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**FIGURE 7–4.** NRT thresholds superimposed over program levels for reliable adults with excellent performance. *Note*. Image courtesy of Cochlear Ltd.

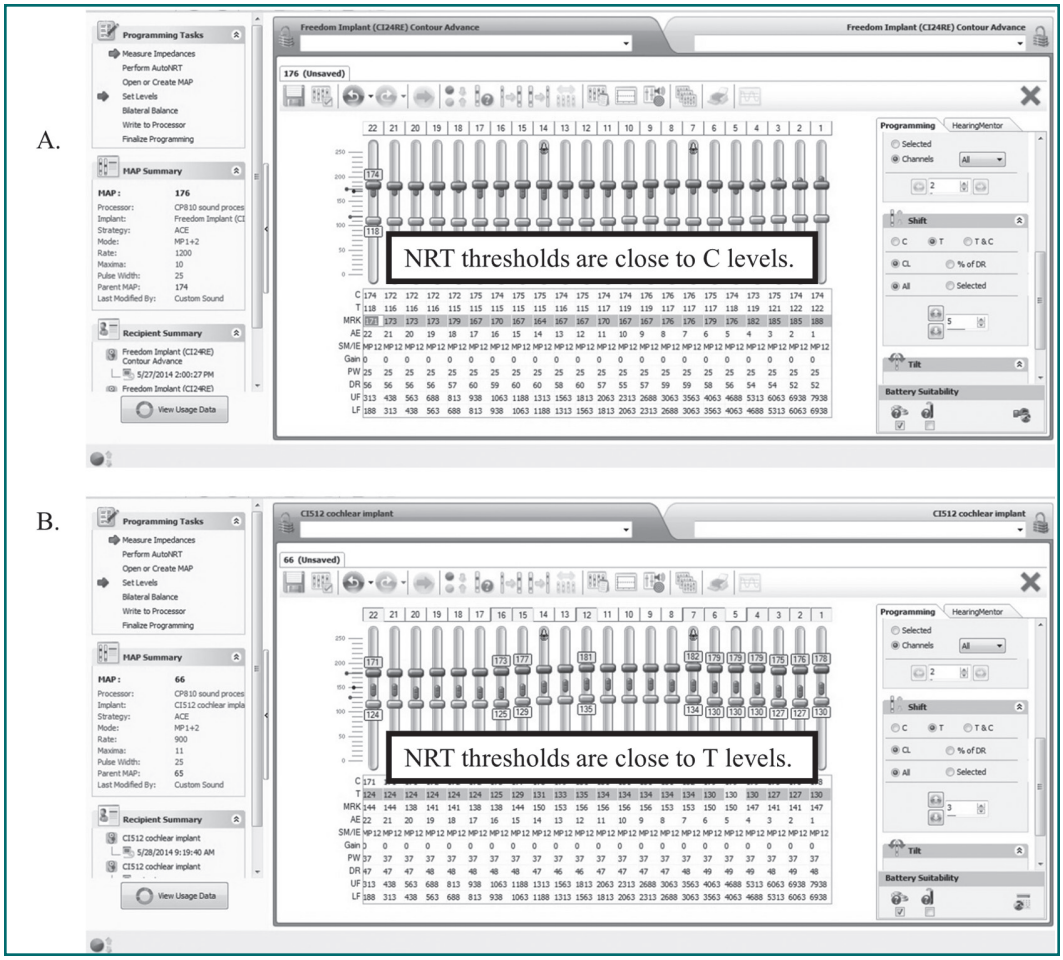



**FIGURE 7–5.** Examples of how the NRT threshold can approach the C levels for one person and T levels for another. *Note*. Image courtesy of Cochlear Ltd.

measurement of behavioral stimulation levels. In these cases, the authors of this book have adopted a modified approach that is largely influenced by the protocols proposed by both Smoorenburg et al. (2002) and Botros and Psarros (2010a, 2010b). First, the clinician should measure the ECAP threshold for all intracochlear electrode contacts across the array (Figure 7–6a). Second, the T- and upper-stimulation levels should be set closely to the profile of the ECAP thresholds (Figure 7–6b). It should be noted that when abrupt differences in ECAP thresholds exist across the array (i.e., ECAP thresholds that are markedly different at one or a small subset of electrodes relatively to the remaining or neighboring electrodes), the clinician should consider arbitrarily "smoothing" the T- and upper-stimulation-level profiles so that there are not significant fluctuations in the T- and upper-stimulation levels across the array (Figure 7–7a). Third, the T- and upper-stimulation-level profiles are offset by a small amount (e.g., 10 CL/10% of ECAP threshold—if ECAP threshold is 150 CU, then offset by 15 charge units), and both are globally decreased to a level that the clinician is certain will be inaudible to the recipient (Figure 7–7b). Fourth, the implant is activated in live speech mode, and the T- and upper-stimulation levels are gradually increased while the programming clinician and assistant observe for a behavioral response from the recipient.

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**FIGURE 7–6.** The authors' procedure for setting program levels: **a.** NRT thresholds measured across the electrode array; **b.** T and C levels set to the same profile as the NRT. *Note*. Image courtesy of Cochlear Ltd.

T levels are set at the minimal level that elicits a response (Figure 7–8a). Fifth, upper-stimulation levels are set to a level that the clinician believes will allow for adequate loudness normalization (i.e., loud sounds will be loud but comfortable to the recipient) and optimal speech recognition, while avoiding loudness discomfort. As upper-stimulation levels are increased, the clinician should arbitrarily flatten the upper-stimulation-level profile so that it continues to roughly approximate the ECAP threshold profile but is smoother or flatter in shape (Figure 7–8b). This adjustment is based on the Botros and Psarros (2010a, 2010b) work.

Again, as stimulation levels approach higher current levels, it is likely more appropriate to provide a flatter upper-stimulation-level profile because the increased channel interaction associated with higher stimulation levels will lead to more similar requisite levels (i.e., flatter shape) from one electrode to the next. The clinician should use his or her knowledge about typical stimulation levels of a system to serve as a guide for estimating upper-stimulation levels. For example, most Nucleus recipients have an electrical dynamic range approaching 40 to 50 CL. As a result, an appropriate goal would be to set upper-stimulation levels so that the electrical dynamic range is

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**FIGURE 7–7.** The authors' procedure for setting program levels, continued: **a.** T and C levels "arbitrarily smoothed" to account for abrupt changes that exist in the NRT thresholds from one channel to the next; **b.** T and C levels are set 10 CL apart and globally decreased to a level that is inaudible so the recipient will not hear stimulation when processor is activated. *Note*. Image courtesy of Cochlear Ltd.

approximately 40 CL. The clinician should be reticent about adjusting upper-stimulation levels to a point that is markedly higher than what is necessary for the typical recipient with a given make and model of implant/electrode array although there may be some exceptional cases in which it is appropriate to do so. Of course, the clinician should avoid setting upper-stimulation levels to a point that elicits aversive responses (e.g., eye blinking, anxiety, discomfort, tendency to remove the device). The reader is referred to Chapter 3 for more guidance on setting upper-stimulation levels for young children and adults. Admittedly, the aforementioned protocol requires a certain degree of subjective judgment on the part of the clinician, which is why the authors recommend the use of behavioral measures and the ESRT to determine cochlear implant program levels. However, in instances in which behavioral levels and the ESRT cannot be measured, the authors have found the modified version of the Smoorenburg et al. (2002) and Botros/Psarros (2010a, 2010b) approaches to be a reasonable substitute for estimating program levels.

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**FIGURE 7–8.** The authors' procedure for setting program levels, continued: **a.** T and C levels are increased until the recipient responds to the stimulation, and T levels are set to the level at which the recipient responds; **b.** C levels are globally increased to a level that provides sufficient stimulation without signs of discomfort or aversiveness, and the C-level profile is smoothed to account for the fact that upper-stimulation levels are often flatter than T levels. *Note*. Image courtesy of Cochlear Ltd.

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## ADDITIONAL HABILITATIVE/ REHABILITATIVE CONSIDERATIONS

Developing an ideal cochlear implant program with optimal stimulation levels is certainly important for success with a cochlear implant, but receiving specialized audition-based therapy or aural rehabilitation is, in many cases, equally important to promoting a successful outcome. This specialized therapy or rehabilitation facilitates maximal performance with the device for two reasons. First, the electrical signal from a cochlear implant is much different than the acoustic signal from a hearing aid. Second, auditory skills acquired with cochlear implants are very impressive but still do not match that of a person with normal-hearing sensitivity. Therefore, to enhance satisfaction, performance and benefit, aural rehabilitation for adult recipients is recommended during the first few weeks or months of cochlear implant use. Auditory-Verbal therapy is recommended for children throughout the first few years of life and

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until it is established that the child's speech, language, and auditory abilities are consistent with their normal hearing peers who have a similar nonverbal IQ.

The rehabilitation program should be directed by a LSLS or by an audiologist with experience in aural rehabilitation for adults with severe-to-profound deafness. In addition, the cochlear implant manufacturers have developed very effective computer-based aural rehabilitation programs, which are available on their websites or compact disc (CD-ROM). These manufacturer-designed programs are often critical for adults without access to individualized aural therapy but also supplement rehabilitation of adults enrolled in formal programs. In cases in which more intensive home-based aural rehabilitation is needed, the clinician should consider evidence-based, computer facilitated programs, such as LACE, which have been designed to facilitate auditory-skill development in adults with hearing loss. Home-based aural rehabilitation programs and activities are particularly important for elderly recipients who live alone and who have limited social and occupational activity. These recipients often have sparse exposure to spoken language, and as a result, their auditory systems are not consistently subjected to the intelligible speech necessary to facilitate the auditory nervous system's acclimatization to the signal from the cochlear implant.

The programming audiologist also should be equipped with several rehabilitative tools to facilitate early success with the cochlear implant. For example, when the implant initially is activated in live speech mode, the user is asked to listen as the clinician says the days of the week or months of the year. The clinician stops on a given day or month and asks the user to say the following day/month. In this example, the recipient's auditory memory is used to support the auditory input. Most users can accomplish this exercise, which helps them associate meaning to the sound and builds their confidence with the cochlear implant. It also serves as an excellent confidence builder because it allows the recipient to realize the potential benefit of the signal he or she is receiving from the new implant. Adults also are encouraged to seek out sounds during the first few weeks of cochlear implant use. For example, they can turn on the water faucet to associate meaning to the sound it generates. Recipients also should seek out sounds to which they cannot associate meaning. If they cannot identify a sound, family members or friends should be asked to assist them in identifying its source. The interested reader is referred to Table 3–1 for examples of aural rehabilitation examples the clinician can do with new CI recipients.

Auditory/visual tracking is another effective rehabilitation activity in the early stages of implant use. In this exercise, the recipient listens and follows along visually as the clinician reads a book, magazine, or newspaper aloud. The clinician stops reading somewhere within the passage, and the recipient is asked to say the next word in the passage. Obviously, recipients may rely on visual input to support the auditory signal from the cochlear implant, but this exercise will help them associate specific words to novel sounds. Family members should be encouraged to conduct this tracking exercise with the recipient at home. If the recipient lives alone or is unable (or does not want) to practice tracking with a family member, then he or she may still engage in the tracking exercise by obtaining an unabridged version of an audio book recording (compact disc, MP3, etc.) as well as the paper version of the book. The recipient can follow along visually in the book while listening to the audio recording. Many public libraries carry unabridged audio copies of the paper books on their shelves.

Familiar phrase cards are also beneficial during initial stages of implant use. Two to four cards are placed on the table, and the clinician reads the phrase or words on one of the cards aloud. Recipients are asked to select the card that was read. This task can begin at the level of pattern perception (i.e., cards having words with different syllable length) and progress to closed-set word identification. Although the previously mentioned exercises seem somewhat elementary, they help to build confidence and auditory-skill level at the initial activation sessions. The aforementioned exercises do not represent a comprehensive aural rehabilitation program for the adult patient. Again, if the programming clinician is not well-versed in audiologic rehabilitation, he or she

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should consider referral services from a specialist experienced with working with cochlear implant users, particularly when the recipient does not experience rapid success with cochlear implant.

Audition-based habilitation/rehabilitation is absolutely critical for young children with a cochlear implant. Although the programming audiologist can provide support, all children should be managed closely by an LSLS who will guide the child and family toward optimal development of speech, language, and auditory skills. As previously stated, the audiologist should correspond closely with the LSLS to ensure the child's progress. At the initial activation appointment, the programming audiologist is responsible for ensuring that the child's caregivers have a thorough understanding of the use, care, and maintenance of the cochlear implant system as well as the importance of using it during all waking hours. Oral and written instruction should be provided at the initial session, and the audiologist should revisit the basics of implant use at subsequent appointments. In addition, the audiologist and family should determine the most appropriate wearing option for retention of the sound processor to the child's head or body.

On the first day of cochlear implant use, families should be encouraged to walk around the home and point out the different sounds they encounter. For instance, they can knock on the door, point to their ear, and say, "I heard that." The LSLS often refers to this activity as a "listening walk," which helps children associate meaning to new sounds. Other examples of sounds on a listening walk include ringing from the telephone, running water, and sounds from toys. If the child appears to respond to speech or an environmental sound, the caregiver should provide enthusiastic positive reinforcement, while pointing to her own ear and saying, "I heard that!" Parents should also be encouraged to provide an oral description of even the most mundane events of life. For example, when walking into a room, the parent should say, "Okay, we're walking in the room, and now we're going to turn on the light. Oh, the light came on." Consistent descriptions of daily activities provide the child with a rich model of speech and language development. Another wonderful exercise to recommend for auditory skill development in young children is the simple act of reading books. Reading aloud to children serves as an excellent model for language and literacy development.

Finally, caregivers should attempt to optimize the acoustics in the home. For example, if no one is watching the television, it should be turned off to reduce competing noise. Also, to provide a good signal-to-noise ratio during the early stages of implant use, the parent should attempt to remain in close proximity to the child and direct speech to the side of the implant when communicating. Obviously, the aforementioned suggestions are not exhaustive and only introduce a few ideas for building an "auditory lifestyle." For more information on Auditory-Verbal therapy for children with cochlear implants, the interested reader is referred to *Auditory Verbal Therapy: Teaching Spoken Language Through Listening*, which is a companion of this book in the Plural Publishing, Inc. Cochlear Implant Core Clinical Concepts in Audiology Series.

## CONSIDERATIONS FOR BILATERAL COCHLEAR IMPLANTATION

The basic fundamentals of cochlear implant programming also apply to the programming of two cochlear implants; however, a few special considerations must be taken into account. First, bilateral cochlear implant programming appointments are time intensive. When recipients receive two cochlear implants in the same surgical procedure (i.e., simultaneous implantation), the initial activation appointments should be increased to 3 hours for the first day and 2 hours for the second day. All subsequent appointments should be scheduled for 2 hours. Most adults are able to complete both the evaluation of implant function (e.g., booth testing) and cochlear implant programming during one full session. Children under the age of 12, however, often have a difficult time attending to the behavioral tasks for a 2-hour session. As a result, two appointments on separate days may be necessary, one hour-long session for behav-

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ioral testing and one 60-to-90-minute session for programming. For families who travel a great distance to the programming center, the assessment and programming sessions may be scheduled for the morning and afternoon, respectively. It is very important that the audiologist have a variety of entertaining games and toys available to facilitate conditioned play audiometry and cooperation during objective measures.

Determination of cochlear implant stimulation levels and other adjustable parameters are similar to the practices used for unilateral recipients. The clinician should try to match adjustable parameters, such as stimulation rate, input dynamic range (IDR), and frequency allocation. However, published studies and anecdotal experience suggest that most recipients receive bilateral benefit, even with interaural differences in cochlear implant program parameters (Dorman & Dahlstrom, 2004).

Balancing loudness between the two ears is intuitively appealing, but there are no standardized protocols for how this should be accomplished. Most clinicians query the recipient about the loudness difference between the ears while listening in live speech mode and make global adjustments to upper-stimulation levels to achieve a balanced loudness percept between ears. Tyler et al. (2006) describes an alternative approach in which the recipient listens to external, frequency-specific signals (e.g., warble tones or narrowband noise at 500 and 2000 Hz) presented from 90 and 270 degrees in the horizontal azimuth and reports whether they are matched in loudness or if the right- or left-sided signal is louder. Programming adjustments are made to achieve equal loudness for the signals at each frequency. More research is needed to determine the most effective strategy to optimize bilateral stimulation levels.

Many recipients experience bilateral summation with the use of two cochlear implants (Schafer, Amlani, Seibold, & Shattuck, 2007). However, no published studies indicate the standard magnitude of this summation or how it should be managed in the programming process. In most cases, bilateral summation can be addressed by providing a small global decrease in upper-stimulation levels for each ear relative to those determined in the unilateral conditions. If the decrease results in a significantly reduced electrical dynamic range in one ear, larger decreases to upper-stimulation levels may be necessary for the ear with the wider dynamic range.

Following the 1-month postactivation appointment, sound-field detection thresholds should be evaluated for each ear separately at the beginning of each session to confirm access to low-level sounds. Because the recipient likely wears both cochlear implants during all waking hours, it may not be apparent that audibility is insufficient for one ear. Speech recognition in quiet and in noise also should be assessed for each ear separately and in the bilateral condition. These conditions will confirm bilateral benefit, and more importantly, rule out the very rare possibility of bilateral interference.

No consensus exists regarding the most appropriate wearing schedule for people who receive two cochlear implants in separate surgical procedures (i.e., sequential implantation). Clinicians may recommend full-time use of both implants, periods of use with the new cochlear implant alone, or only the newly implanted ear for several weeks. More research is necessary to determine the ideal schedule of implant use for recipients with varying preimplant histories.

Finally, many cochlear implant recipients wear a hearing aid on the opposite ear (i.e., bimodal arrangement). Once again, it is desirable to balance loudness between the two ears. Information on optimizing the signal of each device is described by Ching and colleagues (Ching, Incerti, & Hill, 2004; Ching, Psarros, Hill, Dillon, & Incerti, 2001).

It is worth noting that insurance reimbursement for bilateral cochlear implant assessment and programming is a pressing issue in our field. Many insurance companies will only reimburse for assessment and programming of one cochlear implant. This policy adversely impacts cochlear implant programming centers that attempt to complete all services for both ears on one day because these services are more time intensive for bilateral recipients, but reimbursement is the same as what is obtained for a unilateral recipient. Many clinics have adopted a policy in which the recipient must attend the center in two separate days so

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that programming may be completed on each ear separately in order to maximize reimbursement. Indeed, these policies are often necessary to prevent the center from absorbing a financial loss. Unfortunately, this is one of many examples in which health care economics adversely impacts patient outcome. Most recipients are better served when both implant are programmed on the same day so that they implants may be programmed to work together to promote optimal bilateral function.

## KEY CONCEPTS

Following this chapter, the reader should recognize concepts associated with the following topics:

- The clinician must consult with the surgeon about the pre- and postimplantation imaging results and other medical needs.
- The cochlear implant activation appointment should be scheduled on two separate days to complete the necessary measurements and assessments.
- The 1-week and 1-month postactivation appointments should be followed by different appointment schedules for adults and for children.
- Although behavioral stimulation levels typically facilitate excellent performance when they are measured in a reliable and valid manner, the clinician should also use objective measures as a cross-check to behavioral cochlear implant program levels.
- In the cases of young children and other recipients who cannot provide reliable feedback about the signals they receive from their cochlear implant, the clinician may have to rely on a combination of objective and behavioral measures to create a functional cochlear implant program. However, the ECAP threshold should never be used as the primary tool to determine stimulation levels when valid behavioral levels (and in the case of upper-stimulation levels, the ESRT) can be obtained.
- Habilitation/rehabilitation is necessary for adults and children to achieve their full potentials with the cochlear implant.
- There are several specific considerations for programming bilateral cochlear implants including the need for increased programming time and balance of adjustable parameters and loudness levels between ears.

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# 8

# Patient Complaints and Complications

Advances in cochlear implant technology and clinical procedures, expansion of candidacy, and earlier identification of hearing loss in children have resulted in considerable benefit for the overwhelming majority of recipients. In some instances, however, children with cochlear implants do not develop age-appropriate speech and language skills, and adults do not achieve excellent open-set speech recognition. Although these undesirable outcomes may be attributed to factors outside of the clinician's control (e.g., low nonverbal IQ, additional disabilities, poor familial support, abnormal anatomy), it is also possible that poor outcomes may be related to a suboptimal program/MAP that fails to meet the recipient's needs. The interested reader is referred to several publications that highlight factors associated with outcomes for children and adults with cochlear implants (Ching & Dillon, 2013; Geers, Strube, Tobey, & Moog, 2011; Holden et al., 2013).

Some recipients experience initial difficulties with their cochlear implants, but after adept management by the programming clinician, they achieve substantial benefit. It is fairly common for even the most successful recipients to experience temporary setbacks, which may be attributed to faulty external equipment or minor programming needs. These issues typically are addressed efficiently and effectively by the seasoned programming clinician. Chapter 11 provides several examples in which substantial programming changes resulted in a significant improvement in implant recipient outcomes. The goal of this chapter is to discuss the audiologic management of recipients who experience suboptimal outcomes, complications, and temporary and chronic difficulties. Suggestions also are provided for the management of difficult-to-program recipients.

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## MANAGING RECIPIENTS WHO EXPERIENCE DISAPPOINTING OUTCOMES

Numerous factors may be responsible for deficient performance with a cochlear implant, but they most often are related to patient characteristics. In other cases, unsatisfactory performance or temporary difficulty may be attributed to an extrinsic factor (e.g., faulty hardware, poor program). Therefore, the following potential extrinsic causes should be evaluated.

### External Hardware

Although the reliability and durability of external hardware is exceptionally good, malfunction of the external sound processor and its components

(e.g., cable, transmitting coil) should always be considered as a potential cause for poor or limited performance. Cochlear implant recipients typically use their sound processors during all waking hours, and as a result, the external hardware is frequently subjected to adverse elements, such as heat, body oil, moisture (from perspiration, humidity, rain, bathing, swimming pools), and being dropped on hard surfaces. The clinician should routinely conduct an examination of the function of a recipient's external hardware. This should take place at least every 3 months for children and one to two times a year for adults. The examination should include the following components.

First, a visual examination of the external hardware should be conducted to check for signs of wear or faults. It is especially important to inspect the transmitting cable(s) for evidence of anomalies as the wires inside the cable often are susceptible to breakage. In fact, the transmitting cable is typically the weakest of all of the external components, and as a result, when recipients present with no signal or an intermittent signal from their implant system, the most likely cause is a faulty transmitting cable. When the function of the transmitting cables is in doubt, the clinician should switch to a clinic loaner cable or the recipient's backup cable, or if troubleshooting over the telephone, then the recipient should be directed to switch to his or her backup cable. For young children who cannot report on the integrity of the signal, the transmitting cables should be replaced every 6 months.

Next, the clinician visually should assess the appearance of the sound processor's microphone port(s) or covers. Dirty microphone ports should be cleaned. If it is impossible to clean the microphone ports, then the clinician should have the processor replaced. Some sound processors feature covers that protect the microphone from moisture, debris, and wear and tear (e.g., Cochlear Nucleus 5 and Nucleus 6 microphone covers, Advanced Bionics T-Mic 2 cover). These covers should be clean and free of debris and should also be periodically replaced. For active recipients and those living in humid/tropical environments, the covers should be replaced every 2 to 3 months. For the typical recipient, the covers should be replaced every 4 to 6 months, and for inactive recipients or those living in arid or relatively dry environments, every 6 months.

When possible, the clinician should also assess the microphone function with a listening check (i.e., Cochlear monitor earphones/Advanced Bionics listening check/MED-EL microphone test device). Additionally, as discussed in previous chapters, it is imperative to measure sound-field detection thresholds to ensure that the sensitivity of the microphone allows for access to low-level sounds across the speech-frequency range (i.e., 250–6000 Hz). In the initial stages of microphone deterioration, sound-field thresholds usually increase in the high frequencies.

The clinician then should confirm that the transmitting coil is sending and receiving signals to the internal device using a signal-check wand, a diagnostic remote control (e.g., CR110), or by performing telemetry measure via the computer programming software interface. Then, the clinician should inspect the sound processor batteries and battery contacts for signs of corrosion or faults. Furthermore, the clinician should verify that all controls and indicators are functioning properly. Finally, most recipients own a backup sound processor for troubleshooting purposes, and in the event that performance and progress are poor, the clinician should always remember to ask the recipient to switch to the backup external equipment. If the problem is addressed, the clinician should replace systematically each of the sound processor components to identify the faulty link. If the recipient does not have a backup processor, then the clinic's loaner equipment should be used for troubleshooting.

In regards to ensuring optimal function of external hardware, the clinician and recipient will both benefit from the old adage that suggests that "an ounce of prevention is worth a pound of cure." As previously suggested, the recipient should be counseled to routinely replace the microphone covers on schedule as recommended by the clinician. Also, the clinician should counsel the recipient about the frailty of the transmitting cable to encourage the recipient to exercise caution when removing the coil from the head (i.e., do not pull the processor off of the head by grabbing

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the coil). The recipient should be provided with a thorough understanding of the water resistance of his or her sound processor as well as suggestions for facilitating reliable sound processor function when in the presence of water. Regardless of the water resistance rating of the sound processor, the recipient should be counseled to place the sound processor in a dehumidifier every night. Routine use of a dehumidifier will markedly improve the reliability and function of sound processors, even for those that are waterproof. Furthermore, clinicians should ask the recipient to hold the sound processor a short distance over a soft surface (if possible) when the processor is being handled (e.g., changing batteries, adjusting controls). This will reduce the likelihood that the processor falls from a high distance onto a hard surface. Finally, the clinician should assist the recipient in identifying the necessary means for ensuring that the processor securely adheres the recipient's ear/head. Most older children and adults can simply wear the processor on the ear without concern of it falling off. However, active recipients (e.g., sports, manual labor) may need assistance in identifying accessories (e.g., Snugfit, earmold, double-sided tape, Ear Gear, retention cords) or special wearing configurations (e.g., body-worn configurations) that will facilitate adequate retention during periods of strenuous activity.

## Determine Wear Schedule and Facilitate an Auditory Lifestyle

To achieve their full potential with a cochlear implant, young children must use the device during all waking hours, and the family must provide an enriching model for speech, language, and auditory development. Children in families that enforce only sporadic use of the implant achieve fair progress at best. In addition, children in families that do not stress audition and have limited expectations for spoken language typically will lag far behind the development of their peers with normal hearing.

When inquiring about a wear schedule and the communication lifestyle at home, questions should be worded carefully to elicit a forthright response. If one asks whether the child wears the cochlear implant during all waking hours, the caregiver likely will answer affirmatively, as the question implies that the objective is full-time use. In contrast, an inquiry about the number of times or hours the child wears the implant or "takes a break" from the implant is likely to yield a more accurate estimate from the caregiver. The clinician should gently and tactfully reinforce the importance of full-time cochlear implant use. Some sound processors, such as the Naída CI Q70 and the Nucleus 6, feature datalogging that indicates the number of hours the sound processor is used per day as well as the number of hours the child is exposed to speech. This information can be critically important when a young child is making undesirable progress. The clinician must have a frank discussion with the child's caregivers to strongly recommend implant use during all waking hours and also to provide a sufficient model of intelligible speech. When a family decides that the child will use spoken language as his or her mode of communication (either by way of an Auditory Oral, Auditory Verbal, or Total Communication approach) anything short of implant use during all waking hours is neglect of the child's development.

Likewise, the clinician should inquire about the number and types of visual supports used to assist communication. If the family relies heavily on visual input, the clinician may provide some suggestions for supporting communication through audition-based strategies. Finally, the clinician should correspond with the child's Listening and Spoken Language Specialist (LSLS) or speech-language pathologist to ensure the family is equipped with effective strategies to facilitate listening at home.

For the elderly adult who lives alone, it may be difficult to establish consistent social interaction with friends or loved ones. As previously mentioned, a new recipient requires a period of time to adjust to the new signal. If the user is unable to practice listening and have routine conversations, the progress will be slow and the adjustment of the central auditory system to novel signal from the implant will be delayed. For those with limited social interaction, the clinician should suggest activities or structured therapy to facilitate adjustment to the device.

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## Evaluate the Appropriateness of the Cochlear Implant Program

When the external equipment is functioning normally and the recipient is using the device properly, the clinician should reevaluate the appropriateness of the user's program. Electrode impedances should be within normal limits, have unremarkable morphology (e.g., no large variability in impedance between contacts), and be similar to previous measurements. Next, the clinician should confirm that threshold (T) levels and upper-stimulation levels are set accurately. Of all adjustable program parameters, the threshold and upper-stimulation levels will have the greatest influence on patient performance and sound quality. Again, the electrically evoked stapedial reflex threshold (ESRT) serves as the best indicator of appropriate upper-stimulation levels (see Chapters 3 and 11).

If the stimulation levels are set appropriately, the clinician may consider revising the signal coding strategy or adjustable program parameters, such as stimulation rate, maxima, and input dynamic range (see Chapters 3, 4, 5, 6, and 7). Finally, the clinician should obtain objective measures of auditory function (e.g., ECAP) and compare them to previous results to identify any unwanted changes. If degradation is observed, the clinician should schedule a medical evaluation and an integrity assessment of the internal device. As mentioned in Chapter 7, the clinician should also be concerned when upper-stimulation levels are substantially lower or higher than the recipient's ECAP threshold. When adjustments of the aforementioned parameters are successful, the programming audiologist, recipient, family, and ESLPs should notice considerable improvement in performance immediately, but definitely no later than 3 months.

## Assessment of Internal Hardware

When all other factors are ruled out as potential causes of poor progress, the clinician should scrutinize and assess the function of the internal device. All three manufacturers offer technical support in providing relatively thorough assessments of internal device integrity. Many faults are identified successfully with these assessments, but some recipients will have "normal" findings with a faulty internal device.

The clinician also may request that the otologist order another postoperative plain film x-ray to confirm that the internal device has not migrated or moved from its original position. This recommendation is especially warranted for sudden changes in stimulation levels or electrode impedance values. The otologist will determine whether shifts in internal device placement may be responsible for the negative changes in recipient performance.

Finally, recent evidence suggests that more advanced imaging procedures, such as computerized rotational tomography and 64-slice computerized tomography scan imaging, may provide valuable information about electrode array placement and the cochlea's condition after implantation. Specifically, research shows this imaging can be used to determine whether the electrode array has remained in the scala tympani or has dislocated to other chambers within the cochlea (Holden et al., 2013; Lane, Driscoll, Witte, Primak, & Lindell, 2007; Postnov et al., 2006). Research suggests that recipient performance is related inversely to the number of electrodes outside the scala tympani (Finley et al., 2008; Holden et al., 2013). This imaging also may be used to identify the presence of fibrous tissue growth or ossification after implantation.

# IDENTIFYING "RED FLAGS"

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Amy McConkey Robbins introduced the term "red flag" as an indicator of an implant recipient's slow or limited progress (Robbins, 2005). A red flag occurs when a certain skill lags behind the development of normal-hearing peers or below expectations for a given time period after activation. A red flag signifies the need for more proactive monitoring of pediatric recipient performance, but it is also applicable to adult recipients.

In order to identify a possible red flag, the clinician must be aware of typical progress with

a cochlear implant and normal speech and language development. The most desirable objective of cochlear implantation is to allow for communication via spoken language across a variety of listening environments, and this goal is achieved for most recipients. Most children who receive a cochlear implant at a relatively young age (≤3 years) and have normal cognitive and neurological function develop some speech, language, and auditory skills within 1 year of implant use. This goal also should apply to children who received considerable benefit from hearing aids and received a cochlear implant at a later age. When a child's progress does not meet this benchmark, the cochlear implant team should explore the reason(s) for the limited benefit.

Most adults with postlingual deafness and a limited duration of deafness (i.e., <20 years) develop good-to-excellent, open-set speech recognition and spoken-language skills. In fact, the vast majority of adults who use recent technology and obtain regular audiology services should be able to converse over the phone and achieve ceiling performance (i.e., 100% correct) on open-set sentence-recognition tests after 3 months of implant use. Certainly, the adult's performance should be significantly better than that obtained with hearing aids immediately prior to implantation. Once again, if these goals are not met, the cochlear implant team actively should explore the reasons for limited progress and suggest means for improved performance.

## Factors Influencing Cochlear Implant Outcomes

Outcomes from cochlear implantation are closely related to personal factors (e.g., preimplant hearing, otologic history, age, and family support) and intrapersonal characteristics (e.g., cognitive and neurological status, personal motivation, and use of communication strategies). In addition, previous research suggests variable outcomes for some recipients with additional disabilities, a history of bacterial meningitis, aplastic auditory nerves, inner ear malformations, long durations of deafness (i.e., >20 years and severe high-frequency hearing loss), advanced age, and prelingual deafness. Although the aforementioned patient characteristics may influence performance, they should not be considered as absolute contraindications for cochlear implantation. We have personal experience with recipients in each of these categories who still show remarkable benefit from a cochlear implant. Finally, performance certainly will be influenced if the integrity of the internal cochlear implant is compromised.

### Additional Disabilities

Approximately 40% of children with hearing loss have an additional disability. Many of these children do well with a cochlear implant, but a low nonverbal intelligence or depressed neurological or cognitive status may have deleterious effects on the development of spoken-language skills. Examples of conditions that are known to limit spoken-language development in children are below average intelligence, mental retardation, autism, and neurological effects associated with cytomegalovirus (CMV). The extent to which these conditions impact cochlear implant outcomes is directly related to the severity of the associated neurological and cognitive deficits. When additional disabilities are identified prior to surgery, it is important for the audiologist to discuss the potential for diminished outcomes following implantation. Even when a young child does not exhibit any obvious secondary disability, it is good practice to counsel families that speech and language outcomes often are impacted by neurological and cognitive status. Specific guarantees to parents regarding outcomes cannot and should not be made.

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### Bacterial Meningitis

Published reports indicate mixed cochlear implant outcomes for children who were deafened secondary to bacterial meningitis (Durisin, Arnoldner, Stover, Lenarz, & Lesinski-Schiedat, 2008; Mitchell, Psarros, Pegg, Rennie, & Gibson, 2000). Some of the children in these studies had similar speech and language outcomes as compared to the general pediatric cochlear implant population. For this reason, an etiology of bacterial meningitis should not prevent cochlear implantation.

However, other subjects had poorer outcomes than the typical pediatric population with cochlear implants, which likely are related to cochlear ossification or neurological deficits. A full insertion of the electrode array may not be achieved in recipients with significant ossification. First, ossification may result in significant degeneration of spiral ganglion dendrites and adversely may affect the delivery of electrical current from the electrode contacts to the auditory nerve. Second, it is well known that bacterial meningitis may cause significant neurological insults. When this occurs, language outcomes will be affected. Finally, for children with a history of bacterial meningitis, reduced outcomes may occur if duration of deafness before implantation is longer than 6 months (Durisin et al.). During this 6-month period, the potential for significant ossification is high; therefore, the surgeon may not achieve a full electrode array insertion. As a result, cochlear implantation should occur as soon as possible following deafness secondary to bacterial meningitis.

### Aplastic Auditory Nerves

An emerging body of literature suggests poorer outcomes for cochlear implant recipients who have absent or severely deficient auditory nerves as compared to those with other etiologies (Walton, Gibson, Sanli, & Prelog, 2008), and other research shows no differences in outcomes between groups (Jeong, Kim, Kim, Bae, & Kim, 2007; Peterson et al., 2003). This condition may be associated with a preimplant diagnosis of auditory neuropathy spectrum disorder (ANSD). Contemporary magnetic resonance imaging (MRI) can determine the status and integrity of the auditory nerve prior to implantation. This imaging is warranted for children with ANSD or limited responses on behavioral and objectives assessments of auditory function. Identification of a unilateral cochlear nerve deficiency would be critical for determining the most appropriate ear for implantation (Buchman et al., 2006). Although outcomes associated with cochlear nerve deficiency are typically poor, the family may choose to move forward with cochlear implantation. In this case, the family should be counseled that spoken-language outcomes may be poorer than expected, the implant may improve only sound awareness, and they should support a visually based communication mode (i.e., sign language). The electrically evoked transtympanic auditory brainstem response (ABR) assessment may serve as a prognostic indicator of the system's ability to respond to electrical stimulation in a synchronous manner (Gibson, Sanli, & Psarros, 2009; Kileny & Zwolan, 2004).

### Inner Ear Malformations

Unlike the previously mentioned conditions, children with inner ear malformations often have successful cochlear implant outcomes (Buchman et al., 2004; Loundon et al., 2005). For example, children with enlarged vestibular aqueduct syndrome typically perform as well or better than the general pediatric population with cochlear implants (Bent, Chute, & Parisier, 1999). Children with Mondini malformations also do well with cochlear implants, but the outcome generally is related to the extent of the malformation (Arnolder et al., 2004; Buchman et al.). Finally, many children who have common cavity deformities experience positive outcomes with cochlear implants (Buchman et al.), but they have more complications than children with less involved cochlear deformities (Mylanus, Rotteveel, & Leeuw, 2004). Because stimulation levels are more likely to fluctuate for this population, more frequent programming and assessment appointments are necessary (i.e., every 3 months).

### Prelingual Deafness and Duration of Deafness Prior to Implantation

Adults with prelingual deafness and limited listening and spoken-language experience often exhibit poorer outcomes when compared to adults with postlingual deafness. Numerous studies show limited open-set speech recognition and poorer spoken-language abilities for adults with prelingual deafness (Kaplan, Shipp, Chen, Ng, & Nedzelski, 2003; Teoh, Pisoni, & Miyamoto, 2004). In addition, some teenagers with prelingual deafness eventually may become nonusers because of limited benefit, cosmetic issues, or peer pressure. However, an implant almost certainly will improve detection of speech and environmental

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sounds, and speech perception will improve relative to preimplant scores (Santarelli, De Filippi, Genovese, & Arslan, 2008). Adults and teenagers in this category must be self-motivated to pursue a cochlear implant and have realistic expectations. If the potential candidate's family serves as the primary impetus toward cochlear implantation, the likelihood of a poor outcome and limited implant use is even greater. As such, the clinician should provide deliberate and truthful information regarding realistic expectations for this population in the candidate's preferred communication modality. If the user has realistic expectations and wishes to pursue a cochlear implant to enhance sound detection and support manual communication, the person may be considered for implantation.

In addition to limitations associated with prelingual deafness, several researchers report that durations of deafness exceeding 5 years may negatively influence cochlear implant outcomes (Green et al., 2007; Oh et al., 2003), but certainly exceptions to this rule can be found. As the duration of deafness begins to approach and exceed 20 years, it is more likely that the recipient's performance may be relatively limited. At any rate, an extended duration of deafness should not be a contraindication for cochlear implantation, but rather an indicator of the need for preimplant counseling regarding expectations. The authors also have encountered several recipients with prolonged high-frequency deafness who initially exhibit poor performance with their implant. This finding may be related to significant degeneration of spiral ganglion dendrites in the basal region of the cochlea or to deleterious changes in the central auditory system. A longer interval of implant use may be necessary before substantial benefit is attained in these recipients. For recipients who continue to struggle after several months of implant use, sound quality and speech recognition may be improved by disabling the most basal electrodes and reallocating the high-frequency inputs to more apical electrode contacts.

### Advanced Age

Several published reports describe the considerable benefits of cochlear implantation for elderly adults in speech recognition and quality of life (Migirov, Taitelbaum-Swead, Drendel, Hildesheimer, & Kronenberg, 2009; Noble, Tyler, Dunn, & Bhullar, 2009); therefore, an advanced age should never dissuade a cochlear implant team from recommending implantation. However, when compared to the general adult population with implants, some recipients who have advanced age will exhibit poorer performance (Noble et al.). Therefore, more extensive counseling regarding expectations and longer appointments may be necessary. The diminished outcomes for this population may be related to the duration of deafness, changes in central auditory processing, and the neurological or cognitive status.

### Internal Device Function

Although the vast majority of cochlear implant internal devices function properly, instances have been reported of compromised device integrity resulting in poor recipient performance. At some point, every manufacturer has produced an internal device with a widespread faulty component, design flaw, electrode array anomaly, or electrical defect. Clinicians should be aware of these potential issues and closely scrutinize the performance of recipients with these devices. Manufacturer representatives can provide information about the faults in past devices and the names of recipients who may have a potentially compromised device. In some cases, defects will result in complete failure of the internal device (i.e., hard device failure). In other instances, a compromised internal device may continue to provide sound for the recipient, but he or she may experience consistently poor speech recognition and/or sound quality, regression in speech recognition and/or sound quality, intermittent device function, and extraneous sounds (e.g., popping, static, crackling). The reliability data of contemporary internal devices is very high and may be found on each manufacturer's website (Advanced Bionics Corporation, 2013; Cochlear Corporation, 2013; MED-EL Corporation, 2013).

If a recipient experiences a limited outcome and has a device with a suspect reliability record, then the cochlear implant team should consider replacing the internal device with a new system

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(i.e., cochlear implant revision). Published reports and the authors' extensive anecdotal experience suggest that cochlear implant revisions frequently are successful when based on sound clinical judgment (Zeitler, Budenz, & Roland, 2009). After exhausting the extensive troubleshooting and assessments discussed in this chapter, cochlear implant revision should be considered for recipients who have as outcome that is significantly poorer than what would be expected for the preimplant history. Of course, the decision to revise a cochlear implant is much more difficult for recipients who have any of the aforementioned factors associated with limited outcomes. Prior to the decision, the cochlear implant team should consider the recipient's intrapersonal characteristics, his or her family support, the reliability of the internal device, and the integrity of the device according to assessments conducted by programming clinician or a manufacturer representative.

## Programming Adjustments for the Most Common Complaints and Complications

Prior to making programming adjustments to address recipient complaints or difficulties, the clinician needs to consider the typical periods for acclimatization following initial activation. First, sound quality and speech recognition often improve over the first few months of cochlear implant use. As a result, programming adjustments may not result in optimal sound quality during the early stages of implant use. The troubleshooting guidelines in this section may be used to improve sound quality and performance, but if the complaint persists, additional counseling may be necessary to remind the recipient that performance should improve with continued use and experience. Second, recipients usually require several days to completely adjust to any programming modifications. If a user has minor complaints about sound quality or performance immediately following a programming session, the clinician needs to use his or her clinical judgment regarding the need for additional changes to the program. The audiologist may ask the recipient to use the modified program for 7 to 10 days before additional adjustments are made. Following this period of time, most users adapt to the changes and will report better hearing than before the programming session.

Table 8–1 provides step-by-step instructions for various programming adjustments to address the most common user complaints. When a particular complaint is resolved by one of the steps, the clinician should cease the adjustments.

## KEY CONCEPTS

The following key concepts were discussed in this chapter:

- Most recipients will achieve remarkable success, but others will require special programming to perform at their highest potential.
- The user's preimplant characteristics will play an important role in the postoperative implant performance.
- Appropriate and thorough preimplant counseling is absolutely necessary to help the patient set forth realistic expectations.
- Common complaints and complications often can be addressed with step-by-step modifications to stimulation levels and other programming parameters.

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**Table 8–1.** Troubleshooting Guidelines to Address Common Recipient Complaints

| *Recipient Complaint* | *Programming Adjustments* |
|---|---|
| Echo or boomy sound quality | 1. Common at initial activation, allow for acclimatization<br>2. Global reduction in USL; if sound too soft, return to original USL<br>3. Ensure balanced loudness at USL across electrode array<br>4. Reduce stimulation levels in LF channels<br>5. Reduce stimulation levels in HF channels<br>6. Reduce stimulation rate<br>7. For CIS-type strategies, reduce gains in LF channels*<br>8. For CIS-type strategies, reduce gains in HF channels* |
| Ringing, pinging, or tinny sound quality | 1. Ensure balanced loudness at USL across electrode array<br>2. Reduction stimulation levels in HF channels<br>3. Reduce stimulation rate<br>4. For CIS-type strategies, reduce gains in HF channels*<br>5. Sweep to locate electrodes with ringing percept; disable them<br>6. Disable most basal electrode(s)<br>7. This may be a counseling issue. If recipient is newly activated, then it may be beneficial for him or her to acclimate to high-frequency stimulation |
| Sound is muffled, fuzzy, or not clear | 1. Global increase in USL; if speech is too loud, decrease USL; if no change, return to original USL<br>2. Ensure balanced loudness at USL across electrode array<br>3. Increase USL in HF channels<br>4. Reduce USL in LF channels<br>5. Reduce stimulation rate<br>6. For CIS-type strategies, increase gains in HF channels*<br>7. Increase IDR<br>8. Increase microphone sensitivity<br>9. Global increase in threshold (T) levels<br>10. Increase loudness growth function (maplaw) if not optimized |
| Patient's own voice is too loud, but speech from others is comfortable | 1. Ensure balanced loudness at USL across electrode array<br>2. Global decrease in USL; if speech of others too soft, increase sensitivity, T levels, or loudness growth for low-level inputs<br>3. Decrease USL in LF channels; increase USL in HF channels<br>4. Global increase in gains<br>5. Reduce LF gains; increase HF gains* |
| Buzzing, frying, or humming sound | 1. Determine ambient noise or electrical stimulation: reduce sensitivity<br>2. If solved, recipient needs acclimatization to low-level sounds; if still unhappy, slightly reduce microphone sensitivity and/or IDR<br>3. If not solved, adjust T levels to balance loudness at 50% of dynamic range<br>4. Global decrease in T levels<br>4. Ensure balanced loudness at USL across electrode array<br>5. Global decrease in USL<br>6. Decrease USL in LF channels |

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*continues*

**Table 8–1.** *continued*

| *Recipient Complaint* | *Programming Adjustments* |
|---|---|
| Roaring | 1. Global reduction in USL; if solved, but speech too soft, return to original USL<br>2. Ensure balanced loudness at USL across electrode array<br>3. Adjust T levels to balance loudness at 50% of dynamic range<br>4. Global decrease in T levels |
| Sound is hollow | 1. Global reduction in USL; if sound too soft, return to original USL<br>2. Ensure balanced loudness at USL across electrode array<br>3. Reduction in stimulation levels at LF channels<br>4. Increase in stimulation levels in HF channels<br>5. Increase in stimulation levels in MF channels<br>6. For CIS-type strategies, reduce gains in LF channels<br>7. For CIS-type strategies, increase gains in HF or MF channels |
| Overall poor sound quality/performance | 1. Global increase in USL; if speech is too loud, reduce USL; use ESRT to estimate USL; if not solved, return to original USL<br>2. Ensure balanced loudness at USL across electrode array<br>3. Increase USL in HF channels<br>4. Sweep at USL: ensure pitch percepts in expected tonotopic pattern<br>5. Sweep at USL: find electrodes with poor quality/aversive percept<br>6. If several months past activation, try different stimulation rates<br>7. If several months past activation, try other signal coding strategies<br>8. Increase PW across all channels |
| Facial nerve stimulation | 1. Single-channel stimulation: increase PW on channel or globally; if increased PW negatively influences loudness growth, disable electrode<br>2. Stimulation in live speech mode: (a) sweep at USL to find problem channels; (b) may increase maximum USL permissible to elicit facial stimulation; (c) disable 2–3 electrodes at a time in live speech mode to find problem; (d) increase PW or disable problem electrodes<br>3. If stimulation still exists with 25% electrodes disabled: change the electrode coupling mode (i.e., monopolar or bipolar) |
| Discomfort, pain, or nonauditory percept, such as tinnitus or tactile sensation | 1. Measure ESRT to see if USL too high<br>2. Sweep at USL: find suspect channels, increase PW, remeasure stimulation levels; if not solved, disable electrodes<br>3. Disable 2–3 electrodes at a time in live speech mode to find problem; increase PW or disable problem electrodes<br>4. If problem across several electrodes, increase PW for all electrodes<br>5. Change electrode coupling mode (i.e., monopolar or bipolar)<br>5. Decrease stimulation rate: high rate may cause tinnitus/vibration<br>6. Change signal coding strategy |
| Too loud | 1. Reduce upper-stimulation levels<br>2. Reduce volume control setting |

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**Table 8–1.** *continued*

| *Recipient Complaint* | *Programming Adjustments* |
|---|---|
| Too soft | 1. Increase upper-stimulation levels<br>2. Increase volume control setting<br>3. Remeasure T levels<br>4. Consider situational increase in sensitivity if complaint pertains to low-level speech<br>5. Consider reducing aggressiveness of noise reduction technologies (e.g., ClearVoice, ASC) if complaint primarily occurs in noisy places |
| Elevated sound-field detection thresholds | 1. Ensure volume control and sensitivity settings are appropriate<br>2. Ensure microphone is functioning normally and microphone covers are clean<br>3. Increase T levels in Cochlear programs<br>4. Increase T levels in AB recipients and/or increase IDR<br>5. Increase maplaw setting in MED-EL recipients or measure THR levels |

*Note.* *should not be done for *n*-of-*m* coding strategies; *IDR* = input dynamic range; *HF* = high-frequency; *LF* = low-frequency; *MF* = mid-frequency; *PW* = pulse width; *USL* = upper-stimulation levels.

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# 9

# Hearing Assistance Technology (HAT) and Cochlear Implants

Adults and children who use cochlear implants often achieve good open-set speech perception in quiet listening situations; however, they will experience significant difficulty understanding speech in the presence of background noise. In fact, when compared to average performance in quiet, speech perception in noise is significantly poorer by approximately 40% (Caldwell & Nittrouer, 2013; Fetterman & Domico, 2002; Firzst et al., 2004; Schafer & Thibodeau, 2003, 2004). Similar performance detriments are found when comparing average scores in noise of people with cochlear implants and normal-hearing peers (Caldwell & Nittrouer, 2013; Neuman, Wroblewski, Hajicek, & Rubinstein, 2012). The use of HAT, such as personal frequency modulated (FM) or digital radio frequency (RF) systems, addresses these listening difficulties in noise by directing the signal from the primary talker straight to the listener's ear.

FM systems provide significant improvements in speech-recognition performance in noise for adults by an average of 12 dB or 29% to 44% correct when compared to performance with the cochlear implant alone (Schafer, Huynh, Romine, & Jimenez, 2013; Schafer, Romine, Musgrave, Momin, & Huynh, 2013; Schafer & Thibodeau, 2004; Wolfe & Schafer, 2008a, 2008b; Wolfe et al., 2009; Wolfe, Schafer, et al., 2013). Furthermore, in several studies, adults and children report improved clarity, ease of listening, and sound quality when using HAT and communicating in small and large groups (Fitzpatrick et al., 2010; Schafer, Huynh, et al., 2013; Schafer, Musgrave, et al., 2013; Schafer & Thibodeau, 2004). Similarly, personal digital RF systems, which utilize higher transmitting frequencies (e.g., 2.4 GHz) than traditional FM systems as well as a variable carrier frequency to avoid signal interference with other devices, significantly improve speech-recognition performance of adults and children with cochlear implants over their performance in noise with the cochlear implant alone and over traditional FM systems in the test conditions with the highest noise levels (Wolfe, Morais, Schafer, Mills, Mülder, et al., 2013; Wolfe, Morais, Schafer, Mills, Peters, et al., 2013).

This chapter introduces personal wireless FM and digital RF devices for improving speech-recognition performance in noise and discusses programming considerations for coupling sound processors to HAT. These sections are followed by an overview of devices for improving telephone conversations and hearing audio from other devices.

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## BASIC DESCRIPTION OF PERSONAL SYSTEMS

Personal systems used for improving speech-recognition performance in individuals with cochlear implants utilize one of the following types of transmission: (1) digital RF, (2) traditional FM, (3) infrared (light), or (3) electromagnetic via induction loop (i.e., telecoil). When using any of these personal systems, the talker wears a transmitter and microphone, and the listener wears a specialized receiver with an induction neckloop (body-worn infrared; body-worn FM/RF) or plugs a miniaturized receiver directly into his or her cochlear implant sound processor (digital RF; FM). A personal, ear-level, or body-worn receiver decodes the FM, RF, or infrared signal; converts it to an electrical signal; and sends it through the sound processor. Personal receivers are connected to the sound processor using a design-integrated receiver, manufacturer-specific battery cover, streamer, cable, adaptor, or earhook.

Induction loop receivers, which utilize electromagnetic transmission, may be used with all contemporary sound processors by activating the built-in or accessory telecoil. The transmitter sends the speech signal via infrared or FM or digital RF transmission to a radio receiver. The electrical signal from the receiver is converted to an electromagnetic signal and sent through an induction loop of wire that is placed around the listener's neck (i.e., induction neckloop) or around the room. The listener's active telecoil detects the electromagnetic signal and converts it back into an electrical signal, which then is processed by the cochlear implant. When an induction loop is placed around a room, no specialized equipment is required. Conversely, when using a neckloop, a specialized body-worn receiver with an induction loop must be used.

## OPTIMAL SYSTEMS FOR USE WITH A COCHLEAR IMPLANT

There are several published studies on the benefits of personal HAT and cochlear implants with participants using digital RF, FM, and induction neckloops. To the authors' knowledge, no studies have examined the potential benefits of infrared technology or induction loops around a room for users of cochlear implants.

According to a 2009 meta-analysis—the strongest level of evidence according to the American Speech-Language-Hearing Association (2004)—personal FM systems provide the best speech recognition in noise when compared to sound-field (i.e., loudspeaker) FM systems (Schafer & Kleineck, 2009). The personal system provided a 38% improvement over the cochlear implant alone, as compared to the 17% and 4% improvements from the two sound-field systems, respectively. More recently, three studies have examined and compared the benefit of induction neckloop systems to personal FM systems in individuals with contemporary cochlear implants, and overall, these studies support the use of these types of receivers for providing similar or slightly poorer benefit than personal FM systems (Schafer, Huynh, et al., 2013; Schafer, Musgrave, et al., 2013; Schafer, Romine, et al., 2013; Wolfe, Schafer, et al., 2013). Additionally, recent research suggests that personal digital RF systems result in significantly better speech-recognition performance in high levels of background noise (70, 75, and 80 dBA) in comparison to personal adaptive FM systems (Dynamic FM) at the same high noise levels (Wolfe, Morais, Schafer, Mills, Mülder, et al., 2013; Wolfe, Morais, Schafer, Mills, Peters, et al., 2013).

## DESCRIPTION AND PROGRAMMING FOR PERSONAL HAT AND COCHLEAR IMPLANTS

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### FM and Digital RF Transmitters

Contemporary FM and digital RF transmitters are small rechargeable devices that are worn on a belt, lavaliere, or lapel. Several FM and RF transmitters are shown in Figure 9–1. Each of these transmitters offers various options for the listener including microphone style (Figure 9–2), directionality for the microphone, multichannel capabilities, and programmability. All of these transmitters




**FIGURE 9–1.** Examples of contemporary FM or digital RF transmitters: **a.** Oticon Amigo T5; **b.** Oticon Amigo T30; **c.** Phonak Roger Pen; **d.** Phonak Roger Clip-On Mic; **e.** Phonak Roger inspiro. *Note.* Courtesy of Oticon Inc. and Phonak.

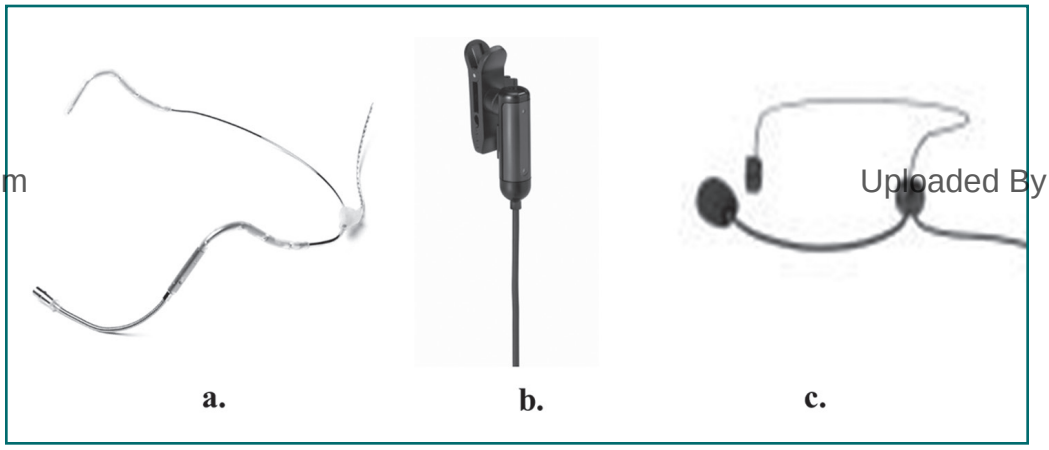



**FIGURE 9–2.** Examples of FM or digital RF transmitter microphones: **a.** Oticon Amigo Binaural Boom Mic; **b.** Phonak iLapel microphone; **c.** Phonak Easyboom. *Note.* Courtesy of Oticon Inc. and Phonak.

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may be used with FM or induction loop receivers, but, to utilize digital transmission, a digital RF transmitter must be used with a digital RF receiver. One recently introduced transmitter may be programmed to deliver the signal of interest via digital RF and analog FM transmission.

Programmable digital FM and RF transmitters offer the audiologist a great deal of flexibility when determining the optimal settings for an adult or child with a cochlear implant. Programming options for FM transmitters include selection of one or more FM transmitting channels, wireless programming of the FM receiver (e.g., channel, FM gain), DataLogging (i.e., storing noise level), creation of wireless team-teaching connections (i.e., combined signal from more than one talker at a time), evaluation of transmitter/receiver function, monitoring of classroom noise, and help with FM channel selection in an environment to avoid interference. The same options are available for digital RF transmitters, but these transmitters do not require the channel changing or monitoring options because this technology continuously changes the transmitting frequency (i.e., frequency hopping) to avoid interference from other digital transmission in close proximity to the transmitter. The Phonak Roger inspiro transmitter is compatible with existing Phonak FM receivers as well as Roger receivers as long as the clinician programs it to deliver both digital RF and analog FM signals. It is important to note that a special cochlear implant module must be selected on the inspiro transmitter (i.e., "Setting 9") in order to obtain appropriate RF output from the Roger X receiver when coupled to the various cochlear implant processors (Table 9–1). This special cochlear implant setting is needed in order to overcome the high input impedance present at the direct audio input (DAI) of cochlear implant sound processors relative to that of hearing aids. The digital RF Phonak Roger Pen and Clip-On Mic, however, can only transmit to Roger receivers.

Most digital RF and FM transmitters have an auxiliary input port to allow for connection to audio sources including iPods/iPads, televisions, computers, and DVD/Blu-ray players. Additionally, when used with the corresponding Phonak receivers, the Phonak FM inspiro and digital RF Roger transmitters have Dynamic functionality, which automatically adjusts the gain of the RF or FM receiver depending on the noise level in the environment. When the noise level exceeds a specified intensity, the transmitter broadcasts a signal to the receiver to automatically increase the gain of the receiver in order to maintain a favorable signal-to-noise ratio at the listener's ear. Several Phonak transmitters also contain Bluetooth technology, which may be paired with Bluetooth compatible mobile phones, computers, and other devices.

It is important to note that, for some Cochlear sound processors, specific FM transmitting frequencies are recommended to avoid interference between the two devices. Hearing professionals should consult the implant and FM system manufacturers to determine the most appropriate channels for the combination of a particular sound processor and FM system.

Given the fact that digital RF and FM system technology continues to evolve and change, cochlear implant and FM/RF system manufactur-

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**Table 9–1.** Phonak Roger X Settings for Various Cochlear Implant Sound Processors

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| *Model* | *Roger X Setting* |
|---|---|
| Advanced Bionics Naída CI via ComPilot | Setting 3 |
| Advanced Bionics Neptune | Setting 1 |
| Advanced Bionics Harmony/ Auria | Setting 1 |
| Cochlear Nucleus 5 | Setting 9 |
| MED-EL OPUS 2 | Setting 2 |
| Body-worn processors using ML CI S | Setting 5 |

ers recognize the need for audiologist, teacher, and parent education. As a result, most manufacturers provide online training and support, knowledgeable representatives, and step-by-step guides to aid the audiologist, and sometimes teachers and parents, with ordering appropriate equipment, procedures for connecting to a cochlear implant, information about using FM systems, and instructions for troubleshooting problems.

## FM, RF, and Induction Loop Receivers

Three contemporary miniaturized FM/RF receivers that may be used with any sound processor are shown in Figure 9–3. All three of these receivers have been tested and shown to provide substantial speech-recognition benefits in research studies with cochlear implant recipients. As outlined in Table 9–2 and shown in Figure 9–4, these and

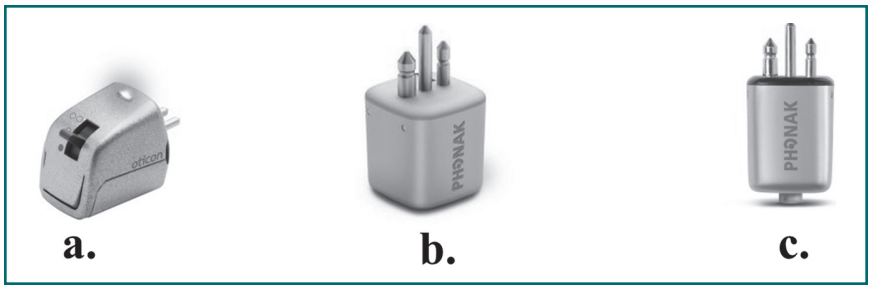



**FIGURE 9–3.** Examples of three contemporary universal FM or digital RF receivers: **a.** Oticon Amigo R2 FM receiver; **b.** Phonak Roger X digital RF receiver; **c.** Phonak MLxi FM receiver. *Note.* Courtesy of Oticon Inc. and Phonak.

**Table 9–2.** Types of Connections for FM and Digital RF Receivers to Cochlear Implant Sound Processors

| *Type of Connection* | *Examples of Equipment* | *Used with Cochlear Implant Sound Processors* |
|---|---|---|
| Specialized Receiver/ Streamer/Battery Pack | • A. Oticon Amigo FM-CI adaptor plus processor-specific cord with universal FM/RF receiver<br>• B. Phonak ComPilot with universal FM/RF receiver<br>• C. Advanced Bionics Mini Battery Pack with universal FM/RF receiver | • A. Cochlear Freedom BTE<br>• B. Advanced Bionics Naída CIQ70<br>• C. Advanced Bionics RONDO |
| Adaptor | • A. Nucleus 5 Euro Accessory Adaptor with universal FM/RF receiver | • A. Nucleus 5 (CP 810) |
| Earhook | • A. Advanced Bionics iConnect earhook with universal receiver<br>• B. Advanced Bionics Auxiliary Audio Earhook with cable | • A. Advanced Bionics Harmony & Auria processors<br>• B. Advanced Bionics Platinum BTE |
| Design Integration | • A. Phonak MicroLink Freedom FM<br>• B. Phonak ML14i<br>• C. Phonak Roger 17 | • A. Cochlear Freedom BTE<br>• B. Nucleus 5<br>• C. Advanced Bionics Naída CIQ70 |
| Direct Connect | • A. All universal FM/RF receivers plugged straight to processor | • A. Advanced Bionics Neptune with Neptune Connect configuration, Cochlear Freedom Body-Worn, & MED-EL OPUS 2 with FM Battery Pack |

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**FIGURE 9–4.** Some examples of types of connections between the cochlear implant sound processor and personal FM/RF system. *Note:* * = cable not shown. Courtesy of Oticon Inc. and Phonak.

other miniaturized receivers connect to body-worn and ear-level cochlear implant sound processors in one of five ways: (1) specialized battery pack/receiver/streamer, (2) adaptor, (3) earhook, (4) design integration, or (5) direct connect.

Some of the programmable options on the universal Oticon Amigo R2 receivers (see Figure 9–3) include 0 to 30 dB of programmable gain (i.e., FM advantage over input from the microphone of the implant sound processor), multichannel capa-

bility, programmable default channel, programmable switch, LED status indicator, sleep mode, manual channel seek button, tamper resistant switches, and wireless programmability. Programmable options on the universal Phonak MLxi and the design-integrated MicroLink Freedom FM and ML14i receivers include multichannel capability, adaptive FM gain (i.e., Dynamic FM when used with the inspiro transmitter), FM gain adjustment, and sleep/stand-by mode. Finally, some of the programmable options on the universal Phonak Roger X and design-integrated Roger 17 include Dynamic capabilities, auto-connect to nearby Roger transmitters, gain adjustment, link quality measurement, and sleep/standby mode. Previous generations of universal Phonak receivers (e.g., Micro MLxS and MLxS) are also compatible with contemporary sound processors and offer multichannel capability, FM gain adjustment, soft squelch (turns off receiver when transmitter off), and sleep mode.

Three examples of contemporary induction loop receivers with built-in neckloops are provided in Figure 9–5. The Phonak MyLink+ is a multichannel, rechargeable FM receiver that includes a noise blocker to reduce the input from the FM system when in a quiet environment, a sleep mode to reduce the input from the FM transmitter when it is not in use (i.e., avoid hissing noise), wireless synchronization, Dynamic FM, and an input port for regular headphones to allow for troubleshooting from a normal-hearing listener. The Phonak Roger MyLink is a digital RF receiver with the same options and characteristics as the MyLink+. The Oticon Arc is a multichannel, rechargeable FM receiver with sleep mode, wireless synchronization, wireless programming, adjustable receiver gain, tamper-resistant switch, channel seek function, and status indicator. All three of these devices allow for connection to auxiliary input from other audio sources including computers, televisions, and CD players.

## Programming Considerations for FM/RF Receivers and Sound Processors

### *Programming FM or Digital RF Receivers*

On programmable FM/RF receivers, the gain may be adjusted to alter the relationship of the signals from the FM/RF system and processor microphones. The range of available gain settings vary across manufacturers and receivers, but in general, a higher gain setting provides greater advantage for the signal from the FM/RF system relative to the input from the sound processor microphone. For some Phonak receivers, this gain setting is only applicable in low-noise environments and is prior to Dynamic processing.

Previous research suggests that higher FM gain settings allow for significantly better speech perception in noise than lower gain settings (Schafer, Wolfe, Lawless, & Stout, 2009; Wolfe & Schafer, 2008a, 2008b). According to the results of these studies, optimal performance and comfort

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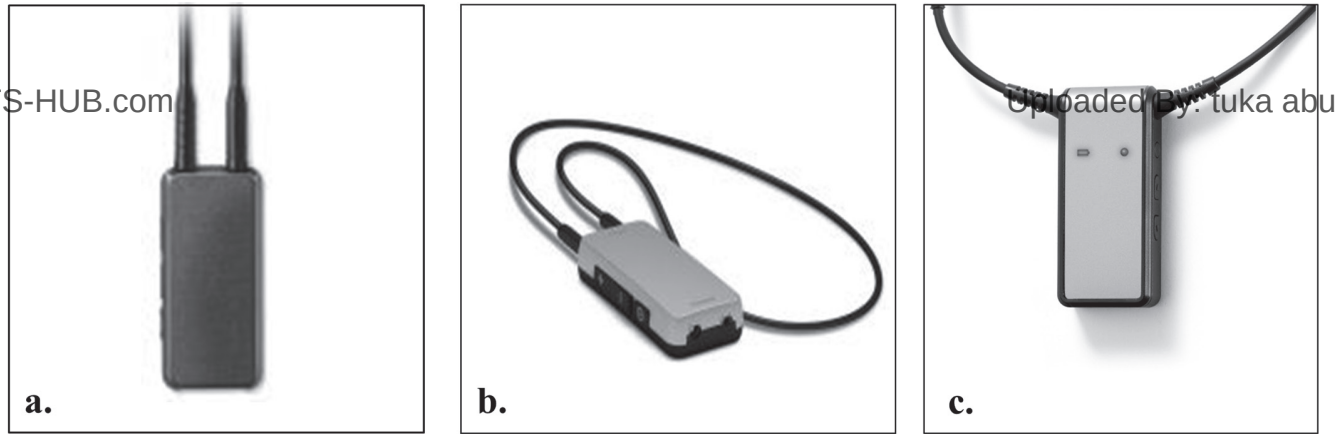



**FIGURE 9–5.** Three examples of contemporary induction neckloop receivers: **a.** Phonak MyLink+; **b.** Phonak Roger MyLink; **c.** Oticon Amigo Arc. *Note.* Courtesy of Oticon Inc. and Phonak.

for nine adults and five children using Advanced Bionics implants was achieved with +14 and +16 receiver-gain settings. Conversely, increased receiver-gain settings did not provide better performance for eight users of Cochlear sound processors. Differences in performance between the two manufacturer groups were related to sound processor settings and will be addressed in the following section. No studies have been conducted on the effects of receiver gain for users of MED-EL cochlear implants, but similar results are expected as those obtained with other devices.

Additional research suggests that Dynamic FM results in significantly better speech recognition in noise than traditional FM systems set to +10 dB gain for users of Advanced Bionics devices (Wolfe et al., 2009). In the initial experiment, users of Cochlear Freedom did not show significant benefit from either type of FM system. However, when Autosensitivity Control (ASC) preprocessing was activated in the Freedom sound processor, speech recognition results were similar to those using Advanced Bionics implants. According to the results of this study, Dynamic FM provides even greater benefit to users of cochlear implants than traditional, fixed-gain FM systems. However, users of Cochlear sound processors must be using ASC preprocessing, which is a form of input processing that is designed to improve listening in noise. ASC is currently enabled in the default program of contemporary Cochlear sound processors.

Regardless of the type of personal FM receiver (traditional FM, digital RF, Dynamic, neckloop), the gain of the FM/RF receiver should be set to achieve transparency or equal output when providing the same inputs (i.e., 65 dB SPL speech signal) to the sound processor microphone and the FM/RF transmitter microphone. A transparent fitting of an FM/RF system will guarantee a standardized fitting approach across patients and will ensure an optimal signal-to-noise ratio at the child's ear when the system is in use. Published guidelines from the American Academy of Audiology (2008) exist for hearing aids coupled to personal FM systems, and recent research provides recommendations for cochlear implants coupled to personal FM systems (Schafer, Musgrave, et al., 2013). Specifically, an audiologist can measure acoustic output from a cochlear implant sound processor with monitor earphones, specialized equipment from cochlear implant manufacturers, and a hearing aid analyzer with a HA-1, 2 cc coupler. Adjustments to the FM/RF gain or volume can be made until equal outputs (i.e., transparency) are measured with 65 dB SPL speech inputs (1) to the sound processor and (2) to the FM/RF transmitter microphones. In the Schafer, Musgrave, et al. study (2013), behavioral testing in four adults with cochlear implants provided preliminary support to the proposed electroacoustic test protocol. However, at the time of this writing, the specialized equipment for conducting these measures is cumbersome and somewhat unreliable. Further technological developments in measurement tools and equipment are necessary to improve and validate a fitting procedures these systems to sound processors.

### *Sound Processor Programming and HAT*

Several parameters in sound processor programming may significantly influence performance with a personal FM/RF system including the audio-mixing ratio, input dynamic range (IDR), and sensitivity.

**Audio-Mixing Ratios.** Similar to FM/RF gain, audio-mixing ratios in the sound processor determine the relationship or ratio between the inputs from the sound processor and the FM/RF transmitter microphones (Aux/Mic). Although this discussion is focused on FM/RF systems, most of the ratios are applicable to telecoil settings for any auxiliary-audio device (e.g., iPod; iPad, MP3 player) that is electrically coupled to the sound processor (see Chapters 4, 5, and 6 for more details).

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In Advanced Bionics processors, there are five clinician selectable, fixed audio mixing ratios (Mic/Aux) available for a given program: (1) microphone only ("Mic Only"), no auxiliary input 30/70; (2) 50/50, equal emphasis is provided for the signals from the sound processor microphone and auxiliary input, with 6 dB of attenuation provided for each input; (3) 30/70, the input from the sound processor microphone is attenuated by 12 dB, whereas 3 dB of attenuation is provided to the input from the auxiliary source; (4) auxiliary

only ("Aux Only"), no processor microphone; and (5) auxiliary input only attenuated ("Aux Only [Attn]"), processor microphone attenuated by 20 dB. The 50/50 option is the default setting.

The Cochlear Freedom, Nucleus 5, and Nucleus 6 processors have several mixing ratios some of which include a (1) 1:1-equal emphasis, (2) 2:1-sound processor microphone attenuated by 6 dB, (3) 3:1-processor microphone attenuated by 10 dB, and (4) 10:1 processor microphone attenuated by 20 dB. In the Nucleus 5 and 6, the mixing ratios may be adjusted with the wireless remote controls, but the ratios for the Freedom must be adjusted in the clinician software. The default mixing ratios are 2:1 for adults and 1:1 for children. However, it is important to consider that when an FM/RF system is coupled to the Nucleus 5 and Nucleus Freedom processors, the floor of the IIDR is automatically attenuated by 10 dB (i.e., the T-SPL is raised from 25 to 35 dB SPL), even when set to 1:1. The elevation of the T-SPL does not occur when an accessory is in use with the Nucleus 6 processor when set to a 1:1 mixing ratio.

According to previous research on 12 adults using Advanced Bionics sound processors and FM receivers, no statistically significant differences in speech perception in noise or subjective benefit exist between a 30/70 and 50/50 audio-mixing ratio (Wolfe & Schafer, 2008a, 2008b). However, when attempting to hear low-level speech in quiet that was directed to the sound processor (i.e., not to the FM transmitter), the adults performed significantly poorer when using the 30/70 mixing ratio. Pilot data collected in the authors' laboratories (Wolfe & Schafer, 2010) show similar results for adults using Cochlear Freedom sound processors when comparing the 3:1 and 1:1 ratios. Therefore, to ensure audibility of FM/RF and environmental signals, a 50/50 or 1:1 mixing ratio is recommended for users of Advanced Bionics and Cochlear processors, respectively. This recommendation particularly applies to children using FM/RF in classroom settings, so that they are able to adequately hear their peers in the classroom as well as their own voice.

It is worth noting that when digital RF/FM receivers are active with Nucleus Freedom and Nucleus 5 sound processors, the T-SPL parameter is increased from 25 to 35 dB SPL, even when the mixing ratio is set to 1:1. As indicated in Chapter 5 of this book, the T-SPL parameter determines the input level that is mapped at T Level. As a result of the increase in T-SPL associated with digital RF/FM use, the recipient's access to low-level sounds may be compromised. In contrast, the T-SPL remains at 25 dB SPL when directly coupled digital RF/FM is used with the Nucleus 6 sound processor. However, the T-SPL of the Nucleus 6 sound processor is increased from 25 to 35 dB SPL when the telecoil is engaged, even when the mixing ratio is set to 1:1. As a result, Nucleus 6 users may experience a reduction in the audibility of low-level incidental sounds when using an induction neckloop digital RF/FM receiver.

MED-EL RONDO, TEMPO+, OPUS 1, and OPUS 2 sound processors do not offer various mixing ratios for FM/RF systems. A 1:1 audio-mixing ratio is enabled automatically. If the user desires a setting with mostly FM/RF input, the sensitivity on the TEMPO+ sound processor knob may be turned to the off position (i.e., 11:00) or the Fine-Tuner remote control may be used to turn down the sensitivity on the RONDO, OPUS 1, and OPUS 2 processors. Reduction of the sensitivity will increase the compression kneepoint, which will make soft sounds presented to the sound processor microphone less audible. Although there are no published data on the effects of audio-mixing ratios for individuals with MED-EL cochlear implants, one recent study demonstrates the benefits of personal digital RF systems over personal FM systems for users of MED-EL OPUS 2 sound processors (Wolfe, Morais, Schafer, Mills, Peters, et al., 2013).

**Input Dynamic Range (IDR).** As discussed in the section on receiver programming, findings from two studies revealed greater FM benefit for users of Advanced Bionics versus Cochlear processors (Schafer et al., 2007; Wolfe et al., 2009). However, in the Wolfe et al. (2009) study, the researchers resolved the manufacturer differences by adjusting the IDR settings in Cochlear sound processors. The IDR is the range of acoustic-input levels (i.e., 30 to 60 dB) that are coded by the sound processor within the user's electrical dynamic range. Signal levels that exceed the upper end of the IDR are subjected to high-level compression, and low-level

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signals below the IDR are not processed and are likely inaudible to the listener.

Advanced Bionics implants have a default IDR of 60 dB SPL in conjunction with an automatic gain control with dual time constants. As a result, inputs ranging from 25 to 85 dB SPL are mapped into the person's electrical dynamic range. The goal of this wide IDR is to provide access to a wide range of everyday sounds including audibility for low-level inputs as well as comfort for moderate and high-level inputs without the use of compression with a high compression ratio. MED-EL sound processors use a similar wide IDR of approximately 55 dB SPL. In contrast, Cochlear uses what is referred to as an instantaneous input dynamic range (IIDR), which refers to the range of instantaneous, short-term acoustic fluctuations that are mapped into the recipient's electrical dynamic range without high-level compression. The default IIDR of Nucleus Freedom, Nucleus 5, and Nucleus 6 sound processors is 40 dB and ranges from 25 to 65 dB SPL. The goal of a narrower IDR is to focus on coding the rapid fluctuations in a static speech (which typically has a range of 40 dB between the peaks and valleys for average conversational level speech) into the recipient's electrical dynamic range, while discarding low-level noise.

The IDR may influence the signal from an RF/FM system for two reasons: (1) typical signals from an RF/FM system are often at a higher intensity (i.e., 75 dB SPL), and (2) the IDR influences the signals coded at the upper end of the dynamic range. If a recipient were to use a narrow IIDR (i.e., Cochlear) with an upper end of 65 dB SPL, the signal from the FM/RF system likely will be subjected to compression with a high compression ratio. Furthermore, this compression will diminish any increases in receiver gain provided either by clinician-adjusted manual increase or by an adaptive (i.e., Dynamic) increase. In other words, speech recognition in noise with a receiver gain of +10 dB would produce similar performance as receivers programmed for +14 or +16 gain. Compression of the FM/RF signal is less probable for users of Advanced Bionics implants because the upper end of the IDR is 85 dB SPL, which allows for processor coding and user perception of increases in receiver gain. Further research is necessary to determine an ideal IDR for users of cochlear implants coupled to personal FM/RF systems. At this time, the IDR parameter is not typically adjusted for FM/RF use.

**Processor Sensitivity.** The effects of a narrow IDR may be addressed by changes to the third parameter referred to as sensitivity. The sensitivity determines the gain for inputs from the sound processor microphone (i.e., softest sound that will be picked up by the microphone), and in the case of the Nucleus 5 and Nucleus 6 sound processors, also for inputs from the FM/RF. When sensitivity is increased from the default setting, lower intensity sounds from a distance will be mapped into the user's electrical dynamic range. A decrease to the sensitivity, however, will reduce the user's ability to hear soft and distant speech, which may improve hearing in background noise. In addition, some evidence shows that lower sensitivity settings provide significantly better speech recognition in noise for users of Cochlear implants than higher sensitivity settings when using an FM system (Aaron et al., 2003).

Although reductions to sensitivity may increase the audibility of the FM/RF system, it also may influence negatively the user's ability to hear environmental sounds. Therefore, it is important to find a good balance between the two signals, which may require behavioral testing or a good reporter (i.e., older child or adult). For younger children, access to the sensitivity may be disabled on the sound processor and fixed in the sound processor programming. When this is the case, it is best to leave the sensitivity at the typical user setting, even when a FM/RF system is in use.

The effect of the sensitivity control varies between the Cochlear Nucleus 5/6 and Freedom sound processors. For the Freedom processor, the sensitivity setting affects only the signal from the sound processor microphone with no attenuation provided to the FM/RF signal. As a result, lower sensitivity settings will provide greater emphasis on the FM/RF signal. However, the high-level signal from the FM/RF system will be subjected to compression at the input of the Freedom processor. In contrast, the sensitivity setting for the Nucleus 5 and 6 processors affects the gain applied to the signals from both the sound processor

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microphone and the FM/RF system. Therefore, lower sensitivity settings prevent both signals from being subjected to compression. Because the FM/RF input should be higher than the ambient noise level in many instances, the FM/RF advantage should be preserved. Also, at lower sensitivity settings, an increase in RF/FM receiver gain provided through manual adjustments or with the adaptive Dynamic feature should be more beneficial to the user because the increased strength of the RF/FM signal is not compressed at the input of the sound processor. In MED-EL and Advanced Bionics sound processors, changes to sensitivity influence only the signal at the processor microphone; therefore, a reduction to sensitivity may result in greater emphasis for the FM/RF signal.

Another programmable option for users of Cochlear processors is an input-preprocessing feature known as Autosensitivity (ASC). This optional setting automatically reduces the sensitivity of the processor microphone according to the noise level in the environment and the signal-to-noise ratio at the microphone. The goal of ASC is for the peaks of speech to exceed the long-term-average spectrum of the noise by at least 15 dB when the noise level in the environment exceeds 57 dB SPL. According to previous research with Cochlear processors, the use of ASC will improve significantly speech-recognition performance in noise with FM relative to a FM condition with no ASC (Wolfe et al., 2009). Also, when the ASC parameter was enabled, speech recognition with Dynamic FM was no longer significantly different between users of Advanced Bionics and Cochlear processors. This occurred because ASC reduces the effects of the narrow IDR for users of Cochlear processors, and it allows for improved speech-recognition performance in noise with automatic increases to receiver gain (i.e., Dynamic feature). Results of another condition in this study revealed that use of ASC also improved performance when the speech signal was presented to the processor microphone with the FM transmitter muted. In 2013, Wolfe, Schafer, et al. (2013) report significantly better speech-recognition performance in noise when using multistage ASC with two types of FM systems (neckloop and electrically coupled FM receivers) in users of the Cochlear Nucleus 5 sound processor. This two-stage approach applies ASC to the sound processor microphone before its signal is mixed with the FM signal and then applies ASC again once the FM and processor signals are mixed. The goals of the two-stage approach were to reduce the level of the background noise within the processor signal and to provide an optimal signal-to-noise ratio for the mixed signal. In summary, this research supports the use of one-stage or preferably two-stage ASC for users of Cochlear processors in noise with and without the use of a personal digital RF or FM systems. It should be noted that the two-stage ASC preprocessing is also used in the Nucleus 6 sound processor.

### Fitting Procedures for Personal FM/RF Systems

Prior to fitting the personal FM/RF system, the audiologist may consider using the proposed electroacoustic test protocol in the Schafer, Musgrave, et al. study (2013), which was described in the section on programming digital RF or FM receivers.

Given the improved compatibility of personal FM/RF systems and sound processors, connecting the two devices is fairly simple. To aid with these connections, cochlear implant and FM/RF system manufacturers offer step-by-step guides, online and in print, as well as onsite and phone support from representatives. As a result, the majority of this section focuses on fitting procedures following the successful connection of a personal FM/RF system to a cochlear implant. However, a brief introduction to connecting the devices is provided.

In general, as a first step to connecting the devices, the FM/RF system receiver, FM/RF transmitter, and sound processor are turned off. Second, the FM/RF receiver is connected to the sound processor with the appropriate streamer, battery pack, adaptor, cable, or earhook. Then, for most systems, the speech processor is turned on, followed by activation of the FM/RF receiver and transmitter. Finally, a listener with normal-hearing sensitivity performs a listening check after this connection is made, which is described in the following section on monitoring and troubleshooting. It should be noted, however, that contemporary cochlear implant processors (Nucleus 5

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and 6, Naída CO Q70, MED-EL OPUS 2) are designed to automatically detect the presence of an RF signal at the DAI of the processor, so it is typically unnecessary to power down each of the components of the CI/RF interface in order to receive the RF signal.

At the time of the fitting, the adult or older child should be tested in a quiet and noise condition with and without the FM/RF system. The speech recognition measure needs to have multiple lists that are equivalently difficult in the type of noise that will be used (i.e., multitalker babble). In addition, the speech and noise need to be recorded on separate channels so that they may be presented from spatially separated loudspeakers (e.g., speech at 0 degrees; noise at 180 degrees).

When percent-correct scoring is used, the test material needs to be challenging enough, in quiet and in noise, to avoid ceiling effects (i.e., 100% correct). Adaptive or modified-adaptive test measures (i.e., Hearing in Noise Test [HINT]; Nilsson, Soli, & Sullivan, 1994; Bamford-Kowal-Bench Speech in Noise Test: BKB-SIN, 2005) may be used; however, it may be best if the noise adapts while the speech remains at a constant intensity (i.e., 60 dBA). This accounts for effects of transmitter compression, which will occur during testing because the speech is presented at a typical intensity (i.e., >70 dB SPL at the location of transmitter).

Speech-recognition testing may be conducted in a sound-treated booth or in a classroom setting for students. The single-coned speech loudspeaker should be placed at zero degrees azimuth and at least 3 feet from the listener. One or more noise loudspeakers should be spatially separated from the signal loudspeaker by at least 30 degrees. When the FM/RF system is in use, the microphone of the transmitter should be placed 3 (boom microphones) to 6 (lapel microphones) inches from the center of the single-coned loudspeaker to replicate the distance of the microphone from the talker's mouth.

The results of the two speech-recognition conditions in quiet ought to show negligible differences between the FM/RF and cochlear implant alone conditions or slightly better performance in the FM/RF condition. Conversely, speech-recognition performance in noise should be considerably better in the FM/RF condition when compared to the cochlear implant alone condition. Significant differences between conditions may be determined through the use of 95% confidence intervals, which often are provided in the user manual of the test material. Alternately, an audiologist may wish to collect his or her own normative data when testing in classrooms. If ceiling effects (i.e., 100% correct) occur in any conditions, the signal or noise levels may be adjusted to allow for a more challenging testing situation and for additional headroom for condition comparisons. When favorable results do not occur, the audiologist may consider adjusting the gain of the FM/RF receiver, audio-mixing ratio, or sensitivity of the sound processor. These conditions then can be repeated until the desired results are obtained.

For younger children and individuals with cognitive deficits, it is difficult to perform usual speech-recognition measures without and with FM/RF systems. Instead, the parent or caregiver needs to monitor the functioning of the cochlear implant and FM/RF device once to twice a day using procedures described in the following section on troubleshooting. Additionally, it is imperative that the audiologist receive feedback from the user's caregivers, teachers, speech-language pathologists, and so forth, regarding the child's benefit and performance with the remote microphone RF system. Ideally, this feedback should be sought through both an informal interview process as well as through the completion of evidence-based questionnaires designed to assess a child's listening function in challenging environments.

Some manufacturers and professionals do not recommend FM/RF systems for young children; however, no evidence suggests that a well-fit and monitored FM/RF system will be detrimental to a child's speech and language development. On the contrary, a large body of research supports the use of FM/RF systems for adults and children with cochlear implants for improving speech recognition in quiet and noisy situations. These improvements are likely to aid in speech and language development. In the authors' opinion, the inability of the child to report difficulties with the FM/RF device should not be a reason for nonuse, just as being a good reporter is not a requirement for receiving a cochlear implant. The following section provides several avenues for monitoring and troubleshooting FM/RF and implant devices.

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### Monitoring and Troubleshooting Personal FM/RF Systems for Cochlear Implants

Troubleshooting of personal FM/RF systems may seem to be a challenge; nonetheless, with the correct equipment, audiologists, parents, teachers, and other professionals have numerous ways to verify functioning of the cochlear implant speech processor and FM/RF receiver. To determine proper functioning of the sound processor, the audiologist or parent may use a coil-check wand, transmission symbols or lights on the processor, a Listening Check device (Advanced Bionics), monitor earphones (Cochlear), or Microphone Tester (MED-EL). The Listening Check devices with earphones from Advanced Bionics for Neptune, Harmony, and Naída CI processors and monitor earphones for Cochlear Freedom, Nucleus 5, and Nucleus 6 processors allow a person with normal-hearing sensitivity to listen to acoustic signals presented to the sound processor microphone. To listen to the output of the MED-EL OPUS 2 or RONDO, a personal with normal hearing can use a cable to connect the sound processor to the MED-EL Microphone Tester with earphones.

The signal from the FM/RF system may be verified by conducting a behavioral listening check or by using more objective techniques, which include listening to the signal from an amplifier speaker; extra hearing aid with an audio shoe; headset checker; or sound processor with a Listening Check device, monitor earphones, or Microphone Tester. When conducting a behavioral listening check, the child is asked to repeat words or sounds that he or she hears as a person speaks into the transmitter microphone. This technique verifies functional performance of the FM/RF system and allows the audiologist to determine integrity of the signal once it is processed through the cochlear implant. However, these procedures may not be possible for young children or patients with cognitive delays or disorders.

In rare cases in which an older sound processor is in use or the audiologist does not have access to the listening check devices, the audiologist should verify the signal through some type of amplifier. Several options are available for listening to the signal from the FM/RF system alone. First, when using the Oticon CI-FM Adaptor coupled to a miniaturized FM/RF receiver, the signal may be verified by speaking into the transmitter microphone and listening to the output from the FM/RF receiver through an amplifier. Amplifiers may include an inexpensive, small speaker with a ⅛-inch input (e.g., Radio Shack Mini Audio Amplifier) or an extra hearing aid with the appropriate FM/RF audio shoe (approximately $30–$50). Another amplifier option is a Headset Checker from Phonak that allows for direct connection of a miniaturized RF/FM receiver to headphones. The normal-hearing listener can verify acoustic signals from the FM/RF receiver through the earphones while speaking into the transmitter microphone.

The best way to check an RF/FM signal while connected to a sound processor is with a Listening Check device, monitor earphones, or Microphone Tester. As described previously, all three U.S. manufacturers have devices and/or earphones for listening to the output of the sound processor, and these same devices can also be used to listen to the output from a FM/RF system. Additionally, one Advanced Bionics (Naída CI Listening Check) device as well as Cochlear and MED-EL devices allow the audiologist or parent to hear simultaneously the signals from the speech processor microphone and from the FM/RF transmitter microphone. When checking the RF/FM system, the processor sensitivity may be reduced to verify the FM/RF signal. Conversely, when verifying the signal from the processor microphone, the FM/RF transmitter may be muted, and the sensitivity of the sound processor may be increased. The signal from the device does not represent the final processed signal that the child receives from the cochlear implant; however, these devices allow for verification of the signal from the microphone just prior to entering the sound processor. Additional information about connecting some sound processors and FM/RF systems to Listening Check devices, monitor earphones, or Microphone Testers, may be found in Schafer, Musgrave, et al. (2013).

The devices used to determine the output of the sound processor alone, FM/RF alone, or combined devices may be used to systematically eliminate issues with one or more components of the devices. To verify function of some of the sound processor components, educational audiologists may need to collaborate with a child's cochlear

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implant audiologist. Likewise, to verify function of some of the FM/RM system components, the pediatric cochlear implant audiologist may need to have access to extra FM/RF equipment or will need to collaborate with a child's educational audiologist.

## HAT TO IMPROVE TELEPHONE CONVERSATIONS

Some individuals with cochlear implants successfully converse over the telephone by placing the phone receiver directly over the processor microphone or by using a speaker phone. However, in general, three options exist for improving telephone conversations over what is achieved with the sound processor alone. These options include: (1) direct-audio input of the signal from the phone to the processor with a cable, (2) transmission of the signal from the phone to the processor(s) with a telecoil, and (3) use of a streaming device.

As expected, direct-audio input requires a direct electrical connection between the phone and the sound processor. Each manufacturer provides special cords for connecting sound processors to land lines or mobile phones. The telecoil or streaming options, however, may allow for wireless connections between a land lines/mobile phones and sound processors. The following sections introduce the standard use of telecoil or streaming devices, programming considerations, and accessories for cochlear implants. It should be noted that much of the following discussion also applies to connections to other devices, such as MP3 players, iPads, televisions, and computers.

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### Telecoils

The general procedures for using a telecoil are similar across cochlear implant manufacturers. The first step is to obtain a regular land line telephone or telecoil-compatible mobile phone. All mobile phone manufacturers make telecoil-compatible phones. In fact, as of 2006, the Federal Communications Commission requires that all mobile phone providers have at least two highly rated telecoil compatible phones. After the processor is set to the telecoil setting by using a push button, processor remote control, or automatic telecoil, the telephone conversation may begin. During the conversation, the person may need to adjust the location of the telephone receiver relative to the telecoil (i.e., above or behind the sound processor) to obtain the best orientation and signal from the phone receiver.

As mentioned previously, most contemporary sound processors have built-in telecoils. The telecoil options discussed in this section also are applicable to the use of induction-loop HAT with a room loop or neckloop via FM, digital RF, or infrared technology. In fact, several of the external devices, as described in the following section, use an induction neckloop to send the signal from the phone to the sound processor.

The Advanced Bionics Harmony and Naída CI processors have built-in telecoils and several telecoil settings the audiologist may select from in the programming software (e.g., 50/50, 30/70). Through programming, the audiologist needs to make the telecoil active and its function accessible in one of the switch positions on the sound processor. Users of Harmony processors may consider using the telecoil adaptor plugged into the Direct Connect Earhook when no program is available for the built-in telecoil or when the user does not desire to use a program position for the built-in telecoil. The previous generation processor, the Auria, does not have a built-in telecoil, but a Telecoil Adaptor may be connected to the processor via the Direct Connect Earhook and cable. The Neptune processor does not have a built-in telecoil, but recipients may choose to use a T-Comm accessory, which has a built-in telecoil. The T-Comm comprises of a cable, earhook, and T-Mic (i.e., microphone that is placed near the opening of the ear). Individuals with Neptune processors may also use a telephone adaptor or a patch cable to connect directly to the phone via the Neptune Connect configuration.

The Cochlear Nucleus 5/6 and Freedom processors have built-in telecoils. The Nucleus 5 and 6 telecoils can be accessed manually via a push-button, activated with the Remote Assistant, or

automatically engaged (Auto Telecoil) when in the presence of elecgtromagnetic signals from a telephone or induction neckloop. On the Freedom processor, telecoil access may be placed in one of the program slots, and similar to the Harmony, the same audio-mixing ratios are available as those for FM/RF use.

The MED-EL OPUS 2 and RONDO have a FineTuner remote control, which allows the user to select from three signal-source settings: microphone only, telecoil only, or microphone plus telecoil (1:1 ratio).

## Telecoil Accessories

Several telecoil accessories may be coupled to sound processors with specialized cables. These products are sold by many HAT vendors. Some systems utilize Bluetooth technology along with a neckloop (e.g., ClearSounds Bluetooth Amplified Neckloop) or headset (e.g., Motorola Bluetooth Headset). To use either device, the user needs a mobile phone with Bluetooth capabilities as well as an active built-in telecoil or telecoil adaptor. The speech signal from the telephone is sent via Bluetooth to the Bluetooth receiver worn by the listener. The receiver decodes the signal and sends it through the neckloop or headset. Finally, the telecoil that is built-in or connected to the sound processor detects and decodes the electromagnetic signal from the neckloop/headset and sends it through the processor. Most of these devices also include a microphone, allowing the user to have hands-free conversations.

The second device, a phone neckloop, functions in the exact same manner as the Bluetooth device, but a wired connection exists to the cell or land-line phone. The neckloop will plug into a 2.5- or 3.5-mm jack on the phone. In addition to the neckloop, some devices have a microphone to allow for two-way conversations (i.e., listening and speaking), whereas other devices do not have the microphone (i.e., audio-only neckloops).

Another telecoil accessory, the silhouette, is a lightweight, wafer-thin earpiece that creates an induction connection from the cell or land-line phone to the sound processor. It functions similarly to neckloops whereby the silhouette, which is electrically connected to land line or cell phones, transmits electromagnetic information to the sound processor telecoil.

## Streaming Devices

The newest options for improving telephone connectivity and conversations are available with streaming options in the Advanced Bionics Naída CI and Cochlear Nucleus 6 sound processors. First, users of bilateral Naída CI may use binaural streaming technology (i.e., Binaural VoiceStream) to hear phone conversations in both ears simultaneously. Specifically, when using the audiologist-programmable feature DuoPhone, the user places the phone up to one processor, and the conversation is automatically streamed to the other processor. Second, individuals with unilateral or bilateral Naída CI processors may use the Phonak ComPilot streaming device around the neck, which allows for Bluetooth connectivity to Bluetooth enabled mobile phones and other devices. The signal from the ComPilot to the sound processors may be streamed to one ear of a unilateral recipient, two ears of a bilateral recipient, or two ears of a bimodal user when a compatible Phonak hearing instrument is used on the nonimplant ear. The ComPilot may also be used in conjunction with the Phonak TVLink to provide streaming of TV signals to the Naída CI processor(s) or the Phonak RemoteMic to provide improved speech recognition in noisy situations. It is important to note, however, that the Phonak RemoteMic uses Bluetooth streaming, which is not the same type of digital RF transmission utilized by the Dynamic Phonak Roger transmitters. There are no studies comparing implant users' performance obtained with the RemoteMic with ComPilot compared to the more traditional FM and digital RF systems; however, the authors of this book hypothesize superior performance in noise for the digital RF systems given the use adaptive increases in gain provided by Dynamic digital RF/FM.

The Cochlear Nucleus 6 sound processor is able to connect to a range of wireless audio accessories, including a Mini Microphone (remote mic),

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TV streamer, and other Bluetooth-enabled devices, without the use of a specialized streaming device. The Mini Microphone and TV streaming accessories will also be compatible with some ReSound hearing aids for users of bimodal stimulation. The reader is referred to Chapters 2 and 5 in this book for more information on the Cochlear wireless streaming accessories.

## SUMMARY OF HAT FOR COCHLEAR IMPLANTS

Despite the advances in cochlear implant technology, users continue to experience difficulties with speech perception in noise and over the telephone. However, as discussed in this chapter, several types of HAT are available to improve speech recognition in noise and telephone conversations of people with cochlear implants. When considering types of HAT for speech recognition, evidence-based research supports the use of personal FM/RF systems that electrically couple to sound processors. However, some research supports the use of electromagnetic coupling via neckloop receivers. When the FM/RF system is successfully connected to the sound processor, several programming parameters may need adjusting including the gain of the FM/RF receiver and audio-mixing ratios, IDR, and sensitivity of the sound processor. These parameters have a significant influence on speech recognition in quiet and in noise; therefore, the audiologist will need to use the most up-to-date research and individualized testing to determine the optimal settings for a patient.

HAT for improving telephone conversations and hearing audio from other devices (e.g., iPad, MP3 player) may involve direct-audio input of the signal from the phone to the sound processor, use of a telecoil that is built-in or plugged into the sound processor, or use of a streaming device. Through electromagnetic transmission, the telecoil allows the user to hear land-line or telecoil-compatible cell phone conversations through the implant. Several telecoil accessories also are available to transmit the sound from a land line or cell phone through the use of electromagnetic (i.e., neckloop or silhouette) or Bluetooth transmission. Binaural streaming and a streaming device, such as the Phonak ComPilot, allow for improved hearing over the phone and connectivity to Bluetooth-enabled devices. All three types of HAT may offer particular improvement in telephone performance for bilateral cochlear implant recipients and users of bimodal stimulation, as it may allow these users to listen to telephone output with both devices simultaneously.

## KEY CONCEPTS

The reader should understand how to use HAT with cochlear implants and understand the following key concepts:

- Speech perception in noise of people using unilateral or bilateral cochlear implants is improved significantly when using personal HAT, such as an FM/RF system, relative to performance with an implant alone.
- The most commonly used personal HAT for people with cochlear implants are digital RF, FM, and induction loop systems.
- There are several manufacturer-specific programming considerations for sound processors when coupling to personal HAT.
- The primary ways to improve telephone conversations with a cochlear implant are through the use of a telecoil or a streaming device, which may require special programming or adaptors for the sound processor or additional equipment.

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# 10

# Programming Recipients Using Electric-Acoustic Stimulation

## INTRODUCTION

Traditional acoustic hearing aids may be somewhat beneficial to patients with precipitously sloping or ski-slope sensorineural hearing loss configurations, but the provision of high-frequency gain may not provide improvement in speech recognition in quiet and in noise to these listeners (Cox, Johnson, & Alexander, 2012; Hogan & Turner, 1998; Hornsby & Ricketts, 2006; Stelmachowicz, Pittman, Hoover, Lewis, & Moeller, 2007). Additionally, research has shown that the aided word speech recognition of listeners who have precipitously sloping, severe high-frequency hearing losses is often fair to poor with the use of high-frequency amplification (Glista et al., 2009; Simpson, McDermott, & Dowell, 2005; Simpson, Hersbach, & McDermott, 2006). Patients who have severe-to-profound hearing loss in the high-frequency regions, better residual hearing in the low-frequency regions, and difficulty communicating with hearing aids, may be candidates for combined electric-acoustic stimulation (EAS) from a hybrid cochlear implant.

The primary premise of a hybrid cochlear implant is to provide electrical stimulation to the auditory nerve for high-frequency inputs. Additional objectives of hybrid cochlear implantation include the preservation of low-frequency residual hearing for recipients with the aforementioned audiometric profile and the provision of acoustic amplification to enhance the user's ability to hear low-frequency inputs. In general, hybrid cochlear implants differ from conventional cochlear implants in several respects. First, hybrid cochlear implant systems typically possess an electrode array that differs in many ways from the manufacturer's conventional electrode array. Hybrid cochlear implant electrode arrays are shorter in length and narrower in diameter than conventional electrode arrays (Figure 10–1). Furthermore, a hybrid implant electrode array is flexible and designed for lateral wall placement (rather than perimodiolar placement) to minimize trauma during insertion and to avoid the delicate cochlear structures of the organ of Corti. It should be noted, however, that some reports describe almost complete preservation of low-frequency hearing for some recipients who have received conventional, long electrode arrays and/or perimodiolar electrode arrays (James et al., 2005; Kiefer et al., 2004; Prentiss, Sykes, & Staecker, 2010). In short, although hearing preservation is probably more likely with shorter, lateral wall electrode arrays,

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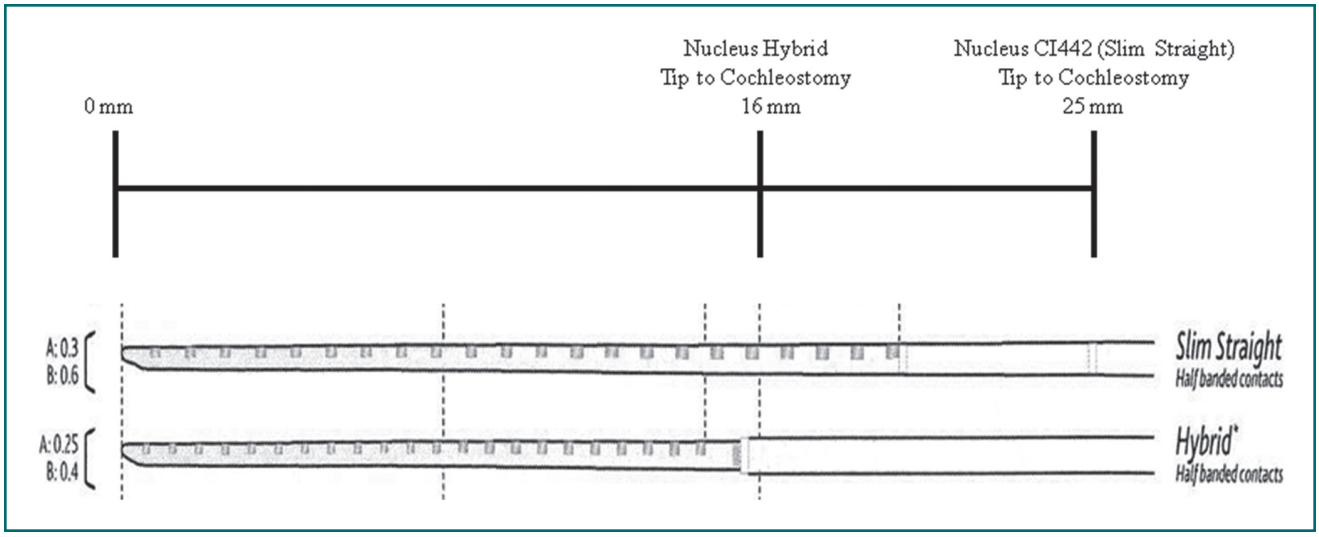



**FIGURE 10–1.** Length comparison between hybrid cochlear implant electrode arrays and conventional electrode arrays. *Note*. Provided courtesy of Cochlear™ Ltd.

these do not appear to be requisite characteristics for hearing preservation for some recipients. The variables associated with the preservation of low-frequency hearing in implant recipients are described in more detail later in this chapter.

The external sound processor of a hybrid implant system features an acoustic component (i.e., an acoustic receiver) used to deliver low-frequency amplification. As shown in Figure 10–2, the acoustic component is typically coupled to the sound processor with a thin cable. However, some EAS users have worn a separate custom in-the-ear hearing aid along with a conventional cochlear implant in order to amplify low-frequency acoustic inputs.

To aid in hearing preservation, surgeons have developed refined surgical techniques for hybrid electrode array insertion. The exact surgical techniques used for hybrid cochlear implant surgery differ substantially across surgeons. Some surgeons prefer to insert the electrode array via cochleostomy whereas others prefer to enter the scala tympani through the round window. When a cochleostomy approach is used, the surgeon selects a small (e.g., 1 mm) diamond burr and slow drill speed (e.g., 5,000–10,000 revolutions per minute) to access the cochlea. Bone is carefully excavated to expose endosteum, which is then gently opened using a metal pick. With a round window approach, the surgeon will often remove the bony overhang near the round window to visualize the



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**FIGURE 10–2.** External sound processor of a hybrid implant system. *Note*. Provided courtesy of Cochlear™ Ltd.

round window membrane. Then, a pick is used to enter the round window membrane.

With either a round window or cochelostomy technique, the surgeon uses copious amounts of irrigation and suction prior to making the cochleostomy or entering the round window in order to ensure that blood, bone, and other debris do not enter the cochlea. Suction is avoided once the cochlea is opened in an attempt to reduce the possibility of a noise-induced hearing loss and disturbance of cochlear fluids. Finally, intraoperative steroid administration is gaining favor in an attempt to reduce cochlear inflammation that may be related to postoperative changes in acoustic hearing. It is worth noting that, due to the success of recipients of hybrid cochlear implants, many surgeons now employ these trauma-reducing surgical techniques when inserting conventional cochlear implant arrays.

Theoretically, the use of low-frequency acoustic hearing could provide the user with better speech-recognition performance in noise, temporal processing (e.g., fine temporal structure), spectral resolution, and music appreciation/perception when compared to a traditional, long-electrode cochlear implant. It should be noted that some patients who receive hybrid implants lose their low-frequency acoustic hearing (Fitzgerald et al., 2008; Gantz et al., 2009; Gstoettner et al., 2009; Reiss, Perreau, & Turner, 2012; Skarzyński, Lorens, Piotrowska, & Anderson, 2007). However, these recipients typically achieve better speech recognition with the use of their hybrid cochlear implant, particularly when used in concert with a contralateral hearing aid, when compared to preimplant performance with binaural hearing aids.

The goal of this chapter is to provide a brief overview of hybrid cochlear implant fundamentals and hardware, to review recent research addressing performance of recipients of hybrid implants, and to discuss recommended procedures for programming and optimizing the fitting of these devices. For a more in-depth view of the development of EAS, the reader is referred to several reviews addressing the topic (Mowry, Woodson, & Gantz, 2012; Turner, Reiss, & Gantz, 2008; von Ilberg, Baumann, Kiefer, Tillein, & Adunka, 2011; Woodson, Reiss, Turner, Gfeller, & Gantz, 2011).

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## ELECTRODE ARRAY LOCATION AND PLACE OF STIMULATION

Prior to a discussion of hybrid cochlear implant hardware, it is important to review the relationship between electrode array location, the site of stimulation in the cochlea, and the frequency of maximum stimulation. The relationship between cochlear place of stimulation and frequency is well established (Greenwood, 1961, 1990; Kawano et al., 1996; Stakhovskaya, Sridhar, Bonham, & Leake, 2007). The Greenwood function correlates the position of hair cells in the cochlea's organ of Corti to the frequencies that give rise to stimulation of corresponding auditory nerve fibers (Greenwood, 1961, 1990). It essentially serves as a map of tonotopic organization along the basilar membrane and has often been used to estimate the frequency range that should be allocated to an electrode located at a specific site (Kiefer et al., 2005; Skarzyński et al., 2007).

However, for multiple reasons, the Greenwood function does not entirely explain the relationship between electrode location and frequency of neural stimulation. For instance, this relationship is substantially influenced by whether the electrode array is designed for lateral wall or perimodiolar placement. Perimodiolar electrodes are designed to deliver stimulation directly to the spiral ganglion bodies of the auditory nerve, so the Greenwood function may not serve as a precise estimate of the frequency that is stimulated by the implant. Lateral wall electrodes are theoretically more inclined to deliver stimulation to a location that is more proximal to the radial nerve fibers innervating the organ of Corti (i.e., adjacent to the basilar membrane), and as a result, the Greenwood function may provide a closer approximation of the frequency place of stimulation.

The pioneering work of Stakhovskaya and colleagues showed that the length of the organ of Corti is much longer (30.5 to 36.87 mm) than the length of the spiral ganglion measured at the center of Rosenthal's canal (12.54 to 14.62 mm) (Stakhovskaya et al., 2007). As a result, a perimodiolar electrode array does not need to be inserted nearly as deep as a lateral wall electrode in order to elicit

a particular frequency of neural stimulation. For example, Stakhovskaya and colleagues estimated that a lateral wall electrode array would have to be inserted about 20 mm into the scala tympani to elicit an auditory neural response at 1000 Hz, but in contrast, a perimodiolar electrode array would only have to be inserted about 14 mm into the scala tympani to evoke the same frequency of stimulation. Although the two different types of electrode arrays are inserted to different depths of the scala tympani, they have both reached approximately the same angular depth of about 360 degrees (re. a reference point of 0 degrees at the round window). It is important to note that Stakhovskava and colleagues (2007) reported that significant intersubject variability existed in the length of the organ of Corti and spiral ganglion. With that in mind, they recommended the use of angular insertion depth as the metric of choice for determining appropriate cochlear implant insertion depth. The interested reader is referred to the publication by Stakhovskava and colleagues for the estimated frequency of stimulation for various angular insertion depths.

It is also important to note that the exact frequency that is stimulated in the auditory system is likely dependent upon the extent of degeneration of auditory nerve fibers, particularly in the case of lateral wall electrodes. Research has suggested that radial dendrites often degenerate secondary to deafness (McMullen & Glaser, 1988; McMullen, Goldberger, Suter, & Glaser, 1988), and as a result, electrical current may have to spread to the spiral ganglion fibers to elicit a response. In such an event, the Stakhovskava et al. (2007) data correlating frequency to spiral ganglion length may be a better predictor of the frequency of stimulation than the Greenwood function. Additionally, innervation of the cochlea becomes more complex and variable at the apical end, so it is more difficult to predict the frequency of the response. Also, cross-turn stimulation (i.e., the spread of electrical current from one turn of the cochlea to a neighboring turn) may elicit complex frequency responses/pitch percepts (Briaire & Frijns, 2006). Furthermore, the stimulation of neighboring neural elements may be influenced in complex fashions by the growth of fibrous tissue and/or bone in the cochlea following surgery (Richter et al., 2001). In short, predicting the characteristic frequency of stimulation at a specific electrode contact location in the cochlea is quite complicated, is dependent on a number of factors related to the design/configuration of the electrode array and the anatomy of the peripheral auditory system, and is likely to be variable across recipients.

## CURRENT DEVICES

Two implant manufacturers, Cochlear and MED-EL, have hybrid devices. At the time of this writing, the Cochlear™ Nucleus® Hybrid™ Implant System is the only hybrid system approved for commercial use by the United States Food and Drug Administration (FDA). The MED-EL hybrid cochlear implant system is currently being evaluated in a multicenter clinical trial, and the data from that study are likely to be submitted to the FDA for review and approval of the device for commercial use with adults with precipitously sloping high-frequency hearing loss in the near future. In the following paragraphs, a review will be provided of all of the hybrid cochlear implant technology developed by Cochlear and MED-EL for use in clinical trials or for commercial use either in the USA or in Europe.

### Cochlear Hybrid Implant Technology

In the late 1990s, Cochlear Ltd. began a collaboration with Bruce Gantz and colleagues at the University of Iowa to develop a hybrid cochlear implant system. The initial device developed from this partnership was known as the Nucleus Hybrid S8 implant and possessed six intracochlear stimulating contacts and two extracochlear reference electrodes. The S8 electrode array is very thin with a cross-sectional diameter of 0.4 × 0.25 mm. It was a 10 mm long straight electrode array that was coupled to the Nucleus CI24M receiver/stimulator and later incorporated the Freedom-based receiver/stimulator. Recipients used the

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cochlear implant along with a custom in-the-ear hearing aid to achieve EAS.

In a USA clinical trial, a total of 87 subjects were implanted with the 10 mm Hybrid S8 cochlear implant. Reports suggested that use of the Hybrid S8 implant was associated with an impressive rate of preservation of residual hearing (about 90% experienced some preservation of their residual hearing), and the majority of subjects achieved significant improvement in monosyllabic word recognition and sentence recognition in quiet and in noise (Gantz & Turner, 2004; Gantz, Turner, & Gfeller, 2004; Gantz, Turner, Gfeller, & Lowder, 2005). However, some of these recipients did lose their residual hearing, and although most performed better in the bimodal condition (i.e., electrical stimulation with short array in the implanted ear and acoustic stimulation from hearing aid in the opposite ear) than they did preoperatively with binaural hearing aids, 17 subjects were reimplanted with longer electrode arrays. There continues to be uncertainty regarding the extent of benefit a 10 mm electrode array provides relative to a conventional-length electrode array when acoustic hearing is lost in the implanted ear.

Cochlear also developed and conducted clinical trials with the Nucleus Hybrid S12 implant, which contained 10 intracochlear electrode contacts positioned along a 10 mm straight electrode array with the same cross-sectional dimensions as the Hybrid S8. In the USA, 24 subjects were implanted with the Hybrid S12 device, and like the S8 study, these subjects typically experienced improvements in speech recognition with the S12 device relative to their preimplant performance (Gantz & Turner, 2011; Gantz, Turner, & Gfeller, 2006; Gantz et al., 2009). However, approximately 30% of the subjects experienced at least a 30 dB shift in their low-frequency residual hearing, and also as with the S8 device, the shallower insertion depth of the S12 array may have limited overall performance for these subjects. Once again, the merits and limitations of 10 mm electrode arrays are still being explored. Investigators will continue to weigh the benefits of hearing preservation associated with a shorter array against the potential limitations of reduced performance that may arise if acoustic hearing is lost and the recipient is forced to rely on hearing across the entire speech frequency range with a short array.

In an effort to balance the goals of providing adequate mid-to-high-frequency electrical coverage along with a relatively high likelihood of preservation of low-frequency residual hearing, Cochlear developed the Nucleus Hybrid L24 cochlear implant (Figure 10–3), which recently received approval for commercial use by the U.S. FDA (March 2014). The commercial name of the Nucleus Hybrid L24 device is the Nucleus Hybrid cochlear implant. However, in this chapter, the authors will continue to refer to the device as the Nucleus Hybrid L24 device in order to prevent confusion with the earlier model Nucleus hybrid cochlear implants. The Nucleus Hybrid L24 implant features the Nucleus Freedom receiver/stimulator coupled to a thin electrode array designed to be inserted 16 mm into the cochlea. The L24 electrode array has a cross-sectional diameter of .25 by .35 mm at the apical end of the array and .4 by .55 mm at the basal end of the array. The



**FIGURE 10–3.** Nucleus Hybrid L24 cochlear implant. *Note*. Provided courtesy of Cochlear™ Ltd.

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thin diameter of this array makes it ideal for round window or cochleostomy insertion. Furthermore, the L24 electrode array possesses 22 half-banded electrode contacts, which are distributed over 15 mm and are designed to point toward the modiolus. Use of half-banded electrode contacts, rather than full ring electrodes, allows for a thinner array carrier and also results in a smooth surface, which may reduce resistance during insertion and promote hearing preservation.

The L24 array is intended to be inserted along the lateral wall of the scala tympani, and like the S8 and S12 devices, it possesses a "stopper" that prevents insertion beyond a desired depth, which in the case of the L24 is 16 mm (i.e., the medial end of the stopper is 16 mm from the tip of the array), which corresponds to an angular insertion depth of about 235 to 270 degrees for a typical round window or cochleostomy insertion, respectively. It also features a fan-shaped handle at the base of the array that is intended to facilitate ease of insertion. Finally, the electrode lead wires are 20 micrometers in diameter, and as a result, the array is quite flexible at the apical end and relatively stiff at the basal end. This characteristic is intended to reduce trauma during insertion, while also enhancing the surgeon's ease of handling.

The L24 hybrid implant system initially included the Nucleus Freedom hybrid sound processor (Figure 10–4). The Freedom hybrid sound processor possesses an acoustic driver at the apex of the processor, which allows for removal of the conventional earhook and a connection to a cable that is coupled to a custom-made receiver in-the-ear (RITE), referred to as the acoustic component. The electroacoustic specifications of this device are provided in Table 10–1.

In 2013, the Nucleus 6 sound processor (see Figure 10–2) was introduced as the contemporary EAS sound processor for use with the Nucleus L24 hybrid implant. The Nucleus 6 sound processor is



**FIGURE 10–4.** Nucleus Freedom hybrid sound processor. *Note*. Provided courtesy of Cochlear™ Ltd.

**Table 10–1.** Comparison of Electroacoustic Characteristics between the Freedom Hybrid and Nucleus 6 Hybrid (CP900)

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| | *Freedom Hybrid* | *Nucleus 6 Hybrid* |
|---|---|---|
| **SPA Output at 90 dB SPL (OSPL90)** | 111.0 dB SPL | 117.1 dB SPL |
| **SPA Full-On Gain at 50 dB SPL (FOG)** | 40.4 dB SPL | 50.6 dB SPL |
| **Reference Test Gain at 60 dB SPL** | 33.2 dB SPL | 41.4 dB SPL |
| **Frequency Range** | <200 Hz and 1491 Hz | <200 Hz and 1781 Hz |
| **Equivalent input noise level (EIN)** | 37.5 dB SPL | 27.1 dB SPL |

*Note*. Special Purpose Average (SPA) frequencies are: 500, 800, and 1250 Hz.

described in detail in Chapter 1. Like the Freedom hybrid sound processor, the Nucleus 6 sound processor possesses an acoustic driver, which may be coupled to a thin wire that connects to an acoustic RITE. The RITE may be coupled to a generic dome or to a custom shell, which may be ordered from an ear mold laboratory. The electroacoustic specifications of this device are provided in Table 10–1. As shown, the maximum power output (MPO), gain, frequency response, and equivalent input noise (EIN) of the Nucleus 6 acoustic component are all superior to that of the Freedom acoustic component.

The clinician may adjust several programmable parameters for the acoustic component of both the Freedom and Nucleus 6 sound processors. Both acoustic components possess multiple frequency bands to allow for precise shaping of the low-frequency response. Specifically, the clinician may adjust overall MPO and gain and may also adjust these two parameters within each frequency band. Additionally, the clinician may control the crossover frequency that delineates the starting point of electric stimulation, the range over which acoustic amplification is provided, and the delay of the acoustic signal may also be adjusted. It should be noted that the delay of the acoustic signal is rarely adjusted.

Although it is not formally considered as a hybrid cochlear implant, the Nucleus CI422 implant (e.g., slim research array—see Figure 1–25 in Chapter 1) was designed to be an atraumatic array and has also been considered as an EAS device (Skarzyński et al., 2012; Skarzyński et al., 2014). As mentioned in Chapter 1, the CI422 device is a straight array designed for lateral wall insertion. It possesses a cross-sectional diameter of 0.3 × 0.35 mm and 0.5 × 0.6 mm at the apical and basal ends, respectively, which makes it relatively narrow in comparison to the Contour Advance perimodiolar array. Additionally, the CI422 possesses two "markers" to guide the surgeon in achieving an insertion depth of approximately 20 or 25 mm. A 20 mm insertion depth for a lateral wall electrode generally results in about a 330 to 360 angular degree insertion depth, whereas a 25 mm insertion depth results in a 430 to 480 angular degree insertion depth, depending on insertion technique.

## MED-EL EAS Technology

European surgeons working with MED-EL implants pioneered the concept of EAS (Kiefer et al., 2004; Skarzyński et al., 2002). Some of the earliest attempts to provide EAS were with the MED-EL COMBI40+ implant with the standard electrode array (31.5 mm in length) (Kiefer et al., 2004; Skarzyński et al., 2002). For instance, Skarzyński et al. (2002) found preservation of at least some low-frequency residual hearing in 21 of 26 recipients who were implanted with a deeply inserted MED-EL COMBI40+ Standard array using surgical techniques that intended to reduce trauma to the cochlea. Kiefer and colleagues (2004) also reported impressive hearing preservation results with the MED-EL COMBI40+ Standard array. Kiefer et al. (2004) inserted the MED-EL COMBI40+ Standard array and a modified version of the Standard array at a shallower depth (19–24 mm) in 14 adults with air conduction thresholds in the 20 to 60 dB HL range from 125 to 500 Hz. Hearing was preserved in 12 of 14 recipients, and the mean decrease in air conduction thresholds were 10, 15, 17.5, and 5 dB at 125, 250, 500, and 1000 Hz, respectively. The results of the Kiefer et al. (2004) study elicited interest in the concept of providing a shallower insertion (18–24 mm) to better facilitate hearing preservation, and as a result, surgeons began to turn their focus toward the use of shorter insertion depths for cases in which hearing preservation was a concern.

At that time, MED-EL also offered the Medium electrode array, which was developed for cases in which a sufficient insertion of the Standard array could not be achieved (i.e., the recipient had a shorter cochlea than normal; the recipient was receiving a MED-EL implant as a replacement for a faulty competitor's implant with had a shorter array; the recipient had fibrous tissue growth prevented a full insertion with the standard array). The distance between the most basal electrode and the tip of the Medium array is approximately 22 mm, which is considerably shorter than for the Standard array (approximately 28 mm). The Standard and Medium electrode arrays each include 12 electrode contacts, but the distance between electrode contacts is shorter for the Medium array (1.9 mm) than for the Standard array (2.4 mm).

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European surgeons who were exploring the feasibility of EAS logically began to use the Medium array to achieve a shallower insertion for recipients who had a substantial amount of functional low-frequency hearing. Insertion of a lateral wall electrode to a depth of 20 mm results in a 360-degree angular insertion depth in the typical cochlea, which would reach to the 1000 Hz place in the cochlea. One research study has suggested that a significant reduction in cochlear trauma is possible with angular insertion depths of 360 degrees and less (Adunka & Kiefer, 2006).

Gstoettner et al. (2008) reported on results from a multicenter study in which the Medium array was implanted in 18 adults with considerable low-frequency hearing and profound high-frequency hearing loss. The Medium array was inserted 18 to 22 mm into the cochlea, and steroids were administered to increase the likelihood of hearing preservation. Gstoettner and colleagues (2008) attempted to provide acoustic amplification at all frequencies with an air conduction threshold of 80 dB HL or better. Three subjects lost all of their residual hearing following the surgery. Another three subjects had some hearing preservation, but their postoperative hearing loss was too severe to consider the provision of low-frequency amplification. Twelve of the 18 subjects (66.6%) did have sufficient postoperative low-frequency hearing to warrant consideration of amplification, but six of these subjects felt that use of the acoustic amplification was detrimental or provided no benefit. The authors noted that the subjects had to use a separate hearing aid with their cochlear implant sound processor, and many may have rejected EAS because use of two devices was cumbersome and inconvenient. Indeed, much like the first subjects in the Gantz Hybrid studies, the earliest MED-EL EAS recipients typically used the TEMPO+ sound processor with an in-the-ear hearing aid to achieve EAS. In response to this limitation, MED-EL developed the DUET EAS/Hybrid processor. Gstoettner et al. (2008) reported significant improvement in performance for the six EAS users relative to their preoperative performance, but performance with EAS was not compared to performance with electric-only stimulation.

Early experiences with the Standard and Medium arrays proved that hearing preservation was possible with MED-EL electrodes. However, surgeons and MED-EL scientists, engineers, and researchers realized that the potential of hearing preservation could be further improved with the development of an electrode array that is specifically designed to promote atraumaticity and hearing preservation. In response to these early reports of the benefits of EAS, MED-EL developed the FLEX electrode technology, which possesses several features that are specifically intended to facilitate to reduce cochlear trauma and enhance the potential of hearing preservation. The unique features of FLEX arrays are as follows:

1. The five most apical stimulation sites consist of single electrode contacts, which reduce the diameter and increase the flexibility/flaccidity of the apical end of the array. The reduction in the diameter of the apical end of the array (i.e., tapering) is important, because the scala tympani becomes narrower as it courses from the basal end to the apex. A smaller apical diameter reduces the chance that the electrode will come in contact with the delicate structures of the organ of Corti. Also, the enhanced flexibility of the apical end reduces the likelihood of cochlear trauma during the insertion process. It should be noted that seven most basal stimulation sites are each composed of two oval-shaped electrode contacts, which is similar to what is found for all 12 stimulation sites of the Standard array.
2. The diameter of the FLEX SOFT Array is narrow (0.36 × 0.48) at the apical tip of the array and increases to 1.3 at the basal end where the electrode array resides in the round window or cochleostomy. These narrow dimensions compare favorably to most any other electrode array available for commercial use at the time of this writing. The tapered design of the FLEX array, which is shown in Figure 1–61 in Chapter 1, allows for increased mechanical flexibility at the tip of the array, insertion into the narrow lumen of the most apical region of the scala tympani, and ease of handing at the basal end on the part of the surgeon.
3. The wires (e.g., electrode leads) in the FLEX series of MED-EL electrode arrays are configured in a wave-shaped pattern, which

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further enhances flexibility and promotes atraumaticity.

At the time of this writing, the FLEX$^{24}$ and FLEX$^{28}$ electrode arrays were both available for commercial distribution in the United States. The FLEX$^{24}$ is a 24-mm-long electrode array with 12 intracochlear stimulating sites. The distance between the most basal electrode contacts and the tip of the FLEX$^{24}$ array is 20.9 mm (Figures 10–5 and 10–6). The cross-sectional diameter of the array ranges from 0.3 to 0.49 mm at the apex to

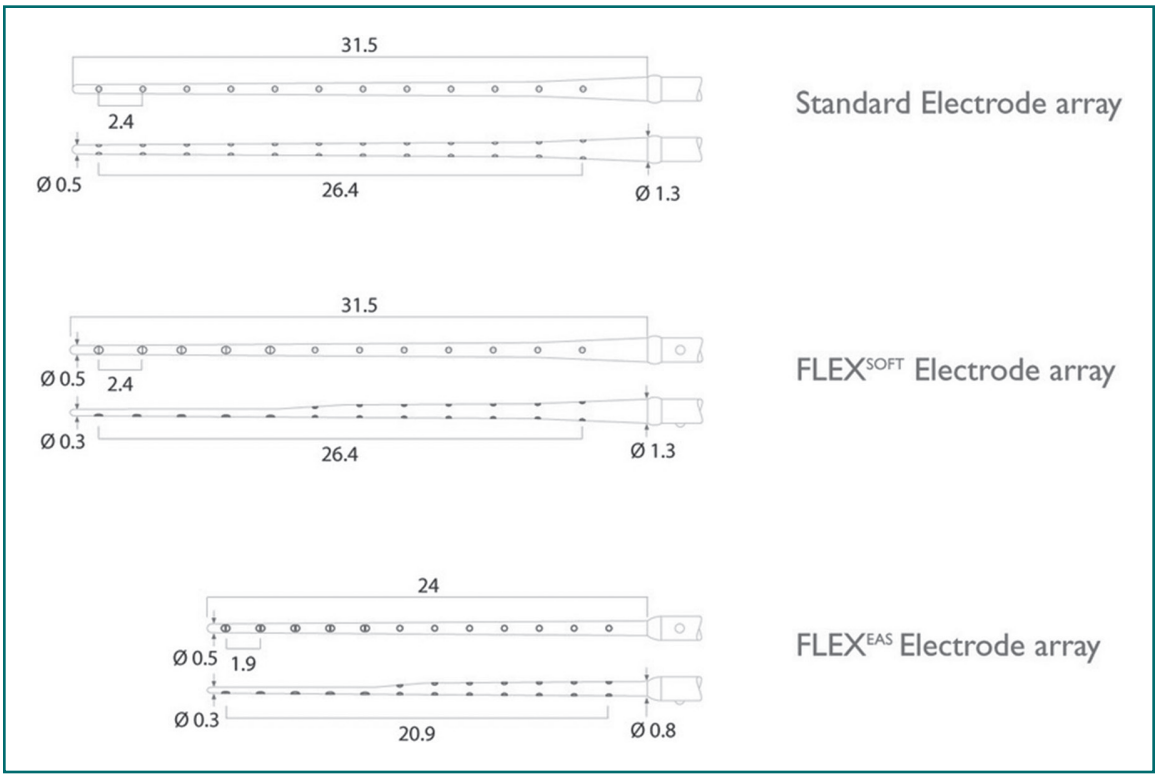



**FIGURE 10–5.** MED-EL electrode arrays. *Note.* Image courtesy of MED-EL Corporation.

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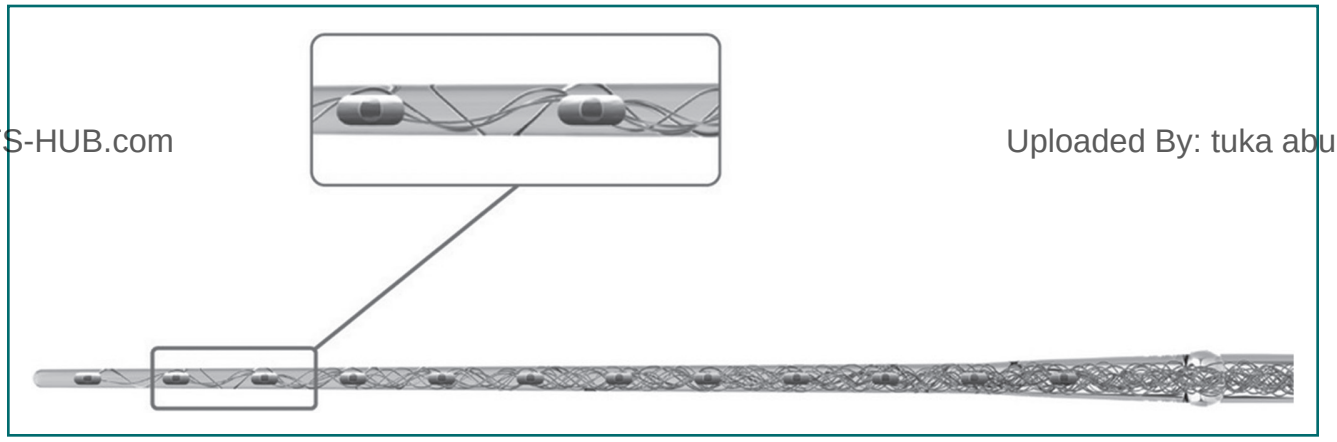


**FIGURE 10–6.** MED-EL wave-shaped wires in the electrode array facilitate flexibility to reduce insertion trauma. *Note.* Image courtesy of MED-EL Corporation.

0.8 at the basal end near the cochleostomy. The electrode leads are arranged in a "wave-shaped" pattern to promote flexibility, particularly at the apical end of the array (Figure 10–7). The five most apical stimulating sites consist of single, flat electrode contacts, a design which reduces the diameter of the apical end of the array and adds to the flexibility (Figure 10–8). The seven most basal stimulating sites are each composed of a pair of electrode contacts, which is similar to the conventional MED-EL arrays. The arrangement of electrode contact pairs at the basal end enhances the stiffness of the basal portion of the array. The flexible apical end paired with the relatively stiffer basal end is designed to allow for easy handling on the surgeon's part while also allowing for greater flexibility, which should reduce insertion trauma. The FLEX$^{24}$ array is designed to be inserted along the lateral wall of the cochlea. Reports have suggested a higher likelihood of hearing preservation with the FLEX$^{24}$ when it is inserted in the round window rather than a cochleostomy (Adunka et al., 2004; Gstoettner et al., 2004). The MED-EL FLEX$^{24}$ electrode array has also been referred to as the FLEX$^{EAS}$ array.

MED-EL also offers the FLEX technology in the FLEX$^{28}$ electrode array and the FLEX$^{SOFT}$ array. The FLEX$^{28}$ is available for use in the United States, whereas the FLEX$^{28}$ and FLEX$^{SOFT}$ are both available in most other parts of the world. The FLEX$^{28}$ and FLEX$^{SOFT}$ electrodes are longer than those typically used for EAS applications, but they do contain the FLEX technology to promote cochlear atraumaticity and hearing preservation. The reader is referred to Chapter 1 for a detailed description of the FLEX$^{28}$ and FLEX$^{SOFT}$ arrays. It should also be noted that at the time of this writing, MED-EL was also developing a FLEX$^{20}$ electrode array, which features the FLEX technology but is 20 mm in length.

At this time, there is no clear consensus as to what is the ideal insertion depth to promote hearing preservation and optimize speech recognition with EAS. Prentiss et al. (2010) evaluated hearing preservation in 18 recipients implanted with the MED-EL PULSAR CI$^{100}$ with Standard or Medium arrays inserted to varying depths ranging from 20 to 28 mm. The examiners employed surgical techniques that were designed to facilitate hearing preservation. All subjects were implanted with a round window insertion. Some residual hearing was preserved in all 18 subjects, and no clear

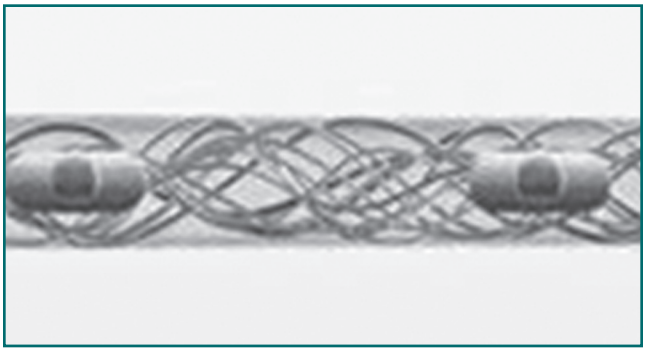


**FIGURE 10–7.** MED-EL electrode leads arranged in a zig-zag pattern to promote flexibility. *Note.* Image courtesy of MED-EL Corporation.

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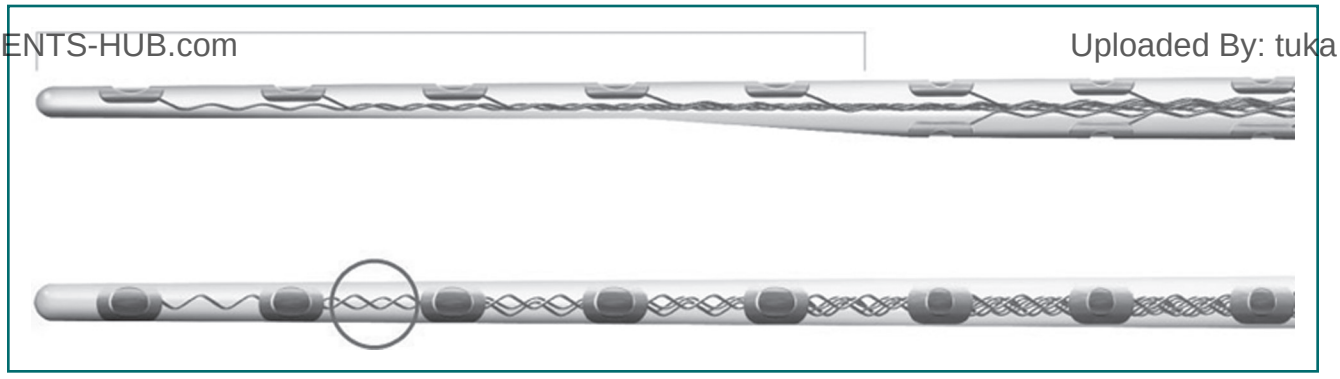


**FIGURE 10–8.** Tapered apical end of the MED-EL electrode array. *Note.* Image courtesy of MED-EL Corporation.

relationship was observed between the insertion depth and the degree of hearing preservation. As a result, the authors suggested that soft surgical techniques may be a primary factor responsible for hearing preservation, and reasonable hearing preservation may be achieved with deep insertions if soft surgical techniques are employed.

Baumgartner et al. (2007) reported on hearing preservation outcomes for 23 adults who had severe to profound hearing loss and were implanted with the FLEX$^{SOFT}$ array. The array was inserted 31.5 mm into the cochlea for 19 of 23 subjects. Shorter insertions occurred for other subjects because the full insertion could not be achieved. At 1-month postactivation, hearing preservation was found in half of the subjects who had measureable preoperative hearing. At 1-year postactivation, hearing preservation was achieved in one-fourth of the subjects who had measurable preoperative hearing. Taken collectively, the results of the Prentiss et al. (2010) and Baumgartner et al. (2007) studies indicate that hearing preservation is possible with deeply inserted electrodes, particularly when soft surgical techniques are employed and the electrode array is designed to facilitate hearing preservation. Additional research is needed to further clarify the relationship between insertion depth and hearing preservation.

MED-EL is currently conducting a multicenter study with the objective of expanding indications for use of the FLEX$^{24}$/FLEX$^{EAS}$ electrode with adults who have precipitously sloping high-frequency hearing loss. At the time of this writing, the trial results have not been submitted to the FDA for consideration of commercial approval, but MED-EL does offer an EAS/hybrid cochlear implant system that is available in most other parts of the world. The MED-EL EAS system consists of the internal CONCERTO titanium internal implant with the atraumatic FLEX$^{24}$ electrode array and the external DUET 2 sound processor with a custom earmold (Figures 10–9 and 10–10). As previously noted, the FLEX$^{24}$ electrode array is available for commercial use in the United States, but the DUET 2 audio processor for EAS is in clinical trial. Internationally, the DUET 2 EAS processor is used for the EAS indication.

The DUET and DUET 2 sound processors possess an acoustic receiver, which is electrically coupled to the DUET sound processor and is acoustically coupled to an earhook that may be attached to a custom-made ear mold (see Figure 10–9). The DUET processor was MED-EL's original EAS processor, but it has been replaced by the DUET 2 EAS processor, which possesses the OPUS 2 housing design. The DUET 2 sound processor also features a built-in telecoil, and it may be used with the FineTuner remote control.



**FIGURE 10–9.** DUET 2 sound processor with a custom earmold. *Note.* Image courtesy of MED-EL Corporation.

The hearing aid portion of the DUET 2 provides approximately 43 dB of gain from 125 to 1700 Hz. The acoustic component of the DUET 2 sound processor is programmable, and allows the clinician to adjust the acoustic settings via four parameters: (1) gain at 500 Hz: 27 to 43 dB, (2) low-frequency slope (0–18 dB/octave), (3) "volume"/overall gain control (0–47 dB; intended to match overall loudness between HA and CI components), and (4) AGC threshold (40–70 dB SPL). The aforementioned acoustic parameters may also be adjusted in the DUET sound processor, but these adjustments are made by way of four trim pots rather than via the programming interface within MAESTRO.

The DUET 2 and DUET processors offer electrical stimulation from 70 to 8500 Hz, with adjustment for the low frequency boundary to

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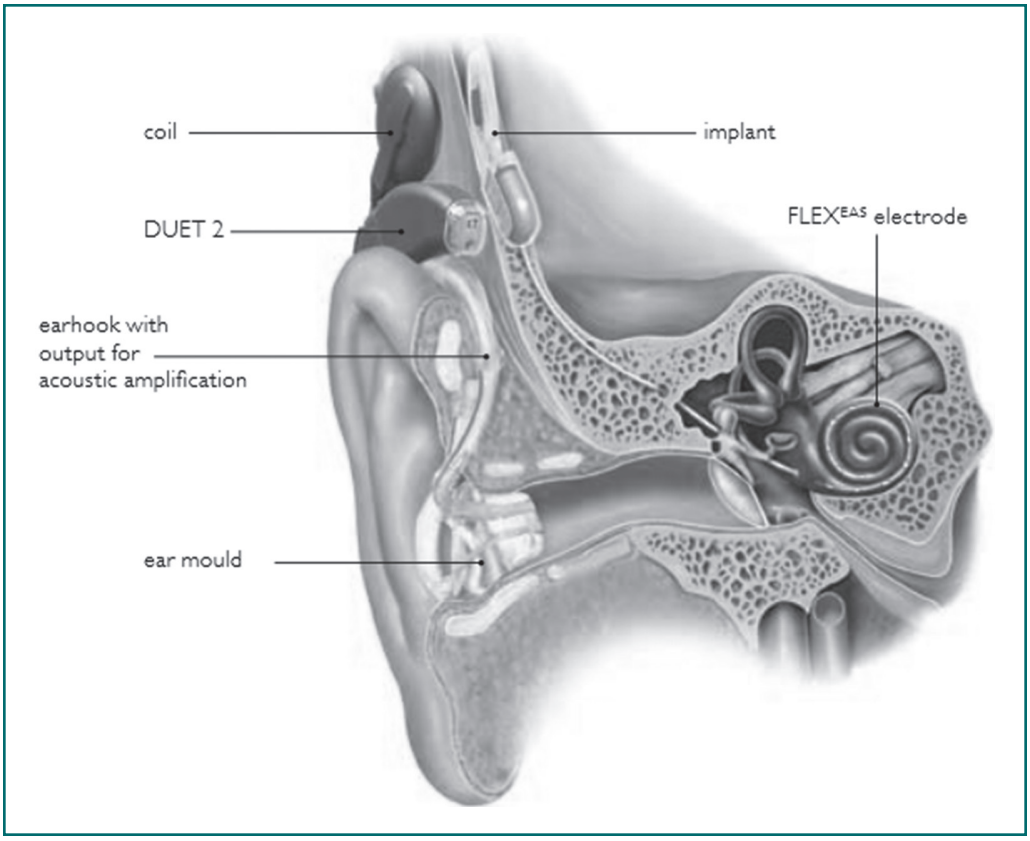



**FIGURE 10–10.** Diagram of MED-EL DUET 2 sound processor and FLEX$^{24}$ (formerly FLEX$^{EAS}$) internal device. *Note.* Image courtesy of MED-EL Corporation.

accommodate EAS users. The DUET 2 is used with the energy efficient D Coil, which allows for up to 7 days of battery life. The DUET 2 sound processor also contains the same front-end processing available on all MED-EL processors including Automatic Sound Management (ASM), a wide input dynamic range (IDR), and a dual-stage automatic gain control (AGC). Additionally, the processor has an audio-input jack to allow connection to other audio sources, such as iPods or MP3 players.

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## PERFORMANCE WITH HYBRID OR MINIMALLY INVASIVE COCHLEAR IMPLANTS

Studies on the efficacy of EAS with hybrid implants suggest significantly better postimplant outcomes as compared to performance with traditional acoustic hearing aids alone and often better performance with EAS as compared to the cochlear implant alone (Büchner et al., 2009; Fraysse et al., 2006; Gantz, Turner, & Gfeller, 2004; Gantz, Turner, Gfeller, & Lowder, 2005; Gantz et al., 2009; Kiefer et al., 2005; Lenarz et al., 2013; Luetje, Thedinger, Buckler, Dawson, & Lisbona, 2007). For example, Keifer et al. (2005) reported an average improvement of 8% for word and sentence recognition in a quiet condition with EAS as compared to the cochlear implant alone in 13 individuals after using MED-EL implants for 1 year. In these same participants, sentence recognition in background noise at a +10 signal-to-noise ratio improved by 23% in the EAS condition versus the cochlear implant alone condition. In another study, Büchner et al. (2009) examined sentence-recognition thresholds in the presence of a single competing talker in 22 patients using the Nucleus Hybrid L implant. The best thresholds (low-

est) were obtained with the combined cochlear implant and hearing aid (3.9 dB) as compared to the cochlear implant alone (10.8 dB) or hearing aid alone (15.9 dB).

In a clinical trial, Gantz et al. (2009) reported results from 85 patients implanted with the Cochlear Hybrid 10 device and revealed that 91% of these patients maintained some usable residual hearing. Improvements in word recognition in quiet and sentence-in-noise thresholds were obtained by half of the patients when using EAS over preoperative hearing aids alone. Interestingly, 91% of the variance in the performance of patients ($n$ = 16) who did not benefit from EAS over hearing aids alone was explained by the preoperative word-recognition performance and duration of deafness in the high-frequency region (i.e., poorer EAS benefit was associated with poorer preoperative word recognition performance and a longer duration of high-frequency deafness—greater than 30 years).

Lenarz et al. (2013) examined behavioral and subjective performance in 66 adults who received Nucleus Hybrid L24 implants. Eighty-eight percent of the participants maintained residual low-frequency hearing after surgery, and 88% continued to use the hybrid devices 1 year following surgery. On average, 65% of these participants realized an increase in speech recognition in quiet whereas 74% improved in noise. Subjective questionnaires yielded significant improvements in ratings of speech understanding, spatial aspects of hearing speech, and the quality of speech.

Hybrid implants are also reported to improve localization abilities and subjective performance. In one study, Dunn, Perreau, Gantz, and Tyler (2010) reported that, when combining the use of a hybrid implant in one ear (implant plus hearing aid) with an acoustic hearing aid on the nonimplant ear, 11 patients with Nucleus Hybrid 10 implants displayed significantly better speech recognition in noise and localization than they achieved with the Hybrid implant alone or a bimodal arrangement (traditional cochlear implant plus contralateral hearing aid). In a study focused on the potential subjective benefits of EAS, Gstoettner et al. (2011) asked 23 patients with MED-EL devices to rate their percentage of difficulty hearing in various listening situations. When compared to the global average preoperative rating of 74% with their hearing aids alone, the patients reported a significant decrease in global hearing difficulty with a rating of 45% when using EAS for 3 months.

Another substantial benefit of hybrid implantation is the preservation of low-frequency hearing in both ears to allow the recipient to capitalize on interaural timing cues to better understand speech in noise when spatial separation exists between the speech and noise signals. Gifford et al. (2013) evaluated speech recognition in noise for 38 subjects who used electroacoustic stimulation in one ear and a hearing aid in the opposite ear. Speech recognition was assessed using an eight-loudspeaker array (each loudspeaker separated by 45 degrees), with the signal of interest presented from the loudspeaker located directly in front of the subject and the noise presented from each of the eight loudspeakers. The noise was presented at 72 dBA, and speech recognition was assessed at fixed signal-to-noise ratios (SNR) of +2 and +6 dB. Also, speech recognition was evaluated adaptively to determine the SNR necessary for 50% correct performance. Furthermore, the testing was completed in a sound treated audiometric test booth with minimal reverberation and also with a simulated reverberation time of 0.6 seconds. For the aforementioned conditions, performance was assessed in several conditions: (1) bimodal—use of the hearing aid for the unimplanted ear along with electrical stimulation only for the implanted ear, (2) combined—use of the hearing aid for the unimplanted ear along with electroacoustic stimulation for the implanted ear, and (3) binaurally aided—acoustic only stimulation for each ear.

Gifford and colleagues (2013) found that electrical stimulation provided significantly better performance than the binaural hearing aid condition. Additionally, the best performance was obtained when the subjects used the combined mode of stimulation (i.e., electroacoustic stimulation for the implanted ear along with the hearing aid for the opposite ear). Compared to the bimodal condition, use of the combined mode of stimulation resulted in a 6 to 10 percentage point improvement in speech recognition at the fixed SNR and a 1.8 dB improvement in the SNR needed for 50% correct performance. They also reported that

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benefit from combined use was correlated with the degree of postoperative low-frequency hearing and the interaural time difference threshold. Gifford et al. (2013) concluded that the availability of binaural low-frequency hearing secondary to the preservation of low-frequency hearing in the implanted ear allowed the subjects to take advantage of interaural timing cues to capitalize on the squelch effect to understand speech better in the presence of spatially separated noise.

Other studies compared performance of patients using hybrid implants to patients with traditional cochlear implants with long-electrode arrays (Dorman et al., 2009; Golub, Wan, Drennan, Worman, & Rubinstein, 2012). For example, Dorman et al. (2009) compared word-recognition performance in 22 patients with the 10 mm Cochlear Nucleus Hybrid implants (EAS) to 25 patients using Cochlear or Advanced Bionics implants with conventional (long) electrode arrays. Although both groups showed significant improvements relative to preoperative performance with hearing aids, the patients using conventional electrode arrays had significantly better performance than the patients using Hybrid implants, which the author related to the broader band stimulation from the conventional versus the hybrid implants. It is important to note that the hybrid recipients in the Dorman et al. (2009) study were implanted with the 10 mm Nucleus Hybrid array, which is substantially shorter than any other electrode that is commercially available and used for EAS application (i.e., Nucleus (L24) Hybrid, Nucleus CI422, MED-EL FLEX[24]). As a result, the findings of the Dorman et al. (2009) study do not necessarily apply to the experience one would expect for the typical EAS recipient seen in clinical environments.

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## SELECTING ELECTRODE ARRAYS FOR HEARING PRESERVATION

At this point, the reader may be wondering which length/angular insertion depth is ideal for optimizing hearing preservation and performance. Although research has yet to suggest a fail-proof protocol for electrode selection for recipients with residual low-frequency hearing, there are some recent findings that may offer clinicians some guidance in this decision-making process. First, although hearing preservation has been demonstrated with long and periomodiolar arrays (Mick et al., 2014; Stanisławek-Sut, Morawski, & Niemczyk, 2013), the likelihood of low-frequency hearing preservation is most likely better with electrode arrays that are shorter, thinner in diameter, and designed for lateral wall placement (Mick et al., 2014). Indeed, temporal bone studies have demonstrated that electrode insertions that extend deeper than 360 degrees (about 20 mm for a lateral wall insertion) are more likely to cause damage to delicate cochlear structures (Adunka & Kiefer, 2006).

A recent publication by Skarzyński and colleagues (2014) examined low-frequency hearing preservation for a group of 35 recipients of the Cochlear Nucleus CI422 implant. The researchers divided the subjects into three groups on the basis of their preoperative air conduction threshold at 500 Hz: (1) better than or equal to 50 dB HL, (2) poorer than 50 dB HL but better than 80 dB HL, and (3) equal to or poorer than 80 dB HL. They reported several interesting findings. First, the vast majority of subjects achieved significant improvements in speech recognition in quiet and in noise with their cochlear implant relative to their preoperative binaural hearing aid performance. However, subjects with better preoperative hearing generally achieved better speech recognition in quiet and in noise with their implant than did the subjects with poorer preoperative hearing. In contrast, the subjects with better preoperative low-frequency hearing thresholds tended to suffer a greater amount of low-frequency hearing loss after surgery than subjects with poorer preoperative hearing (most likely because they had more functional low frequency to lose).

Skarzyński and colleagues (2014) provided a compelling explanation for the greater change in hearing loss suffered by the group with better preoperative hearing thresholds. In short, they contended that the group with better preoperative hearing thresholds was more likely to possess at least some active enhancement of hearing sensitivity secondary of the cochlear amplifier produced by remaining outer hair cells in the apical region of the cochlea. Furthermore, they theo-

rized that the presence of the electrode array may impair the active process of the cochlear amplifier leading to a decrease in outer hair gain and subsequently to hearing sensitivity. More specifically, Skarzyński et al. (2014) hypothesized that greater impairment of outer hair cell function would occur as the electrode array became closer to the location of the functioning outer hair cells, facilitating the response at a given characteristic frequency. The researchers evaluated their theory by examining the relationship between hearing threshold shift at different frequencies across the low-frequency octaves and the location of the electrode as indicated by postoperative spiral CT scans. The estimated characteristic frequency at a specific location was estimated based on the angular insertion depth of the electrode array relative to the Greenwood function. Indeed, Skarzyński and colleagues (2014) found a much higher potential for significant decreases in hearing thresholds at estimated characteristic frequencies that were within 135 degrees of the tip of the electrode array. They attributed this degradation in threshold to a change in cochlear mechanics and function associated with the presence of the array. Based on their findings, Skarzyński et al. (2014) provided the following clinical suggestions. If an electrode is inserted 360 degrees (one full turn along the lateral wall) into the cochlea (e.g., the CI422 inserted to 20 mm), then one would expect a higher chance of impairing outer hair cell function at 1000 Hz and causing a decrease in hearing sensitivity at that frequency. Based on this line of thinking, a shorter array, like the Nucleus L24 Hybrid device, is likely to be more appropriate for recipients with useful residual hearing sensitivity (e.g., 65 dB HL or better) at 1000 Hz. However, if the goal is to preserve useful residual hearing at 500 Hz and below (because hearing sensitivity at higher frequencies is too poor to expect benefit from amplification), then an electrode array that is inserted to 360 degrees is appropriate.

A few additional points are worth discussing. First, there are some reports that suggest that implants with arrays inserted to an angular insertion depth of 360 degrees provide similar speech recognition in the electric-only mode when compared to performance obtained by users with electrode arrays of a conventional length (Balkany et al., 2007; Cochlear Americas, 2013). For instance, 50 subjects in the U.S. FDA clinical trial of the Nucleus L24 Hybrid cochlear implant achieved similar mean monosyllabic word recognition as subjects who received the Freedom implant with the Contour Advance electrode array with a typical angular insertion depth of 430 degrees (Balkany et al., 2007; Cochlear Americas, 2013). Of course, one might contend that the recipients in the L24 clinical trial possessed preoperative otologic histories that were more favorable than the recipients who participated in the Freedom Contour Advance trial, and as a result, their speech recognition should have been better. Although that may be true, Adunka et al. (2010) also found similar speech recognition for a group of users of the MED-EL FLEX$^{24}$ (formerly FLEX$^{EAS}$) users with an insertion depth of approximately 20 mm (360 degrees) when compared to a group of MED-EL recipients with standard arrays (e.g., 31 mm insertion depth). The two groups were matched on several variables including preoperative pure tone air conduction thresholds and aided word recognition. Both groups achieved about 50% monosyllabic word recognition in the ipsilateral electric-only condition 6 months after implant activation, but the hybrid group achieved a mean of score of 69% correct on monosyllabic word recognition in the EAS condition.

Likewise, Helbig, Helbig, Rader, Mack, and Baumann (2009) compared speech recognition for subjects who underwent reimplantation with a longer electrode array after experiencing a significant decrease in residual hearing following hybrid cochlear implant surgery. Based on their findings, these researchers concluded that a minimum electrode array length of 18 mm is needed to provide adequate performance via electric-only stimulation.

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In contrast, research has suggested that conventional-length electrode arrays are likely most appropriate for persons with severe to profound hearing loss across the electrode array (Dorman, Loizou, & Rainey, 1997; Stakhovskaya et al., 2007). Additionally, there are several studies that demonstrate the benefits of perimodiolar electrode array placement, which include a need for lower stimulation levels to elicit a given psychophysical percept, better battery life, and reduced channel interaction (Balkany, Eshraghi, & Yang, 2002;

O'Leary, Richardson, & McDermott, 2009; Wackym et al., 2004). As a result, conventional length electrode arrays and perimodiolar electrode arrays are still likely to be beneficial for candidates with severe to profound hearing loss across the audiometric frequency range. In short, more research is needed to clarify the benefits and limitations of short versus long electrodes for persons with precipitously sloping high-frequency hearing loss.

It should be noted that at the time of this writing, there is limited evidence to indicate the maximum amount of low-frequency hearing loss for which a recipient may have and still receive benefit from low-frequency amplification. The Cochlear L24 Hybrid FDA research has suggested that recipients with hearing thresholds up to 85 dB HL generally receive better speech recognition with use of the acoustic amplification. Recipients with poorer low-frequency hearing typically do not experience improvement in speech recognition, but some do prefer acoustic amplification for subjective sound quality enhancement (Cochlear Americas, 2013).

## OPTIMIZING THE FITTING

There are several programming considerations to take into account for hybrid implants. Some of the more basic considerations include: (1) prescriptive fitting strategy to determine the gain and maximum power output of the hearing aid portion of the device, (2) cutoff frequency for acoustic amplification, and (3) the frequency range over which electrical stimulation will be allocated. Additional considerations include: (1) taking into account the potential differences in compression parameters associated with the acoustic and electric signals, (2) managing contrasting delays that may exist in the delivery of the acoustic and electric signals, and (3) attempting to use pitch and/or loudness scaling to optimize the allocation of the acoustic and electric signals. The pitch or loudness scaling may be necessary to balance the loudness of the acoustic and electric signals provided from the hybrid sound processor and to balance the loudness of the signals provided by the hybrid sound processor and the contralateral hearing aid.

In the following paragraphs, we discuss the results of research studies that have examined the aforementioned considerations. At this point in time, it is important to note that there is a paucity of research examining the effects of advanced parameters, such as delay times between the electric and acoustic signals along with signal processing characteristics (i.e., compression time constants), on recipient performance. As a result, those parameters are not discussed in detail in this chapter. Following the review of existing research pertaining to the programming of EAS devices, a clinical protocol for the programming of hybrid cochlear implants (e.g., EAS) will be provided.

### Frequency Allocation of Acoustic and Electric Stimulation

As indicated in earlier chapters in this book, a conventional cochlear implant provides electric stimulation to convey the entire speech frequency range. In contrast, a hybrid cochlear implant system provides acoustical stimulation for low-frequency inputs and electric stimulation for high-frequency inputs. With a hybrid implant, the clinician is faced with the task of determining how acoustic and electric stimulation should be allocated for sound inputs across the speech frequency range. A number of studies have sought to determine how acoustic inputs should be allocated in the acoustic and frequency stimulation modes, and the following section briefly describes these studies.

Keifer et al. (2005) examined performance of 13 recipients who had air conduction thresholds better than 60 dB HL below 1000 Hz and who received the MED-EL COMBI 40+. During the course of the study, the electrode array of the COMBI 40+ that was conventionally provided at the time of the study was modified so that the distance from the tip to the most basal intracochlear stimulating electrode contact was 21.9 mm (as opposed to a typical distance of 27.4 mm). This resulted in a reduction of the dis-

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tance between intracochlear electrode contacts from 2.4 to 1.9 mm, which allowed for a larger number of electrode contacts to be inserted into the cochlea. Regardless of which electrode array was inserted, the typical insertion electrode array depth achieved in the study was about 20 mm. Two of the recipients lost their low-frequency residual hearing, so EAS results are available for 11 of the 13 participants. These subjects were fitted with digital in-the-ear hearing aids in order to receive EAS. The hearing aids were set with the gain of low-frequency amplification to half of the air conduction threshold (e.g., 30 dB of gain at 500 Hz if the air conduction threshold at that frequency was 60 dB HL—"½-gain rule"). Real ear measures were not conducted to measure the in situ output of the ipsilateral hearing aid.

During word- and sentence-recognition testing in quiet, participants used three frequency allocations for electrical stimulation: full frequency range (300–5500 Hz), 650 Hz low-frequency boundary (650–5500 Hz), and 1000 Hz low-frequency boundary (1000–5500 Hz). In all but one participant, the best performance and subjective benefit was obtained with the full frequency range of electrical stimulation in combination with the hearing aid. The remaining patient performed best with the 650 Hz low-frequency boundary. It should be noted that only 4 of the 11 EAS subjects possessed an air conduction threshold better than 90 dB HL at 500 Hz; therefore, the group's benefit from acoustic stimulation above 500 Hz was likely limited. Because of that and the fact that verification measures were not conducted to ensure in situ audibility for low-frequency speech inputs, the results of this study may not apply to the typical EAS recipient.

In another study, Fraysse et al. (2006) evaluated speech recognition in 27 recipients of the Cochlear Nucleus 24 implant with the Contour Advance electrode array. Soft surgical techniques were employed and low-frequency hearing, which was defined as air conduction thresholds of 80 dB HL or better at 125 and 250 Hz and 90 dB HL or better at 500 Hz, was preserved in nine recipients. These nine recipients were fitted in the implanted ear with custom hearing aids with the gain set to the Desired Sensation Level (Input/Output) prescriptive method. Two implant programs, which differed in the way in which electrical stimulation was frequency-allocated, were created for the subjects using EAS. One program provided electrical stimulation across the entire speech frequency range, which presumably resulted in some overlap between acoustic and electric stimulation provided for low-frequency inputs. In the second program, the only acoustic stimulation was provided for low-frequency inputs, and electrical stimulation was only provided for the high-frequency inputs. The exact cutoff between low- and high-frequency inputs was not defined in this study nor was the method used to determine the cutoff between acoustic and electric stimulation for the second program. The researchers reported that 7 of 9 subjects preferred the program without overlap between the acoustic and electric stimulation, whereas two subjects preferred the program with overlapping stimulation. There was no difference in word recognition in quiet between the two programs, but there was a trend toward better sentence recognition in noise with use of the nonoverlapping program. However, there were insufficient data to allow for a formal statistical analysis. It is possible that the trend toward poorer speech recognition in noise with the overlapping program may have been attributed to masking of the low-frequency acoustic signal by the low-frequency electrical stimulation.

Simpson, McDermott, Dowell, Sucher, and Briggs (2009) compared behavioral performance of five users of the Nucleus Freedom Contour Advance electrode array when using (1) only high-frequency electrical stimulation paired with the hearing aid providing low-frequency information below a specified cutoff versus (2) a map with the full frequency range plus the hearing aid proving the same low-frequency input. With both maps, participants showed significantly better speech-recognition performance in quiet relative to preoperative performance with hearing aids alone, but no significant differences in performance were detected between the two maps for speech recognition in quiet and in noise.

Most recently, Karsten and colleagues (2013) at the University of Iowa examined the effect of varying frequency allocations in 10 users of the

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Cochlear Nucleus Hybrid (S8 or S12) device with a 10 mm electrode array. All subjects used the Nucleus Freedom hybrid sound processor. The Audioscan Verifit real ear measurement system was used to evaluate the output of the acoustic component, which was programmed to provide an output that matched (± 7 dB) the National Acoustics Laboratories Non-Linear 1 (NAL-NL1) prescriptive target for a 65 dB SPL calibrated speech signal. The upper edge of acoustic amplification was defined as the highest frequency that was within 7 dB of the NAL-NL1 target. Acoustic channels above this frequency were disabled. Three programs with differing cutoff frequencies for electric stimulation were created: (1) the "meet" program aimed to set the cutoff frequency of electrical stimulation to match the upper edge of acoustic amplification, (2) the "overlap" program, which set the lower end of the electrical range so that it was less than 50% of the upper edge of the acoustic amplification, and (3) the "gap" program, which set the lower end of the electrical range to be 50% higher than the upper edge of acoustic amplification.

The three different programs were compared in speech recognition in quiet and in noise as well as for the subjects' subjective preference for each program. Although there was a trend toward better speech recognition in quiet with the "meet" program, there was no statistically significant difference in speech recognition in quiet across the three programs. However, recognition of spondees in noise was significantly better in the "meet" condition when compared to the "overlap" condition. Additionally, subjective results yielded better ratings for the "meet" condition over the remaining two frequency allocation conditions. Overall, the authors theorized that, because acoustic hearing provides better frequency resolution than electric hearing, the "overlap" program may have impaired speech recognition in noise by allowing for electric masking of the acoustic signal. The authors also concluded that the "meet" program generally provided better overall performance than the "overlap" program. Finally, it should be noted that this study included subjects with a 10 mm hybrid electrode array. A 10 mm electrode array typically extends to the 4000 Hz place in the cochlea, and as a result, the likelihood of physical interaction between the electric signal and the low-frequency acoustic signal is reduced. In contrast, commercially available EAS implants (e.g., Cochlear Nucleus L24 Hybrid, MED-EL FLEX[24]) are longer; thus, a higher likelihood of physical interaction may exist.

Vermiere and colleagues (2008) conducted a smaller study (e.g., four subjects) to examine how EAS stimulation should be frequency-allocated in both the acoustic and electric domains. This study included recipients of the MED-EL COMBI 40+ Medium electrode array, which was inserted to a depth of 18 mm. The subjects' postoperative air conduction hearing threshold at 500 Hz ranged from 65 to 100 dB HL at the time of the ipsilateral hearing aid. Acoustic stimulation was allocated in two different ways: (1) no amplification was provided at frequencies in which the air conduction threshold exceeded 85 dB HL, and (2) acoustic amplification was allocated for all frequencies with air conduction thresholds better than 120 dB HL. The latter strategy was more likely to allow for amplification of frequency regions often associated with cochlear dead regions (Moore, Vickers, Glasberg, & Baer, 1997). Some studies have suggested that it may be beneficial to amplify one octave into a dead region for subjects with precipitously sloping hearing loss (Baer, Moore, & Kluk, 2002). The output of an Oticon Adapto P in-the-ear (ITE) hearing aid was measured with in situ real ear probe microphone measures to ensure it was appropriate at the frequencies over which acoustic amplification was provided. Electric stimulation was also allocated in two different manners: (1) the "full" range of electrical stimulation was provided as with the conventional MED-EL TEMPO+ processor, and (2) electrical stimulation began at the first air-conduction frequency exceeding 65 dB HL. Sentence recognition in noise and subjective ease of listening were evaluated across all conditions. The subjects generally performed better when electric allocation was limited to the mid- to high-frequencies (i.e., reduced electrical overlap). However, the subjects typically achieved better speech recognition and expressed a greater preference for the program with the broader acoustic allocation (Vermiere et al., 2008). It should be noted that the findings of this study are limited by the small number of study participants and

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also by the fact that postoperative, low-frequency residual hearing was relatively poor.

Turner, Reiss, and Gantz (2008) compared speech recognition of seven Nucleus S8 (10 mm insertion) hybrid recipients with two different programs: (1) electrical stimulation allocated from 750 to 8000 Hz and acoustic allocation from 750 Hz and below, and (2) electrical allocation from 2000 to 5000 Hz and acoustic allocation from 750 Hz and below. It should be noted that allocating electrical stimulation from 2000 to 5000 Hz closely corresponds to the typical frequency place of stimulation in the cochlea where the 10 mm-long electrode array resides. The researchers measured speech recognition in quiet with the two different programs and found mixed results. Three of the seven subjects performed better with the program with the wider range of electrical allocation, whereas one subject performed better with the program with the reduced range of electrical allocation. There was no difference in the mean performance between the two programs. The authors noted that the fact that most of the subjects performed relatively well with the wide range of frequencies allocated electrically over a 10 mm span in the basal region of the cochlea is evidence of the plasticity of the auditory system in its capability to acclimate to a signal that is significantly shifted/distorted in the frequency domain. Again, the results of this study may not be completely applicable to the typical contemporary clinical experience, as current commercially available EAS devices across the world customarily incorporate longer electrode arrays with deeper insertion depths.

In a similar study with three Nucleus S8 (10 mm electrode array with six intracochlear electrode contacts) Hybrid recipients, Reiss, Perreau, and Turner (2012) compared speech recognition and pitch perception for two different programs: (1) electrical stimulation allocated from 688 to 7938 Hz, and (2) electrical stimulation allocated from 188 to 7938 Hz. All three subjects lost their low-frequency hearing within the first year following implant surgery, so they discontinued ipsilateral hearing aid use. On the opposite ear, they did use a hearing aid, which was fitted to the NAL-NL1 prescriptive target for a 65 dB SPL speech signal. Word recognition in quiet and sentence recognition in noise was measured with the two programs. Pitch matching relative to the contralateral ear was also compared across the two programs. The authors noted that the subjects generally realized no improvement in speech recognition in quiet or in noise with the wider allocation of electrical stimulation. However, the subjects did demonstrate changes in pitch perception associated with the changes in frequency allocation. This finding, again, demonstrates the ability of the auditory system to acclimate to a frequency-shifted signal.

Several studies have set the high-frequency cutoff of the hearing aid based on the first frequency at which the air-conduction pure tone threshold reaches the profound range. In general, this approach places the high-frequency cutoff of amplification at the edge of usable residual hearing in the low frequencies. Specifically, several research groups recommend a cutoff frequency where pure tone hearing thresholds exceeded 90 dB HL (Gantz et al., 2009; Golub et al., 2012; Lenarz et al., 2013). To examine the potential effects of hearing aid cutoff frequency Büchner et al. (2009) compared sentence recognition in noise across three cutoff frequencies (300, 500, 700 Hz cutoffs) in eight participants using the Nucleus Hybrid L24 device with a full frequency range for electrical stimulation (i.e., acoustic-electric overlap). The speech recognition results suggested that increasing the cutoff beyond 300 Hz did not significantly improve performance. It should be noted that the subjects in the Büchner study generally had a substantial amount of low-frequency hearing loss, so it is possible that the limited benefit of low-frequency amplification was due to the subjects' poor auditory function across the frequency range under study.

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Overall, there is a paucity of published studies that have systematically examined the cutoff frequency to determine the optimal frequency range over which acoustic amplification should be provided to maximize the performance of hybrid recipients. Furthermore, research has yet to reveal the minimal low-frequency range over which functional hearing must exist in order for the provision of acoustic stimulation to be beneficial relative the electric-only stimulation. In other words, there is still no clear understanding

of whether EAS will provide sufficient benefit for a recipient who only has functional hearing at 250 Hz or below (and profound hearing loss from 500 Hz and beyond), or if substantial EAS benefit is only present for those who have functional hearing out to 500 Hz, 750 Hz, and so forth. Most likely, the extent of EAS benefit will increase as the range over which functional hearing exists also increases. However, additional research is needed to specifically delineate the expected amount of EAS benefit relative to electric-only hearing as a function of the frequency range of usable hearing.

In March, 2014, the U.S. FDA approved the Nucleus L24 Hybrid cochlear implant for commercial use. This approval followed a Cochlear-sponsored clinical trial of the L24 device that was designed to evaluate recipient outcomes to justify the manufacturer's proposed labeling for clinical use in the United States. Fifty adult recipients participated in this trial at 10 different cochlear implant centers in the United States. Subjects met the following audiometric inclusion criteria for participation in the study:

1. Severe to profound (an air conduction pure tone threshold average at 2000, 3000, and 4000 Hz equal to or poorer than 75 dB HL) sensorineural hearing loss for frequencies greater than 1500 Hz along with low-frequency at 500 Hz and below of 60 dB HL or better.
2. Consonant-Nucleus-Consonant (CNC) monosyllabic word recognition scores equal to or better than 10 but no better than 60% correct in the ear to be implanted.
3. CNC word scores in the contralateral ear equal to or better than the implanted ear but no better than 80% correct.
4. No conductive hearing loss.
5. The duration of severe to profound hearing loss cannot exceed 30 years.

These subjects were fitted with the Nucleus Freedom Hybrid sound processor. Postoperative air-conduction and bone-conduction thresholds were measured bilaterally at every octave and interoctave frequency from 125 to 8000 Hz. Acoustic amplification was provided at all frequencies possessing an air-conduction threshold better than 90 dB HL. The output of the acoustic component was fitted to the NAL prescriptive target across the low-frequency range. The low-frequency boundary of electrical stimulation was set at a frequency that was as close as possible to the last aidable acoustic frequency. For instance, if aidable residual hearing existed at 750 Hz but not 1000 Hz, then electrical stimulation began at approximately 800 Hz.

The participants in the study performed significantly better with their L24 hybrid cochlear implant system than they did preoperatively with binaural hearing aids. Specifically, they achieved an average improvement of 37% on CNC word testing at 60 dBA and 36% on AzBio sentences in noise (+5 dB SNR). Additionally, the subjects performed better in the EAS mode than they did in both the acoustic and electric-only conditions. This study included a relatively large number of subjects with an EAS device that is now commercially available. The favorable results obtained by the study subjects lend credibility to the procedures used to program the acoustic and electrical stimulation levels of the Nucleus Freedom hybrid sound processor.

Other researchers have explored the use of pitch scaling and matching as a tool to determine acoustic/electric allocation for EAS users (Baumann, Rader, Helbig, & Bahmer, 2011; Simpson et al., 2009). Simpson and colleagues (2009) determined the acoustic frequency corresponding to electric stimulation (pulse train at 900 pps) on the most apical electrode (electrode 22) for a group of five Nucleus Freedom Contour Advance recipients with residual hearing in the ipsilateral or contralateral ear. Soft surgical techniques were employed, and the Contour Advance electrode array was inserted approximately 17 mm into the cochlea. The pitch matching was completed in the ipsilateral ear for two subjects who had residual hearing in the implanted ear and in the opposite ear for three subjects who had no postoperative residual hearing. Speech recognition in quiet and in noise was compared with two programs: (1) a conventional program with electrical stimulation allocated across the entire speech frequency range (188 to 7938 Hz) and (2) the lowest frequency of electrical stimulation was based on the acoustic frequency that yielded a pitch match to electrical stimulation on electrode 22. Subjects were tested

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in the electric-only condition and also while they used a hearing aid. Subjects with residual hearing in the implanted ear used a hearing aid in the ipsilateral and contralateral ear. The output of the hearing aid was matched to NAL-NL1 prescriptive targets. For the two subjects with ipsilateral residual hearing, the cutoff frequency of amplification was set to the highest frequency possessing an air conduction pure tone threshold better than 90 dB HL. The researchers reported that the subjects performed better with EAS stimulation than they did with the hearing aids or cochlear implant alone. This finding supports the use of hearing aids programmed to an evidence-based prescriptive target to improve speech recognition relative to the use of electric stimulation alone. However, Simpson and colleagues (2009) found no difference in speech recognition between the two cochlear implant programs. The small number of subjects in this study, particularly those with residual hearing in the implanted ear, limits its application to the clinical management of EAS users.

Baumann and colleagues (2009) conducted pitch matching with eight EAS users who had residual low-frequency hearing in their ipsilateral and contralateral ears. These subjects used MED-EL PULSAR CI$^{100}$ or SONATA TI$^{100}$ implants. Four of these recipients were implanted with the MED-EL FLEX$^{24}$ electrode array, whereas the other four were implanted with the MED-EL Standard array (31.5 mm in length). The subjects listened to an 800 pps pulse train presented to a single electrode and adjusted the frequency of an acoustic pure tone to match the pitch of the electrical signal. This assessment was conducted on two to four of the most apical electrodes and was completed for the ipsilateral and contralateral ear.

Baumman et al. (2009) discovered that the difference in pitch matching between the ipsilateral and contralateral ears was quite small (the mean average was 14 Hz). They also reported that the FLEX$^{24}$ recipients matched the electric signal on electrode 1 (the most apical electrode) to an acoustic pure tone frequency of 583 Hz, whereas the recipients with standard electrode arrays matched their most apical electrode to a mean frequency below 250 Hz. As expected, electrodes positioned at a deeper angular insertion depth were associated with a lower pitch percept. Of significant note, these researches noted a large amount of intersubject variability for pitch matching as a function of electrode location. Also, this study did show the potential of using pitch matching to determine the crossover frequency between acoustic and electric hearing. Considering the relatively small number of subjects in this study along with the significant amount of intersubject variability, additional research is needed in order to determine how pitch matching may be implemented in the clinical programming of EAS devices.

### Prescriptive Strategy for Acoustic Hearing Aid

Until future research determines a more appropriate fitting model, evidence-based prescriptive fitting strategies along with real-ear measures are recommended for determining the appropriate hearing aid gain below the low-frequency cutoff for acoustic amplification. Many researchers who have completed studies with hybrid implants reported use of the NAL NL-1 (Byrne et al., 2001) or the Desired Sensation Level (DSL) v5 (Scollie et al., 2005) prescriptive fitting strategies as a starting point for gain in the lower-frequency regions (Dorman et al., 2009; Gantz et al., 2009; Golub et al., 2012; Karsten et al., 2013; Lenarz et al., 2013), and use of these prescriptive strategies resulted in good to excellent speech-recognition performance for many of the patients in these studies. In one study, investigators used the half-gain rule along with nonlinear compression model to determine appropriate gain for frequencies ranging from 125 to 1000 Hz (Kiefer et al., 2005).

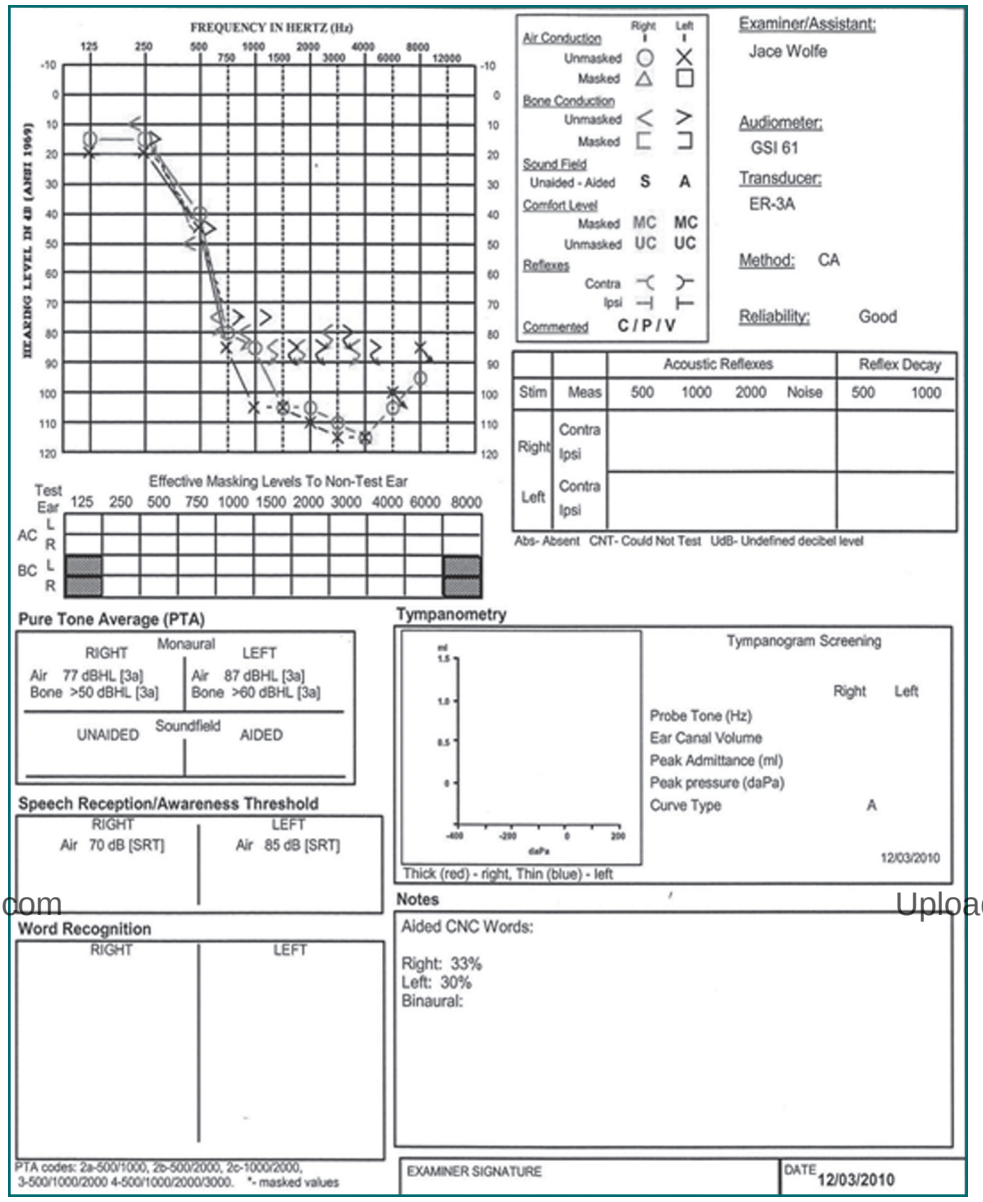
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## CLINICAL PROTOCOL FOR PROGRAMMING EAS DEVICES

The following protocol is based on the aforementioned research studies as well as the authors' clinical experiences with programming EAS devices. The first step involved with the programming of EAS devices is to obtain a pure tone air-conduction

and bone-conduction audiogram for the implanted ear. It is important to measure 125 Hz as well as every octave and interoctave frequency thereafter. Of course, the clinician should complete an otoscopic evaluation prior to the audiometric assessment to ensure there are no contraindications to the placement of the earphone or hybrid acoustic component and also to evaluate middle ear status. It is worth noting that some recipients will have a persistent middle ear effusion for several weeks following cochlear implant surgery, and as a result, it is prudent to conduct tympanometry to evaluate the admittance at the tympanic membrane. The authors generally wait about 4 weeks following surgery to activate the sound processor of EAS recipients in order to allow for the resolution of a potential middle ear effusion.

Figures 10–11 and 10–12 provide examples of a recipient's preoperative and postoperative audiometric profile. This recipient received a




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**FIGURE 10–11.** Examples of a recipient's preoperative audiometric profile.

Nucleus L24 hybrid cochlear implant for her left ear and a Nucleus CI422 cochlear implant for her right ear. As shown in these figures, a significant amount of low-frequency residual hearing remained after receiving each of her cochlear implants. The remainder of this example will illustrate how the Nucleus 6 sound processor was programmed for her left ear with the Nucleus L24 Hybrid cochlear implant.

After completion of the audiometric assessment, the clinician must determine how acoustic and electric stimulation will be allocated for the recipient. Recent research suggests that it may be worthwhile to amplify low-frequency inputs at frequencies with air-conduction thresholds of 85 dB HL or better (Cochelar Americas, 2013). The authors recommend conducting in situ real ear probe microphone measures to evaluate the output of the acoustic component of the EAS device. The recipient's real-ear-to-coupler difference should be evaluated prior to in situ real ear measurement so that the impact of the acoustics

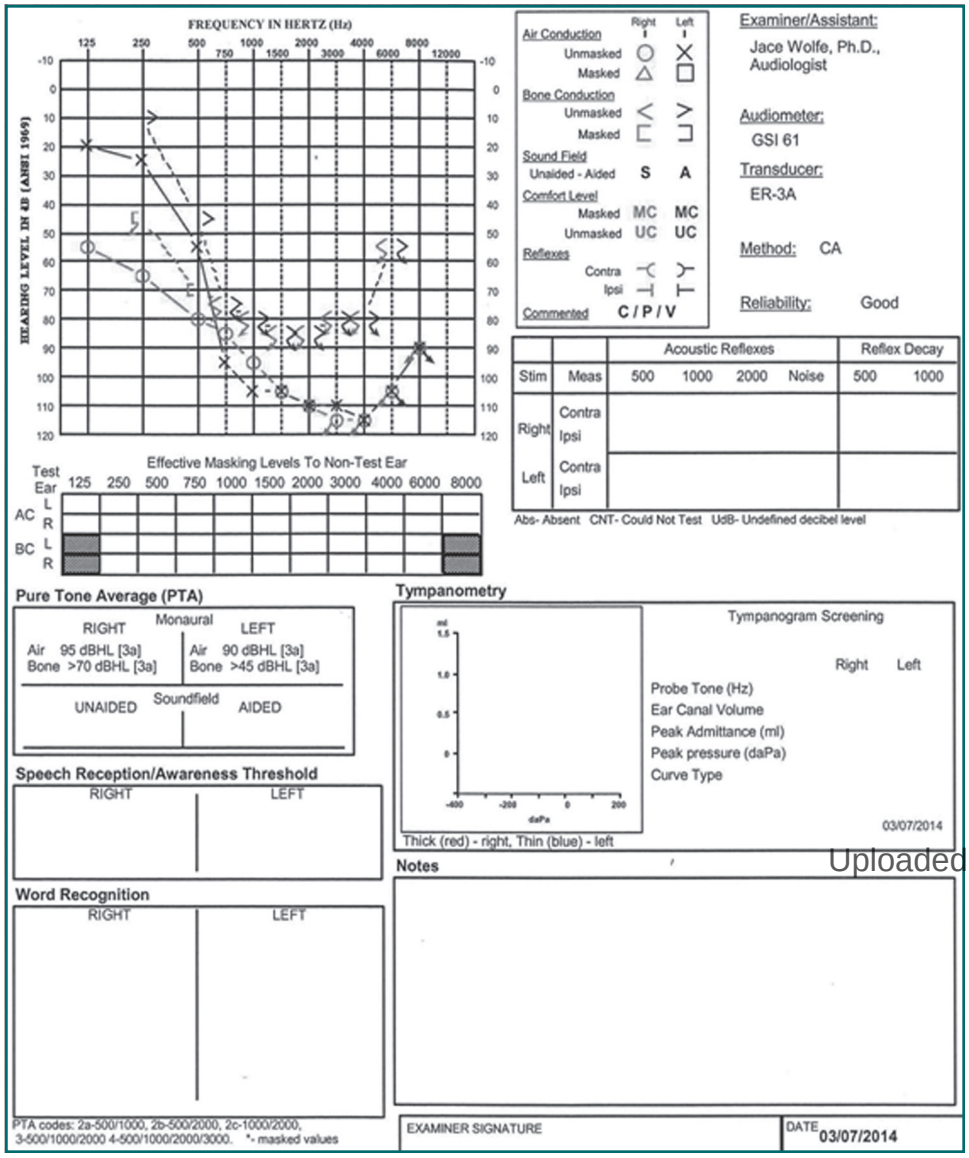
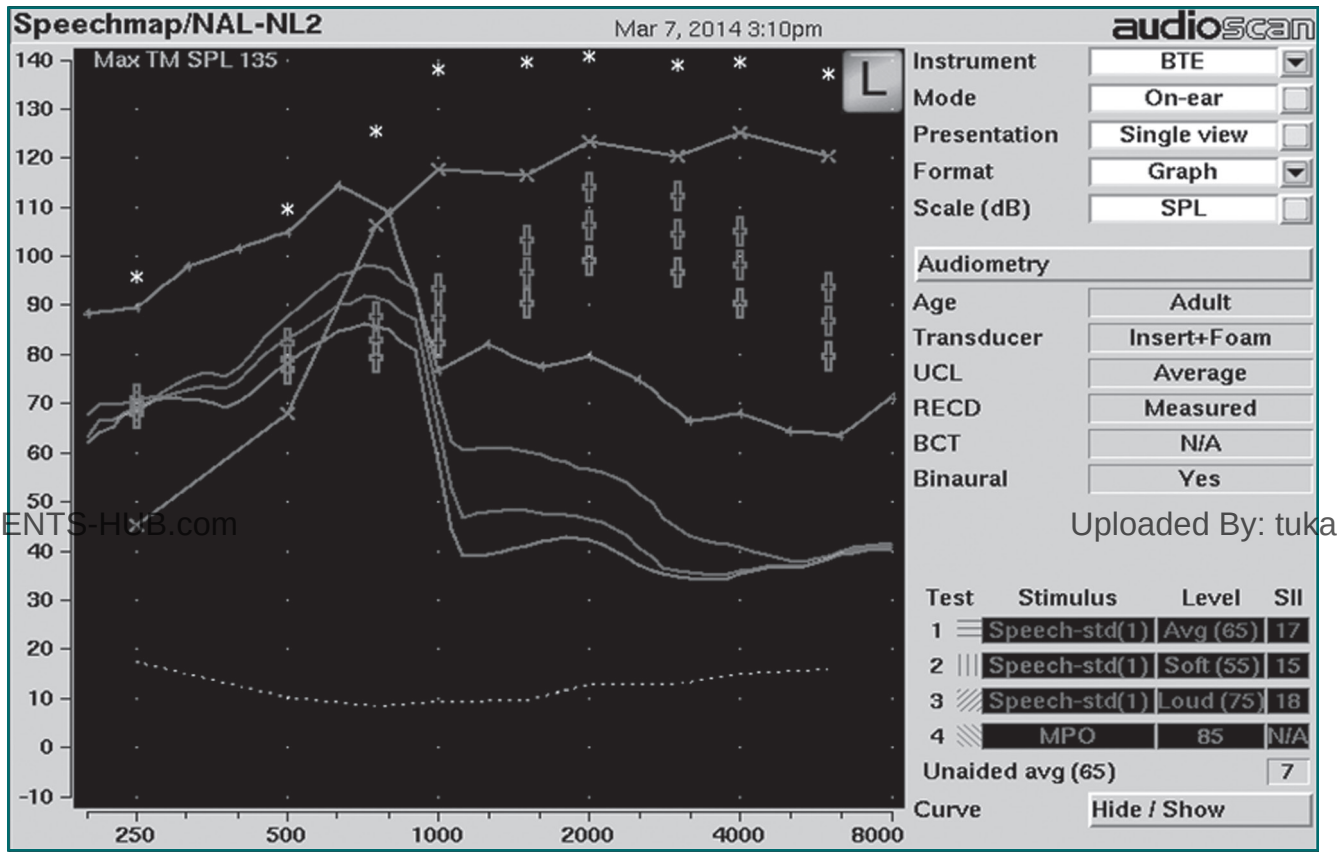



**FIGURE 10–12.** Examples of a recipient's postoperative audiometric profile.

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of the recipient's external and middle ear can be accounted for in the unaided thresholds used for device programming and real ear measurement. Next, the authors recommend programming the acoustic component so that the output is matched (± 5 dB) to an evidenced-based prescriptive target (e.g., DSL v 5.0 Adult, DSL v 5.0 Child, NAL-NL2). Also, it is likely most important that the slope of the acoustic response approximates the slope that is prescribed by the evidence-based generic prescriptive target. Ideally, the output should be measured for calibrated speech signals presented at 55, 65, and 75 dB SPL. Then, a high-level (85–90 dB SPL) swept pure tone should be presented to ensure that the MPO is appropriate (i.e., does not exceed the recommended maximum output of the prescriptive method). Figure 10–13 provides an example of real ear probe microphone results for an EAS recipient. The authors do not typically provide acoustic amplification for frequencies with pure tone thresholds of 90 dB HL or higher. However, some researchers (Baer, Moore, & Kluk, 2002) have suggested that it may be beneficial to provide amplification up to one octave above the start frequency of a dead region. With that in mind, some subjects may benefit from a wider acoustic bandwidth than what is suggested in this text. Figure 10–13 illustrates that audibility for average conversational level speech is limited when provided via acoustic stimulation because of the severity of the hearing loss beyond 500 Hz.

After the fitting of the acoustic component, the authors recommend determining the low-frequency boundary of the electric signal by noting the last acoustic frequency that meets each of the following criteria: (1) possesses an air-conduction threshold of 85 dB HL or better, and (2) the output of the acoustic component meets the prescriptive target for the 65 dB SPL speech signal. Electrical allocation should be provided at the lowest fre-




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**FIGURE 10–13.** Real ear probe microphone results for an EAS recipient.

quency at which audibility cannot be provided by the acoustic amplification from the hearing aid. Figure 10–14 provides an example of electrical allocation based on the results of the real ear probe microphone testing shown in Figure 10–13.

Next, T levels will need to be measured for all functional intracochlear electrode contacts in the same manner as T-level measurements are made with a conventional cochlear implant recipient. Then, C levels are also determined in the electric-only mode in the same manner that C levels are optimized with a conventional cochlear implant recipient. Specifically, C levels can be adjusted globally to the recipient's most comfortable listening level with the volume control set at a level that allows the recipient ample headroom so that she or he may increase or decrease stimulation as needed in real-world use. The clinician should then balance upper-stimulation levels in loudness across the electrode array. Following loudness balancing, the clinician can once again adjust the upper-stimulation levels globally to the recipient's most comfortable listening level (MCL) at a volume control setting that provides for both an increase or decrease in stimulation during real-world use. Further, the clinician should alternate between the ipsilateral acoustic only and electric only conditions to ensure that the loudness is balanced between the two modes of stimulation. Finally, the clinician will need to ensure that the loudness of the signal provided via the EAS device is similar to the loudness provided by the hearing technology used on the opposite ear.

## THE FUTURE OF EAS

Additional research will need to be conducted to continue to examine the behavioral and subjective benefit of EAS with hybrid implants as well as the optimal programming parameters for the electrical stimulation from the implant and acoustic gain from the hearing aid. It is likely that the clinical protocol suggested in this chapter will evolve and change as more knowledge pertaining to the management of EAS devices is acquired.

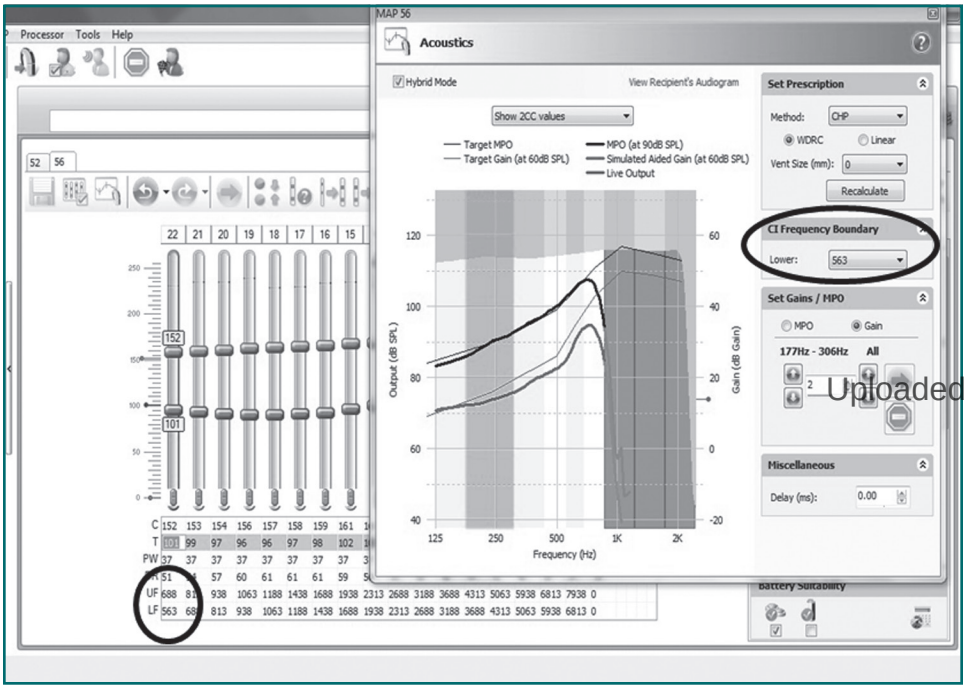



**FIGURE 10–14.** Electrical allocation based on the results of the real ear probe microphone testing.

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## KEY CONCEPTS

The reader should understand the potential benefits of EAS, programming considerations for hybrid implants, and the following key concepts:

- Hybrid implants are currently commercially distributed in the United States, and each manufacturer offers an implant array that is designed to enhance the likelihood of low-frequency hearing preservation. Although Cochlear currently distributes an EAS processor (e.g., Nucleus 6), it is likely that other manufacturers will offer an EAS processor in the USA in the future. MED-EL does commercially distribute an EAS processor in most other parts of the word and is currently conducting a clinical trial with the objective of seeking approval to commercially distribute an EAS processor (e.g., DUET 2) in the USA.
- On average, EAS provides superior behavioral and subjective performance over the preoperative performance with hearing aids, the electrical stimulation alone, or the hearing aid alone.
- When programming hybrid implants, the audiologist will need to consider the frequency range for the electrical stimulation, the prescriptive fitting strategy for the hearing aid, and the cutoff frequency for the hearing aid.

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# 11

# Case Studies

*Jace Wolfe, Erin C. Schafer, and Mila Morais*

The following sections illustrate a variety of cases (some simple and others complex) in which cochlear implant programming influenced recipients' outcomes. The purpose of these cases is to reinforce the content covered in the previous chapters.

## CASE STUDY 1: BASIC EXAMPLE OF CREATING PROGRAMS FOR AN ADULT RECIPIENT

This case study provides a basic example of how programs are typically created for an adult cochlear implant recipient. This adult recipient received a Cochlear Nucleus Freedom cochlear implant for her right ear when she was 65 years old. She had a history of hereditary, progressive hearing loss with an onset at 45 years of age. She used hearing aids for both ears for the 13 years preceding cochlear implantation of her right ear. She reported that she was unable to understand speech over the telephone for 3 years prior to receiving a cochlear implant. Prior to implantation, she scored 32% correct on AzBio sentences in quiet (60 dBA presentation level) in the binaurally aided condition, and 20% and 16% correct in the right and left aided conditions, respectively.

Using typical procedures, a program was created at the initial activation of her sound processor. Her T levels were measured on every third electrode and also on electrodes 1 through 3 using a conventional Hughston-Westlake audiometric technique. T levels were then decreased by 6 clinical units to account for the possibility of suprathreshold responses, which are fairly common on the first day of implant use. T levels of unmeasured electrodes were interpolated based on the values obtained on the measured channels. Then, C levels were set two clinical units higher than the measured T levels across the electrode array (Figure 11–1). The volume control range was set to 50%, and the volume control position was set to the minimum setting. The Nucleus 6 sound processor was activated in live speech mode, and the volume control was set to a setting of "7." Next, C levels were globally increased while the recipient listened to conversational-level speech. C levels were adjusted to the recipient's most comfortable listening level (MCL). Additionally, C levels were balanced in loudness, two channels at a

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time across the electrode array, which resulted in a C-level profile that was flatter than the T-level profile (Figure 11–2).

With this program, the recipient was able to hear all 6 Ling sounds presented from 15 feet away. The eCAP (NRT) was present on all electrodes and were generally within close proximity to the recipient's electrical dynamic range. It should be noted that the pulse width of the signal used in the recipient's program was 37 micro-

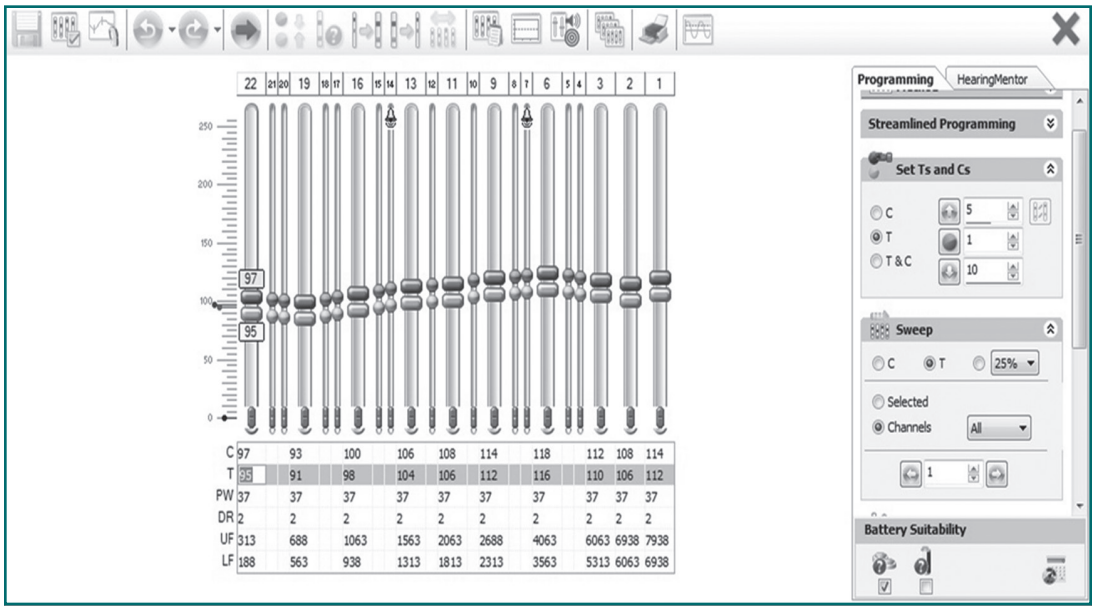
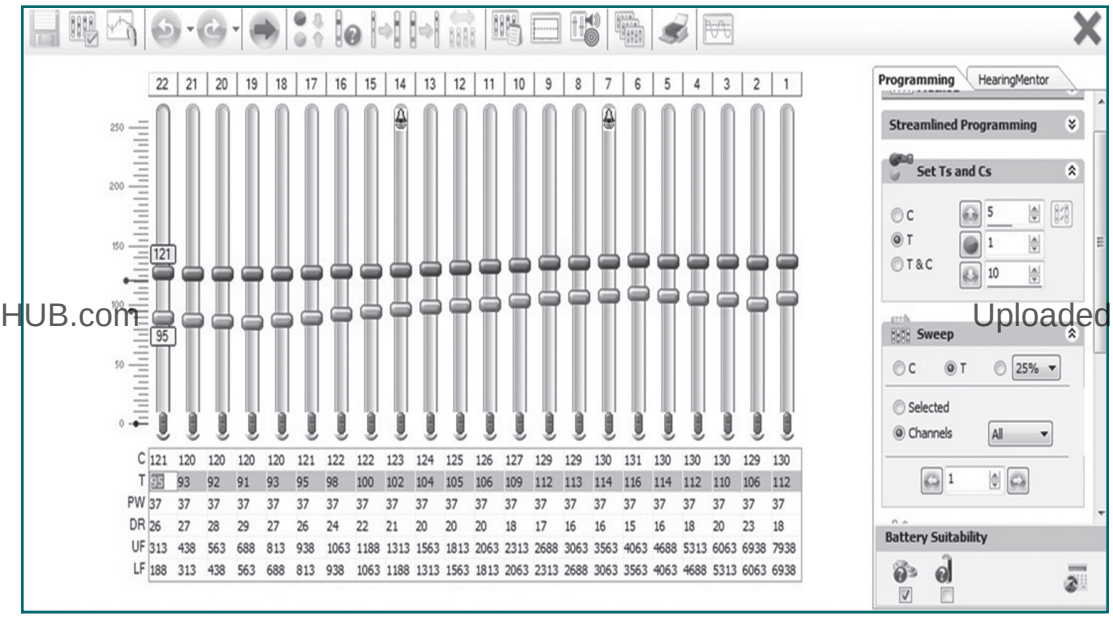



**FIGURE 11–1.** Case 1: C levels set two clinical units higher than the measured T levels across the electrode array. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–2.** Case 1: After C levels were set to the most comfortable listening level, C levels were balanced in loudness across the electrode array. *Note.* Provided courtesy of Cochlear™ Ltd.

seconds as opposed to the 25 microsecond pulse width of the NRT signal.

Figure 11–3 shows the program that was created for the recipient at her 1-week postactivation checkup. Once again, T levels were measured using conventional audiometric techniques. The clinician also measured the electrically evoked stapedius-reflex threshold (eSRT) (with the acoustic admittance probe in the nonimplanted ear) for all even-numbered electrodes along with electrode #1. The values of channels for which the eSRT was not measured were interpolated based on the values of the measured channels. Then, the C levels were globally decreased to a point at which the clinician was certain that stimulation would be perceived with a loudness that was comfortable or softer. Next, the Nucleus 6 sound processor was activated with the volume control set to a position of "7" (volume control range set to 30% to allow for more precise steps with each volume adjustment), and C-levels were adjusted to the recipient's MCL in live speech mode. After C levels were adjusted to the MCL in live speech mode, the C levels were within 15 clinical units of the measured eSRT, which is typical of most users of Cochlear Nucleus implants. While listening in live speech mode, the recipient reported the presence of a persistent low-level noise, which was eliminated by decreasing T levels globally by 4 clinical units.

The recipient returned 3 weeks later for a 1-month postactivation assessment and cochlear implant programming. She reported a significant improvement in her ability to understand speech and even noted that she had resumed use of her mobile telephone. With use of the implant alone, her sound-field detection was 25 dB HL or better from 250 to 6000 Hz, indicating excellent access to low-level sounds across the speech frequency range. Also, with use of the new implant alone, she scored 91% correct on AzBio Sentences with the right implant alone and 97% correct in the bimodal condition.

## Take-Home Points

1. Programming for this newly implanted recipient was straightforward and involved a combination of behavioral and objective measures. This case illustrates the simple nature of measuring T levels to set stimulation levels to ensure that low-level sounds are audible.
2. A variety of behavioral (globally increasing C-levels in live speech mode, balancing C-levels in loudness across the array, etc.) and

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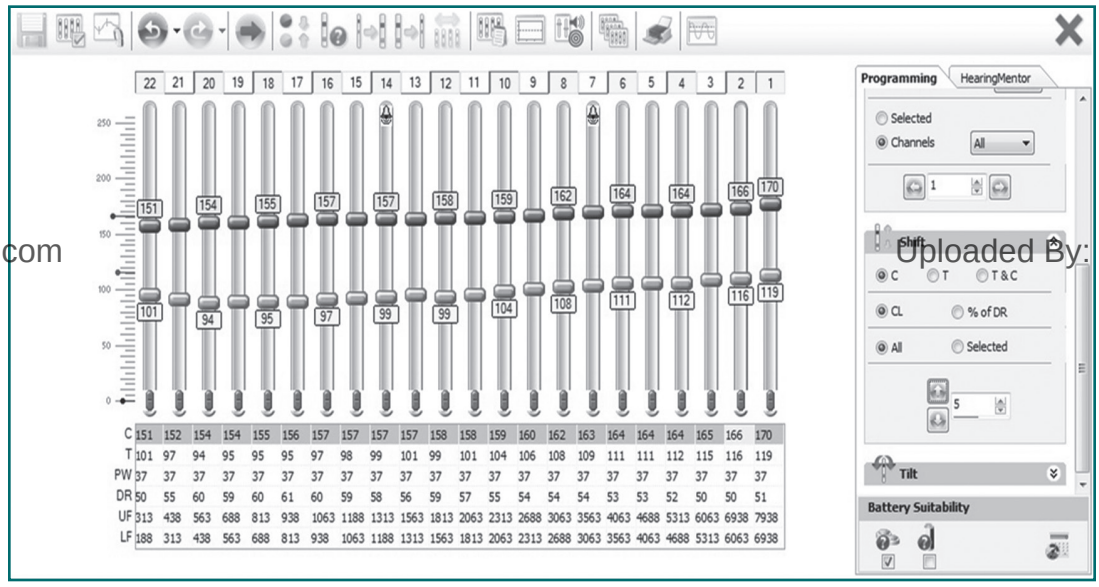



**FIGURE 11–3.** Case 1: Program at the 1-week postactivation checkup. *Note.* Provided courtesy of Cochlear™ Ltd.

objective (eSRT) may be used to determine appropriate upper-stimulation levels.

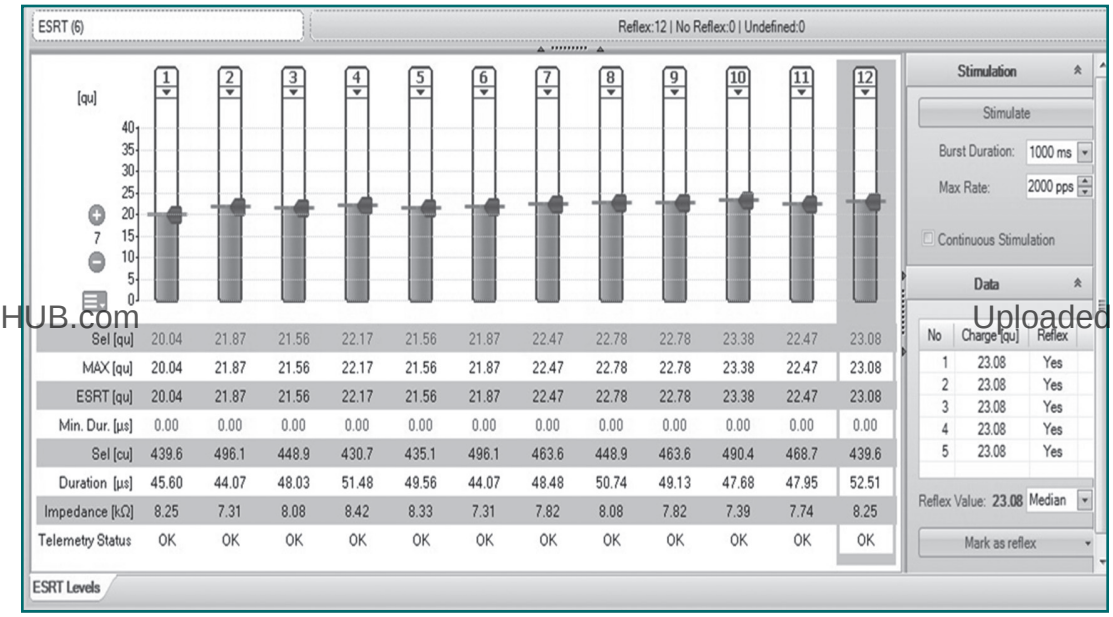
3. NRT may be used to cross-check the stimulation levels the recipient is using. As discussed, NRT responses typically fall within or toward to upper part of the recipient's electrical dynamic range.

## CASE STUDY 2: ESTABLISHING A PROGRAM FOR AN ADULT VIA OBJECTIVE MEASURES

This 63-year-old woman had bilateral profound hearing loss and a received MED-EL Concert implant for her left ear. Her hearing loss, which had an unknown etiology, was progressive over the course of approximately 25 years. She reported limited to no benefit from hearing aids for the 4 to 5 years prior to implantation. She scored 14% correct on the AzBio sentence test in the binaurally aided condition and 6 and 9% correct in the left and right ear conditions, respectively.

Her eSRT response was measured with the MED-EL Maestro software at her 1-month postactivation cochlear implant programming appointment. The eSRT eliciting stimulus was set to a 1000 millisecond on/500 millisecond off-duty cycle in order to produce an eSRT response that was highly salient to the clinician. First, the eSRT response was measured for all 12 electrodes (Figure 11–4). Then, the MCL profile was set to match the eSRT profile and MCL values were globally decreased to a level that the clinician was confident would be comfortable for the recipient. The volume control range was set to 80 to 100%, and the default test setting was set to 100%. T levels were set to default to 10% of the MCL value, and the maplaw coefficient was set to 1000. Next, MCL values were globally increased in live speech mode until the eSRT values were reached (Figure 11–5). The patient reported that these stimulation levels provided satisfactory sound quality and clarity of speech.

She returned within 1 month for a checkup and scored 89% correct on AzBio sentences and 70% correct on Consonant-Nucleus-Consonant (CNC) word recognition assessment with use of the implant alone. Her sound-field detection thresholds were within 25 to 35 dB HL from 250 to 6000 Hz.




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**FIGURE 11–4.** Case 2: Electrically evoked stapedius-reflex threshold (eSRT) response was measured for all 12 electrodes. *Note.* Image courtesy of MED-EL Corporation.

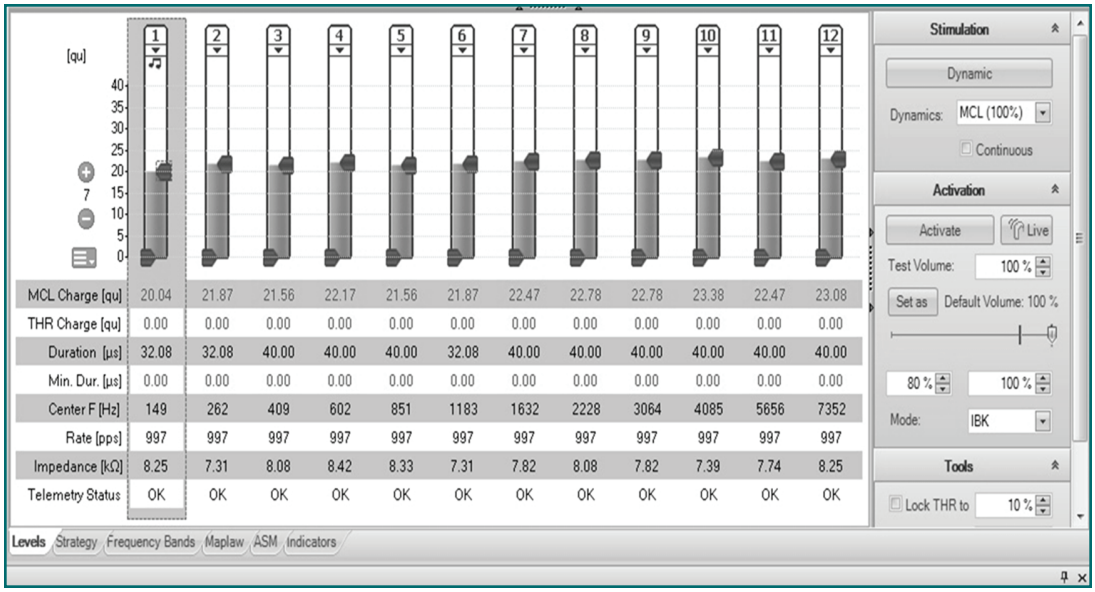



**FIGURE 11–5.** Case 2: Most comfortable listening level (MCL) globally increased in live speech mode until the eSRT values reached. *Note.* Image courtesy of MED-EL Corporation.

## Take-Home Points

1. eSRT measurements may be used to set upper-stimulation levels (i.e., MCL values), even for adult recipients who can provide excellent feedback about the loudness of the signals they receive. Studies have shown a high correlation between MCL and eSRT in MED-EL recipients (Brickley et al., 2005). In fact, because 100 dB SPL inputs are MAPped at MCL, many recipients are satisfied with MCL values set to their eSRTs. Lower-input levels are MAPped to upper-stimulation level for Advanced Bionics and Cochlear recipients, and although a high correlation has also been demonstrated between eSRT and upper-stimulation level for these recipients, upper-stimulation levels are typically set a little lower than eSRT values.
2. MED-EL utilizes a logarithmic MAPping function (maplaw), which estimates the amount of stimulation provided for low-level sounds on the basis of measured MCL values. Many MED-EL recipients are satisfied with their access to low-level sounds when T levels are estimated from the MCL value.

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## CASE STUDY 3: CREATING PROGRAMS FOR A 1-YEAR-OLD IMPLANT RECIPIENT

This case describes the creation of programs for a young child who was activated with a Cochlear Nucleus Freedom implant at one year. This boy was diagnosed at birth with a bilateral profound sensorineural hearing loss. He received limited to no benefit from binaural hearing aid use and was implanted when he was 11 months old.

At his cochlear implant activation, the child's Auditory Verbal therapist (AVT) was present and assisted in the visual reinforcement audiometry (VRA) procedure, which was used to measure minimal response levels (i.e., responses that were most likely slightly suprathreshold) for several electrodes. The T levels of unmeasured channels were estimated using interpolation. Then, T levels were decreased globally by six clinical units to account for the high likelihood that the behavioral responses during T-level assessment were likely minimal response levels.

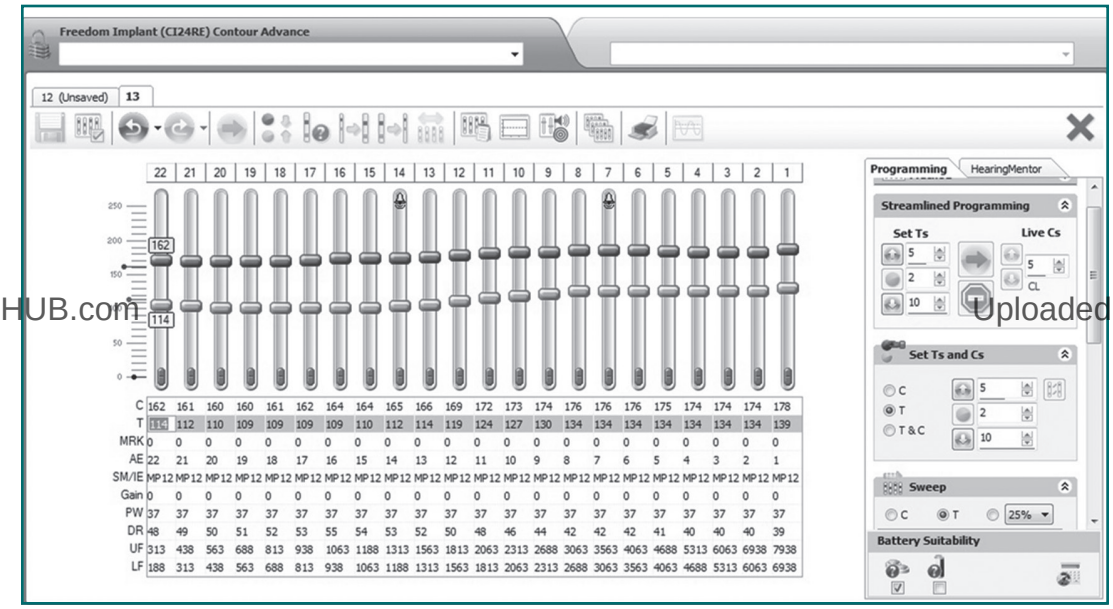
Next, the AVT engaged the child with entertaining activities (e.g., blowing bubbles, playing

with toys with flashing toys, showing a video on an iPad), while the audiologist measured the eSRT across the electrode array. Once eSRT responses were obtained, the unmeasured channels were interpolated, the C levels were globally decreased by 60 clinical units, and the processor was activated in live speech mode. The clinician and the AVT closely observed the child for adverse reactions while C levels were globally increased to 15 clinical units below the eSRT measures (Figure 11–6). The child was able to use this new program with no aversive reactions or discomfort. Additionally, the child was able to detect all 6 Ling sounds when spoken from 15 feet away.

When the child was 2 years old, conditioned play audiometry (CPA) was used to measure T levels across the electrode array. Again, the T levels of unmeasured channels were estimated via interpolation. The stimulus pulse width was set to 37 microseconds to avoid exceeding voltage compliance limits. C levels were once again set to 15 clinical units below the level at which the eSRT was obtained (Figure 11–7). Word recognition was assessed with the Multisyllabic Lexical Neighborhood Test (MLNT) and revealed an open-set score of 92% correct. Sound-field detection thresholds were measured using CPA and were obtained between 15 to 25 dB HL from 250 to 6000 Hz. Furthermore, standardized speech and language assessments indicated that this child's expressive and receptive spoken-language abilities were better than 95% of normal hearing children of the same age (i.e., standard scores exceeding 130).

## Take-Home Points

1. Conventional, pediatric audiometric procedures (VRA and CPA) were used to establish T levels, and eSRT was used to set C levels in this 1-year-old child.
2. The programs were validated by measuring sound-field detection thresholds to ensure access to low-level sounds. Also, speech recognition was assessed through the use of linguistically appropriate open-set speech recognition measures. Finally, speech, spoken language, and auditory skill development were evaluated with norm-referenced, standardized assessments to ensure that the child's progress was commensurate with that of a typically hearing child of the same age.




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**FIGURE 11–6.** Case 3: Initial MAP where C levels were globally increased to 15 clinical units below the eSRT measures. *Note.* Provided courtesy of Cochlear™ Ltd.

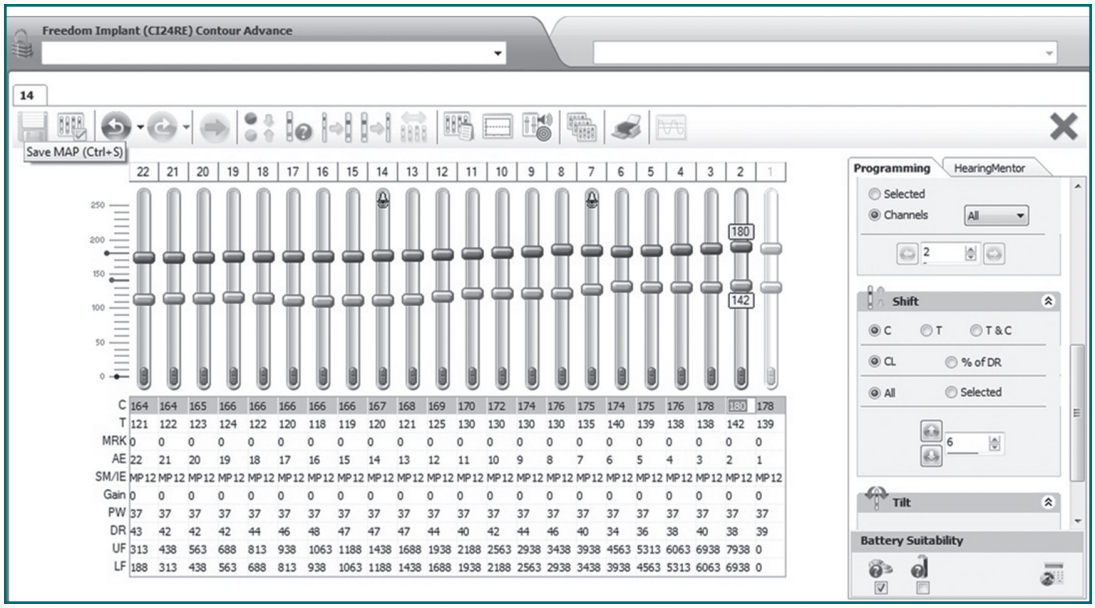



**FIGURE 11–7.** Case 3: C levels set to 15 clinical units below the eSRT. *Note.* Provided courtesy of Cochlear™ Ltd.

## CASE STUDY 4: CREATING A PROGRAM FOR A YOUNG RECIPIENT USING THE ADVANCED BIONICS HIRES 90K ADVANTAGE COCHLEAR IMPLANT

This 3-year-old recipient used an Advanced Bionics HiRes 90K Advantage cochlear implant with the Mid-scala electrode array. She passed her newborn hearing screening but was diagnosed with a mild to moderate, bilateral hearing loss when she was 16 months old. A subsequent MRI indicated abnormally large vestibular aqueducts for both ears. Over time, her bilateral hearing loss progressed to the moderately severe to profound sensorineural hearing loss range, and as a result, a cochlear implant was recommended for the right ear while she continued to use a hearing aid for the left ear. Postoperative audiometric assessment indicated a profound hearing loss for the implanted ear.

The eSRT procedure was attempted in order to determine appropriate M-level settings. Previous research has suggested a strong relationship between the eSRT and M levels for Advanced Bionics users (Buckler & Overstreet, 2003; Wolfe & Kasulis, 2008). The child was restless during & eSRT assessment, so it was unlikely that the measure could have been completed for a large number of single channels. As a result, the eSRT was measured using Speech Bursts as the eliciting stimulus on channels 1 through 4, 5 through 8, 9 through 12, and 13 through 16 (Figure 11–8). In most cases, M-level profiles (e.g., the shape of M levels across the electrode array) are typically flat and generally do not vary greatly from one channel to the next when using monopolar stimulation. As a result, the M levels based on the eSRT measures were arbitrarily smoothed to provide a flatter profile. Next, the M levels were decreased considerably to an overall level at which the clinician was confident that the child would not be uncomfortable. Then, M levels were globally increased until they were about 90% of the measured eSRT values (Figure 11–9). T levels were left at the default setting of 10% of the M-level values. The child exhibited no signs of discomfort with use of the new program.

At a 1-month follow-up assessment, soundfield detection thresholds were measured between 25 to 40 dB HL from 250 to 6000 Hz. Holden,

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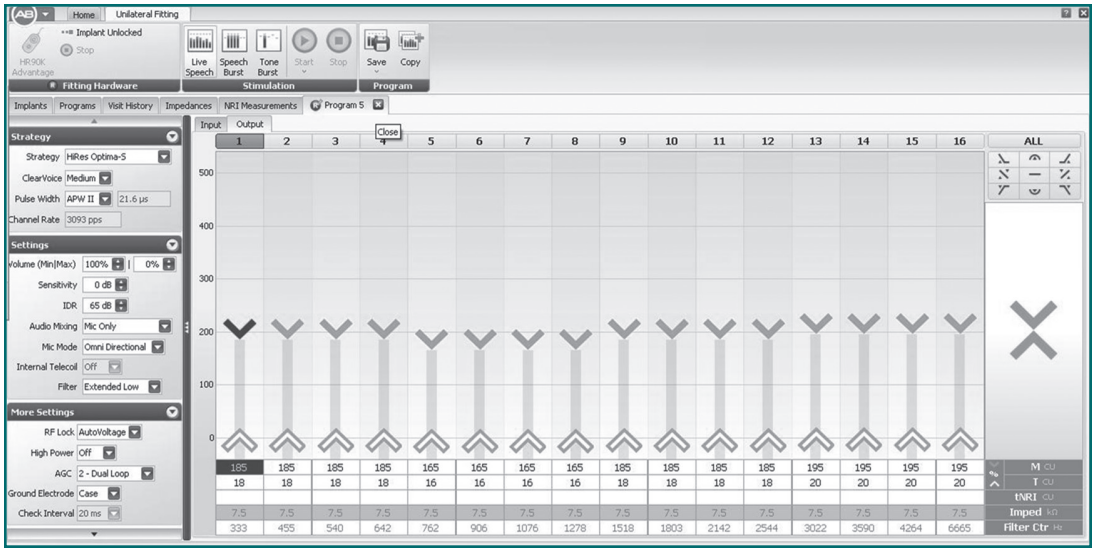
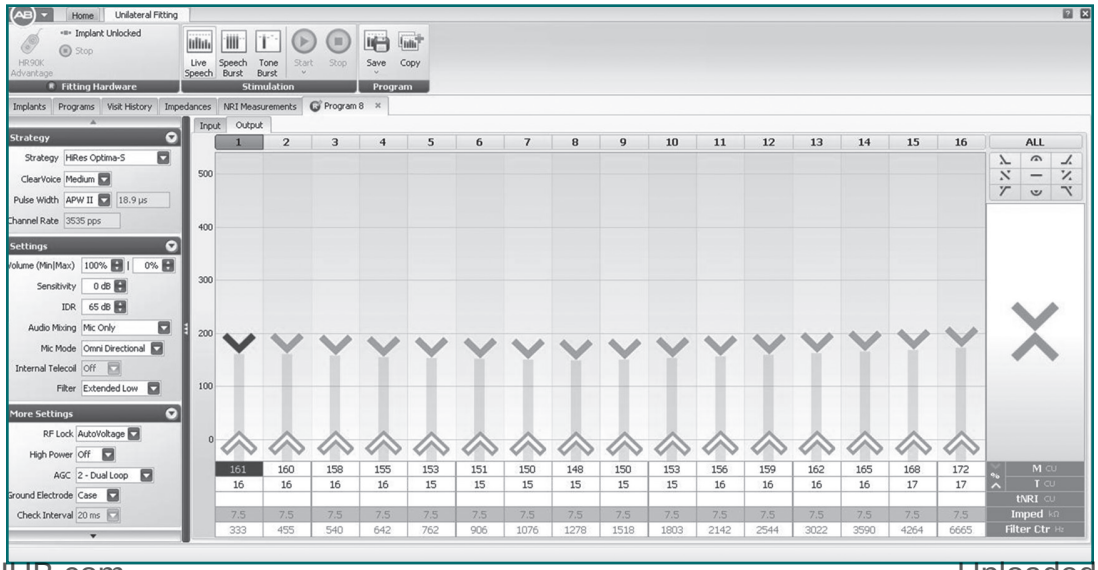



**FIGURE 11–8.** Case 4: The eSRT measurements with Speech Bursts on channels 1 through 4, 5 through 8, 9 through 12, and 13 through 16. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com




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**FIGURE 11–9.** Case 4: Program with M levels were globally increased until they were about 90% of the measured eSRT values. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

Reeder, Firszt, and Finley (2011) have shown that measurement of T levels is sometimes necessary to obtain sound-field thresholds in the 20 to 25 dB HL range. VRA was used to measure T levels to single-channel "tone burst" stimuli for several channels across the electrode array. The values of unmeasured channels were manually interpolated based on the T-level values of the measured channels. Then, T levels were globally decreased by 10 charge units to account for minimal response levels and to reduce the likelihood that the child would perceive a low-level noise (e.g., buzzing, frying, static), which may occur when T levels are set too high in a CIS-based program (Figure 11–10).

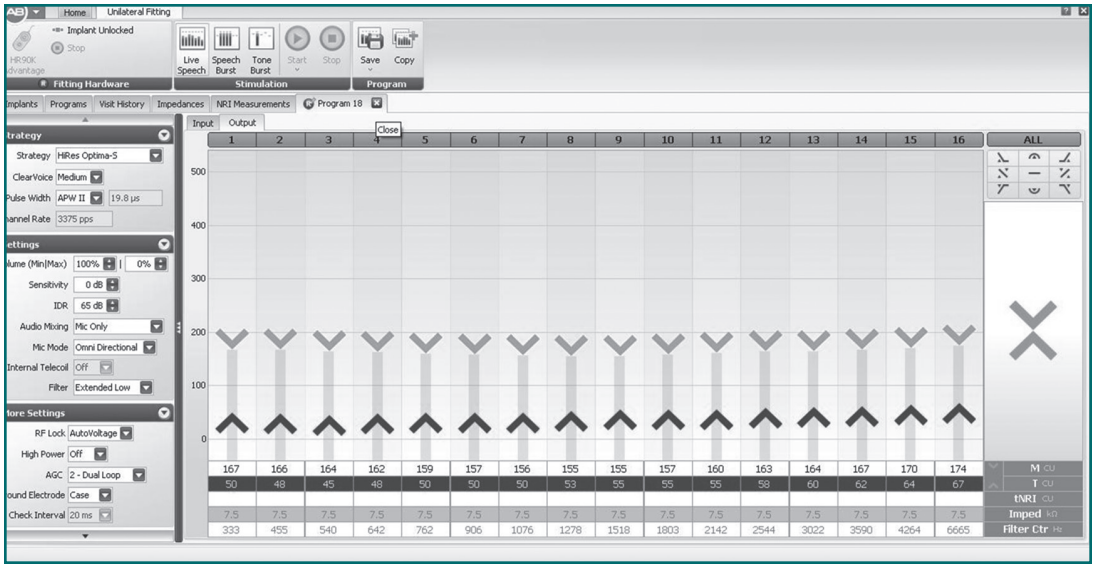



**FIGURE 11–10.** Case 4: T levels globally decreased by 10 charge units to account for minimal response levels and the perception of low-level noise. *Note.* Image provided courtesy of Advanced Bionics.

Two weeks later the child returned, and soundfield detection thresholds were remeasured and obtained in the 20 to 25 dB HL range from 250 to 6000 Hz. Additionally, the child was able to identify and repeat all six Ling sounds without visual cues, and the child's auditory-verbal therapist reported that the child was making excellent progress with her speech and language development.

## Take-Home Points

1. This case is another example of how the eSRT may be used to determine upper-stimulation levels for young children and also illustrates how Speech Bursts may be used to elicit the eSRT across the array in an expeditious manner. When Speech Bursts are used to measure the eSRT or M levels via psychophysical loudness scaling, the upper-stimulation level contour may vary rather significantly from one group of four channels to the next. In this case, it is likely that the channels will not be balanced in loudness across the array. With cooperative recipients, the eSRT should be measured with single-channel tone bursts in order to more effectively capture the contour of upper-stimulation levels from one channel to the next. Loudness balancing should then be completed on a 2-channel basis to equalize loudness across the array. For recipients who cannot provide reliable feedback about the loudness of the signal they receive, the clinician should smooth or flatten the M-level profile.
2. Many recipients possess good access to low-level sounds when T levels are set to 10% of the M-level value. However, some recipients exhibit limited access to low-level sounds (e.g., 20–25 dB HL) when T levels are estimated from the M-level values. In these cases, it may be helpful to measure T levels to optimize access to low-level speech and environmental sounds.
3. The clinician should ensure that the recipient does not experience a constant low-level noise (e.g., buzzing, frying noise, static), which may occur when T levels are set too high in a CIS-based program. When recipients provide verbal feedback that a noise exists, the clinician may decrease T levels globally until the noise subsides. For recipients who cannot provide reliable feedback about what they perceive, T levels may be decreased globally by 10 charge units from the measured values in order to reduce the likelihood of a continuous audible noise.

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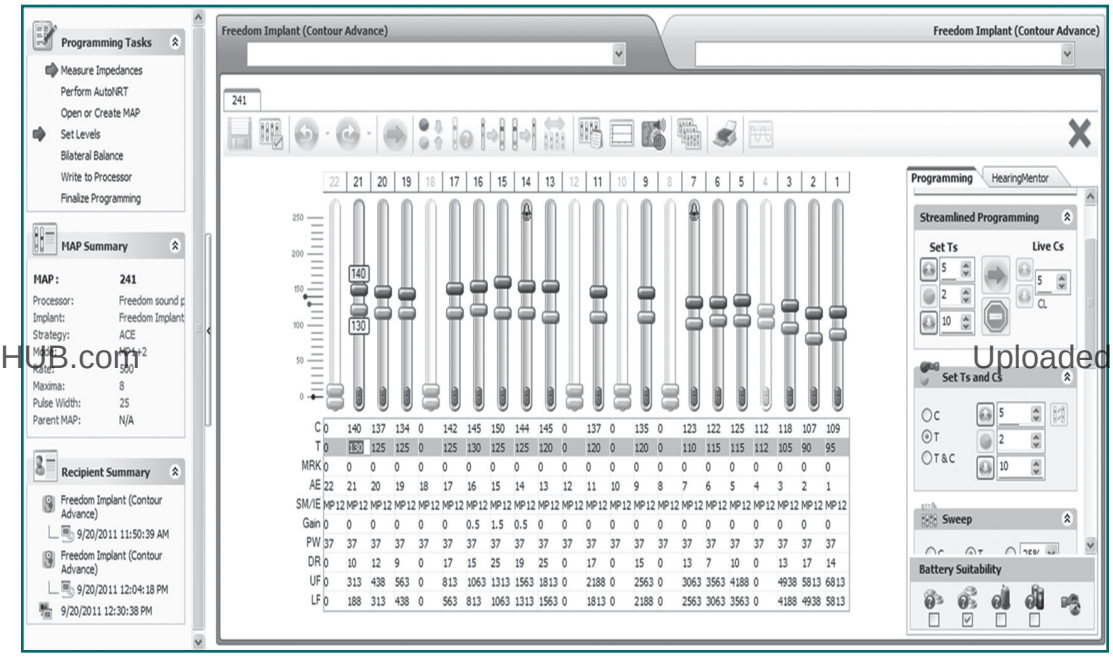
## CASE STUDY 5: INADEQUATE STIMULATION LEVELS AND NARROW ELECTRICAL DYNAMIC RANGE

This 8-year-old boy was diagnosed at birth with severe to profound sensorineural hearing loss in both ears, and a CT scan indicated a Mondini's deformity bilaterally. He received bilateral Cochlear Freedom implants at an early age (right ear at 12 months old; left ear at 17 months). He was first seen when he was 7 years old for audiologic assessment and programming after moving to our area. His mother reported that he had a difficult time monitoring the level of his own voice. She also noted that his speech-recognition performance was poor, and his speech and language levels were significantly delayed.

Sound-field detection thresholds indicated relatively good access to low-level sounds with unilateral warble-tone thresholds of 30 dB HL or better from 250 to 6000 Hz. Bilateral word recognition was 16% correct on the Lexical Neighborhood Test (LNT) with the target words presented at 60 dBA.

Electrode impedances were within normal limits for both ears, and all external cochlear implant equipment appeared to be functioning appropriately. His programs were uploaded from each of his sound processors. From a quick review of his programs, it was evident that his electrical dynamic range was quite narrow for both ears. Figure 11–11 shows the stimulation levels for the right processor. As shown, several electrodes possessed a single-digit clinical unit electrical dynamic range. Typically, the electrical dynamic range of a Nucleus recipient falls within 25 to 60 clinical units with most being around 40 clinical units. He also had several electrodes disabled for each ear. When his mother was queried about the methods that were used to create the implant programs he was using, she noted that programming sessions usually consisted of conditioned play audiometry to set T levels and an attempt at psychophysical-loudness measures to set C levels to the programming tonal stimulus. His mother reported that the patient had never shown any signs of discomfort to stimulation from the implant.

During the appointment, a number of different activities (e.g., throwing plastic fish into a bowl




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**FIGURE 11–11.** Case 5: Stimulation levels for the right processor. *Note.* Provided courtesy of Cochlear™ Ltd.

of water, throwing toy cars down a poster tube, placing birthday cake candles into clay, throwing M&Ms into a cup) were employed to measure T levels, and valid and reliable T levels were obtained across the electrode array. The eSRT and eCAP (NRT) responses could not be obtained for either ear prior to the recipient asking us to stop stimulation because it was too loud. C levels were set to "hug" the T-level profile, and the processor was activated. We gradually increased C levels as the patient played a game on an electronic tablet with an auditory-verbal therapist who was supporting the programming session. As the stimulation levels were increased, the C-level profile was gradually "flattened" relative to the T-level profile (i.e., the C-level profile possessed less variation in level across the array when compared to the T-level profile). C levels were increased so that an electrical dynamic range of at least 25 clinical units was obtained for each electrode. The child was able to tolerate the higher stimulation levels without any discomfort, and he was pleased with the sound quality of his new program. He was scheduled to return to the clinic for audiologic assessment and programming several times over the course of the next 9 months. During this time period, his C levels were increased significantly without discomfort (see Figure 11–12 for program created 9 months after the initial programming session). Although the eSRT was never successfully recorded, the eCAP (NRT) response was recorded and validated the increased stimulation levels. Specifically, C levels were not substantially higher than the eCAP thresholds.

His mother reported a substantial improvement in his vocal quality and speech recognition. Speech and language assessment showed an 18-month improvement in expressive language in a 6-month interval between successive evaluations. Furthermore, he scored 80% correct on the Lexical Neighborhood monosyllabic word recognition test (60 dBA) with bilateral implants.

## Take-Home Points

1. The clinician should be concerned when a recipient's electrical dynamic range is considerably narrower or wider than the typical range. Although exceptions exist, the majority

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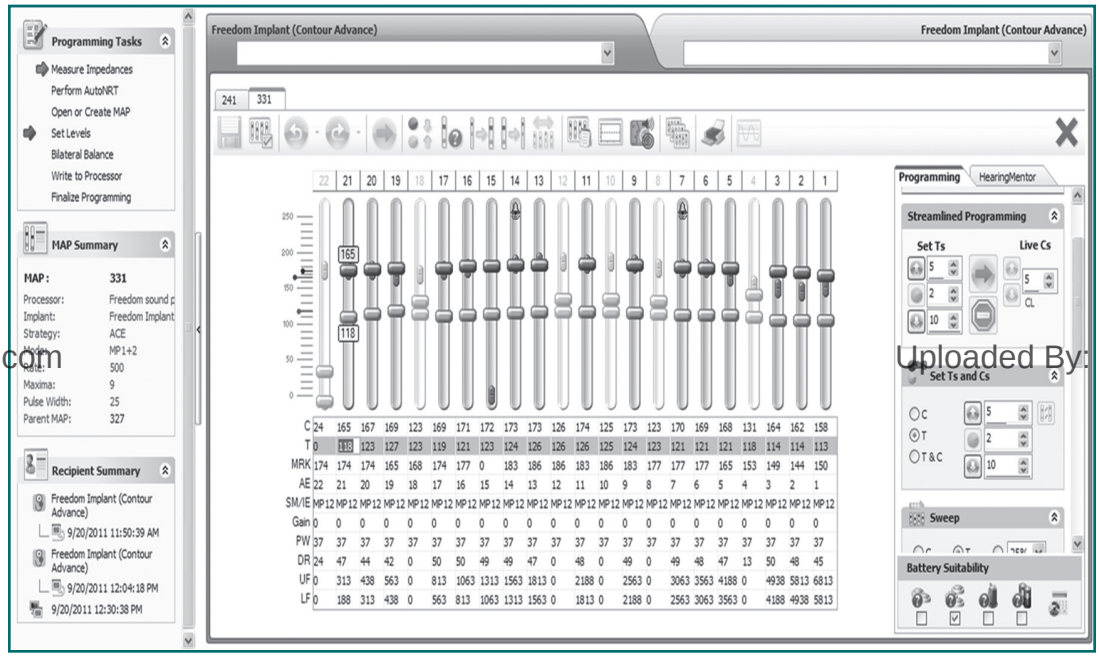



**FIGURE 11–12.** Case 5: Program created 9 months after the initial programming session. *Note.* Provided courtesy of Cochlear™ Ltd.

of Cochlear Nucleus implant recipients have an electrical dynamic range between 25 to 60 clinical units. This recipient's electrical dynamic range was significantly lower than the typical value for most of his electrodes.

2. If the clinician believes that stimulation levels are too low, then multiple programming sessions may be scheduled to gradually increase stimulation over time.
3. When possible, objective measures should be used as a guide to ensure that stimulation levels are appropriate. The eSRT, which is the objective measure with the best correlation to upper-stimulation level, was not present for this recipient. However, the eCAP (NRT) was present and could be used to determine that the upper-stimulation levels were not unreasonably high. Although exceptions exist, the clinician should be particularly concerned when C levels exceed to NRT by more than 30 to 40 clinical units (20–25 clinical units when using a 37 microsecond PW in the CI program). In most cases, C levels are within ± 20 clinical units of NRT when using a program with a 25 microsecond PW.
4. The programming clinician should frequently conduct speech-recognition assessments to ensure adequate performance and progress associated with the programming process. Also, programming clinicians should actively communicate with the child's auditory-verbal therapist (or speech-language pathologist) and parent to ensure that the child is making satisfactory progress and that there are no hearing-related concerns.

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## CASE STUDY 6: USING OBJECTIVE MEASURES TO SET UPPER-STIMULATION LEVELS

An 8-year-old girl was seen for audiologic assessment and cochlear implant programming after moving to our state. She passed her newborn hearing screening but was diagnosed with a severe to profound sensorineural hearing loss at 2 years of age. She received a Cochlear Freedom implant for her right ear shortly before her third birthday and a Cochlear Freedom for her left ear at three and one-half years of age.

Although she used both cochlear implants during all waking hours, she preferred her right ear implant. Unfortunately, her right implant failed and was promptly replaced when she was 6 years old. Her mother reported that the recipient's speech sounded like "deaf speech" since the failure of the first implant. She also noted that the child had significant difficulty understanding speech in most situations.

Unilateral sound-field detection thresholds for warble tones were within normal limits from 250 to 6000 Hz. Lexical Neighborhood Test monosyllabic word recognition was 48% correct for the right ear and 36% correct for the left ear. The recipient's program is shown in Figure 11–13.

Electrode impedances were within normal limits across the array for both ears, and all of her external cochlear implant equipment was functioning appropriately. T levels were measured across the electrode array for both ears, and slight changes were made to T-level stimulation based on the recipient's responses. The eSRT was successfully measured across the electrode array with stimulation to both ears and the acoustic immittance probe in the left ear. The eSRT could not be measured when the immittance probe was placed in the right ear. It is possible that subtle and clinically insignificant changes to the middle ear may have occurred during cochlear implantation of the right ear. These changes may have increased the stiffness of the middle ear, which eliminated that ability to record the acoustic reflex using acoustic immittance measures.

The eSRT for the right ear was obtained at levels that were substantially higher than the upper-stimulation levels the recipient was using in her current programs. As a result, the C-level profile was set to match the eSRT profile (Figure 11–14). Then, the processor was activated in live speech mode, and C levels were increased until they were within 15 to 20 clinical units of the eSRT thresholds (Figure 11–15). The recipient showed no signs of discomfort to the increase in upper-stimulation levels.

The family was asked to bring the recipient back to the clinic in 3 months for audiologic assessment and cochlear implant programming. The family reported a significant improvement

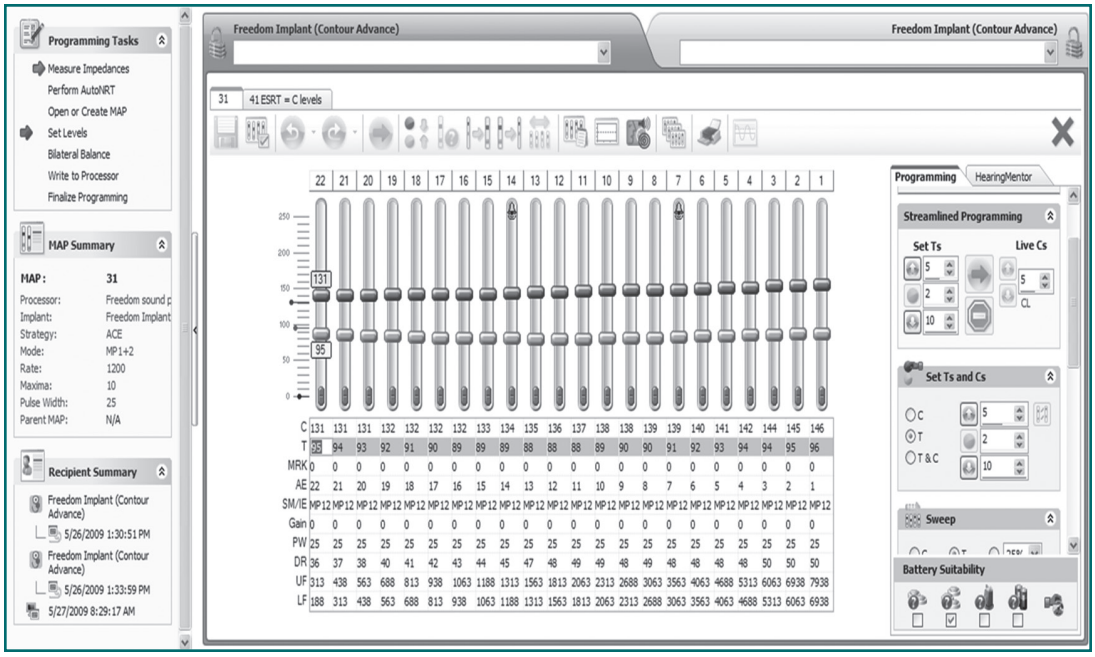
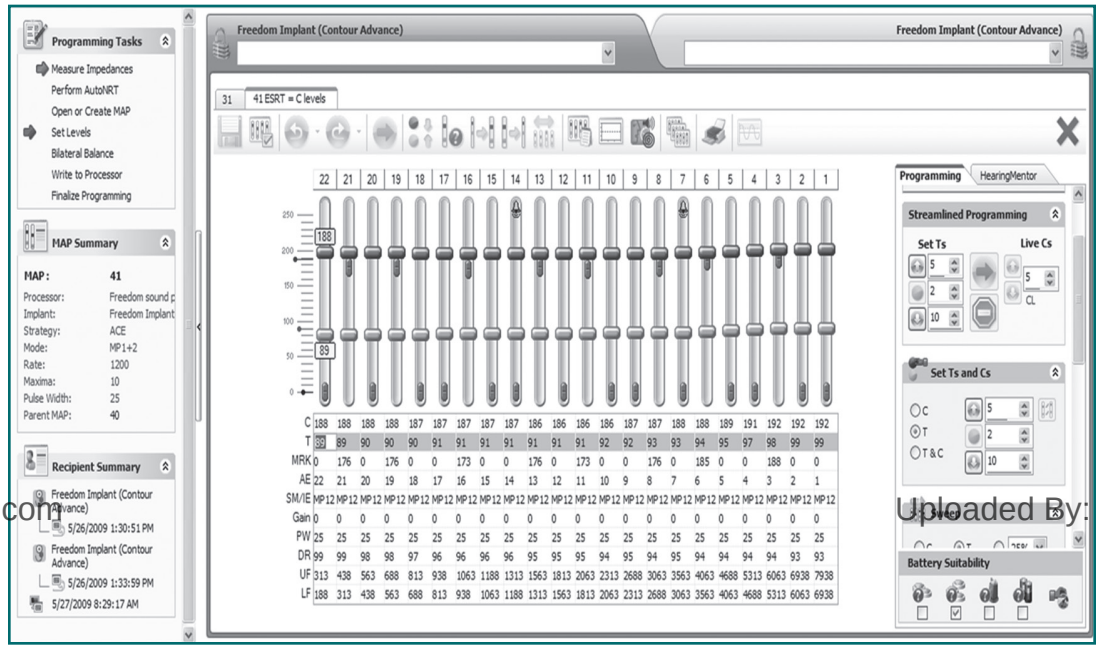



**FIGURE 11–13.** Case 6: Recipient's initial program. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–14.** Case 6: The C-level profile was set to match the eSRT profile. *Note.* Provided courtesy of Cochlear™ Ltd.

in the recipient's speech, language, and auditory abilities. Lexical Neighborhood Test word recognition scores were 84% and 72% for the right and left ears, respectively.

## Take-Home Points

1. Upper-stimulation levels greatly influence sound quality of speech and other external

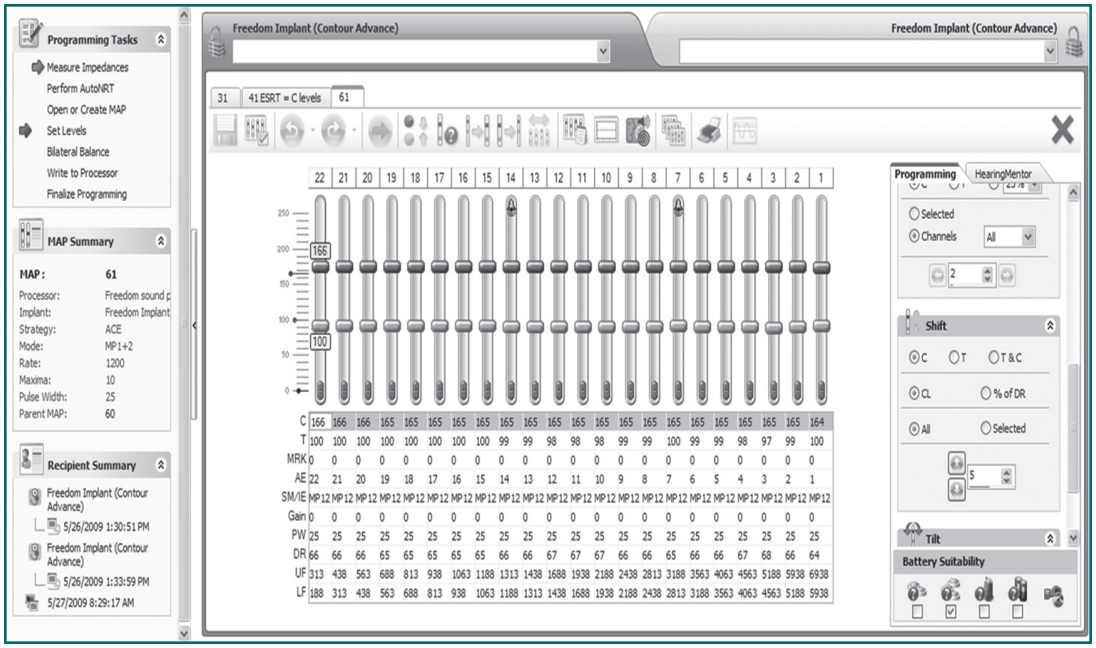



**FIGURE 11–15.** Case 6: C levels were increased until they were within 15 to 20 clinical units of the eSRT thresholds. *Note.* Provided courtesy of Cochlear™ Ltd.

sounds, speech recognition, the overall loudness of the user's listening experience, and the user's ability to monitor his or her own voice.

2. Infants and young children are incapable of providing reliable feedback pertaining to optimal loudness of the signal(s) from their cochlear implants. Additionally, the concept of loudness scaling is often difficult to grasp for older children with congenital deafness and many adults with long-term deafness. The eSRT is the best objective measure for determining optimal upper-stimulation levels for implant recipients who struggle with loudness judgments.

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## CASE STUDY 7: CREATING A PROGRAM FOR A RECIPIENT WITH MULTIPLE DISABILITIES

This 4-year-old boy received a cochlear implant in his right ear at 21 months old in his left ear at 26 months. He transferred to our clinic after moving from another state. This patient has a history of cerebral palsy and neurological/cognitive deficit. He s limited mobility, so he uses a custom-fitted motorized chair that his mother operates. He has modest head control, which allows for a head turn to sound. This patient has no spoken expressive language, and difficulties with fine motor control prevent the use of sign language. However, he does use an augmentative communication board, which revealed that he has developed a robust receptive language capacity. Furthermore, he can express himself by pointing to pictures in a closed set on the board.

Figure 11–16 shows the program the recipient was using for his right ear when he arrived to our clinic. His mother reported that his previous programming sessions consisted primarily of VRA in an attempt to evaluate his responses to programming stimuli. She also noted that his clinician typically measured his "nerve response," which likely referred to the measurement of the eCAP in the NRT module. One remarkable characteristic of his program was the relatively erratic change in C levels that existed across the electrode array.

At his initial appointment, VRA was used to measure T levels, which were then set five clinical units lower than his behavioral responses to account for minimal response levels. Next, eSRT

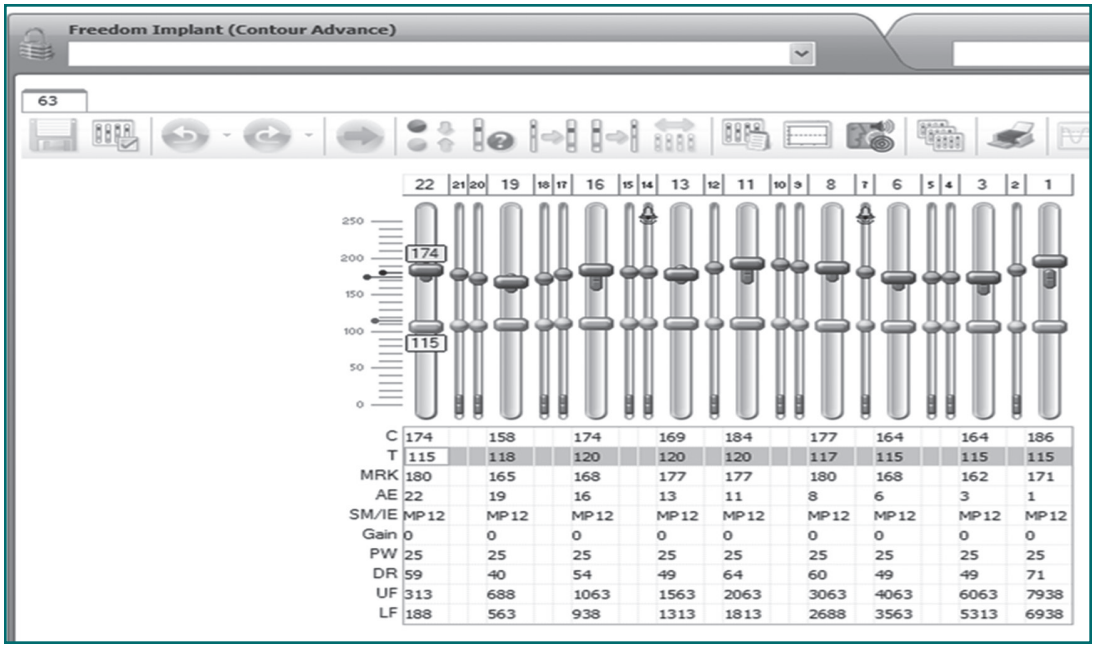



**FIGURE 11–16.** Case 7: Initial program for the right ear implant when recipient arrived to our clinic. *Note.* Provided courtesy of Cochlear™ Ltd.

was measured with the acoustic immittance probe in the left ear. No eSRT response was obtained prior to reaching voltage compliance limits. The immittance probe was, then, placed in the right ear (ipsilateral to the eliciting stimulus), and the eSRT was obtained across the electrode array (Figure 11–17). C levels were set to the eSRT profile and were globally adjusted so that they were 18 clinical units below the eSRT (Figure 11–18). Additional increases of C levels resulted prominent eye blinking in response to environmental stimuli (e.g., knocking on a desktop, speech). Obviously, frequent blinking in response to speech and environmental sounds is not a typical response, and clinicians should avoid stimulation levels that elicit a persistent blinking response. Oftentimes, C levels fall within 10 to 20 clinical units of the eSRT for Nucleus users, so the offset between C levels and eSRT for this recipient would seem to be appropriate.

Several characteristics of this new program suggest that it is likely to be appropriate for the recipient. First, the C levels are based on the eSRT measures, and several studies have suggested a strong relationship between eSRT and upper-stimulation level. Second, the C level configuration (i.e., profile or shape) is relatively flat rather than erratic as with the previous program. Although T-level profiles may show change from one part of the array to another, in the typical case, upper-stimulation levels are rather flat or possess fairly gradual changes across the array. Third, the electrical dynamic range is characteristic of a typical recipient of a Cochlear Nucleus Freedom implant (i.e., typical electrical dynamic range of 25 to 60 clinical units). Finally, the NRT typically falls within the electrical dynamic range of the recipient. Previous research has suggested that NRT is higher than T level and frequently within ± 20 clinical units of C level.

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## Take-Home Points

1. A combination of behavioral (i.e., behavioral observation audiometry and VRA) and objective (eSRT) measures may be used to create a program for a recipient who may be unable to provide specific verbal feedback about the signal provided by the cochlear implant.
2. It is somewhat uncommon to see an erratic change in upper-stimulation levels across the

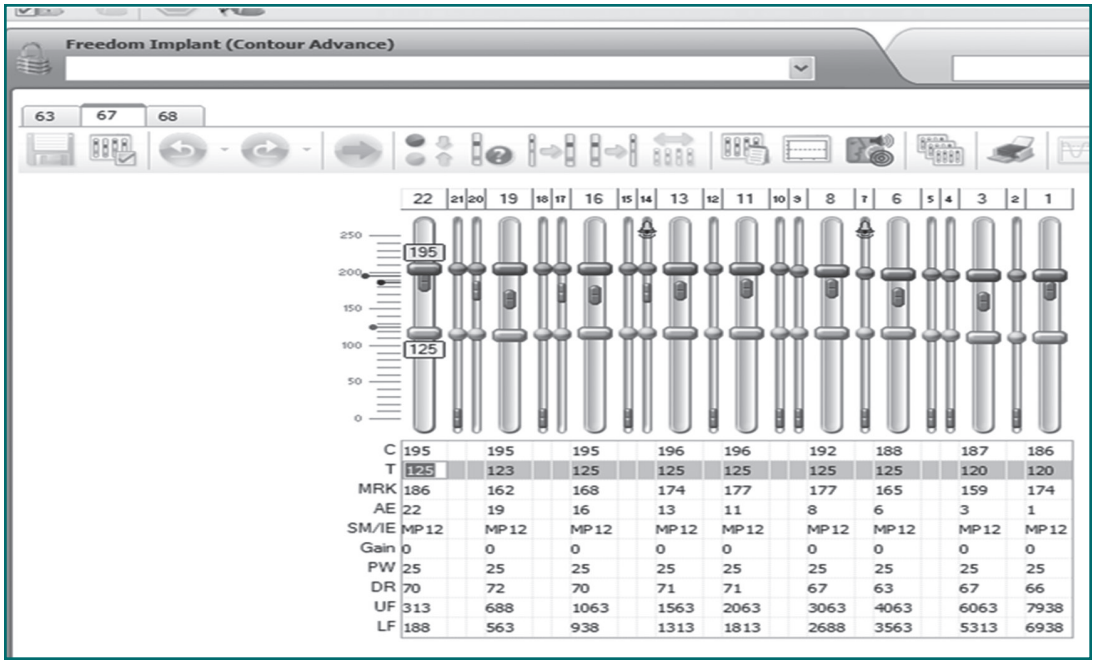



**FIGURE 11–17.** Case 7: The eSRT across the electrode array. *Note.* Provided courtesy of Cochlear™ Ltd.

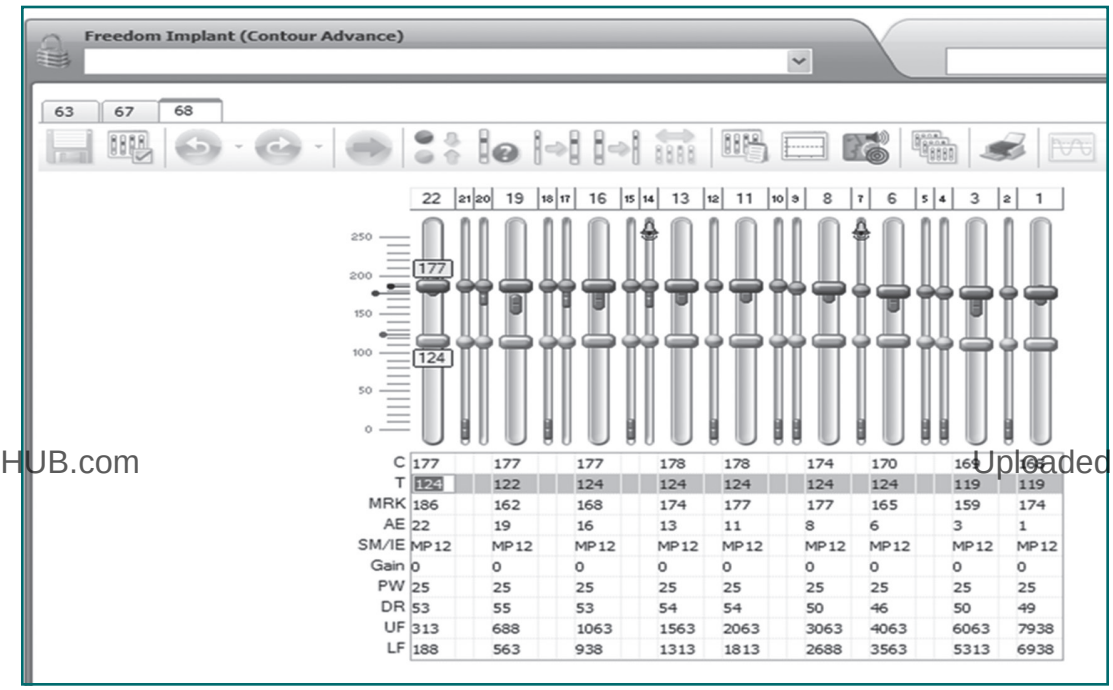



**FIGURE 11–18.** Case 7: C levels set to the eSRT profile and globally adjusted to 18 clinical units below the eSRT. *Note.* Provided courtesy of Cochlear™ Ltd.

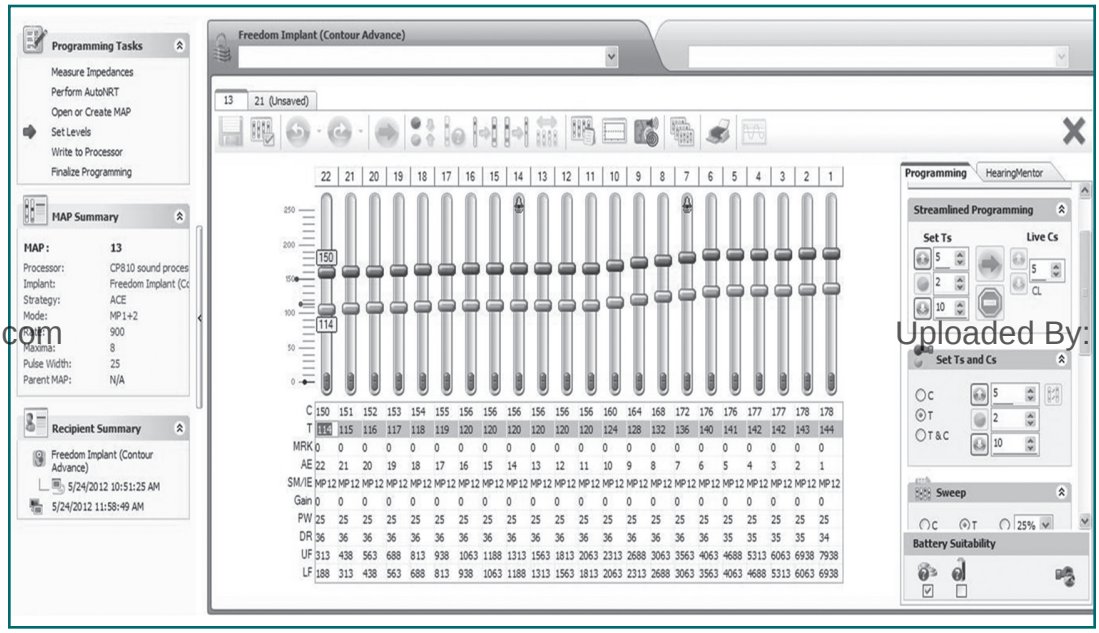
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electrode array as seen in the initial program this recipient was using (see Figure 11–16). Indeed, the eSRT suggested that the upper-stimulation levels should be flatter than what was used in the initial program.

## CASE STUDY 8: USE OF OBJECTIVE MEASURES AS A GUIDE TO AVOID OVERSTIMULATION

A 19-month-old boy was seen for audiologic assessment and programming after transferring to our center from another state. He was diagnosed with bilateral severe to profound sensorineural hearing loss shortly after birth and received a Cochlear Nucleus Freedom cochlear implant for his right ear on his first birthday. His older sister is a bilateral cochlear implant user, but his family has been reluctant to pursue bilateral cochlear implantation for him because he is resistant to device use. Specifically, his mother reported that he immediately pulls his sound processor off his head as soon as it is put in place, even when the volume control of his Nucleus 5 sound processor was set to the minimum setting of "1." She noted that they try to distract him by engaging him with games, movies, or food while placing the device on his head, and occasionally they are able to get him to keep the device on for a minute or two. However, she noted that he will not use it for prolonged periods of time, and he expresses signs of discomfort (e.g., prominent eye blinking, crying, etc.) when the device is placed on his head. She did report that he will keep the device on his head when the processor is powered off (i.e., without stimulation). The cochlear implant program he was using when he arrived at our clinic is shown in Figure 11–19.

Electrode impedances were within normal limits across the electrode array, although absolute values were relatively high. It should be noted that electrode impedances are known to increase globally with periods of nonuse. T-level measurements were attempted via CPA and VRA, but the recipient would not condition to either task. Otoscopy and tympanometry indicated patent PE tubes for both ears, so it was impossible to measure the eSRT. The eCAP response (e.g., NRT) was measured across the electrode array. Initially, the child



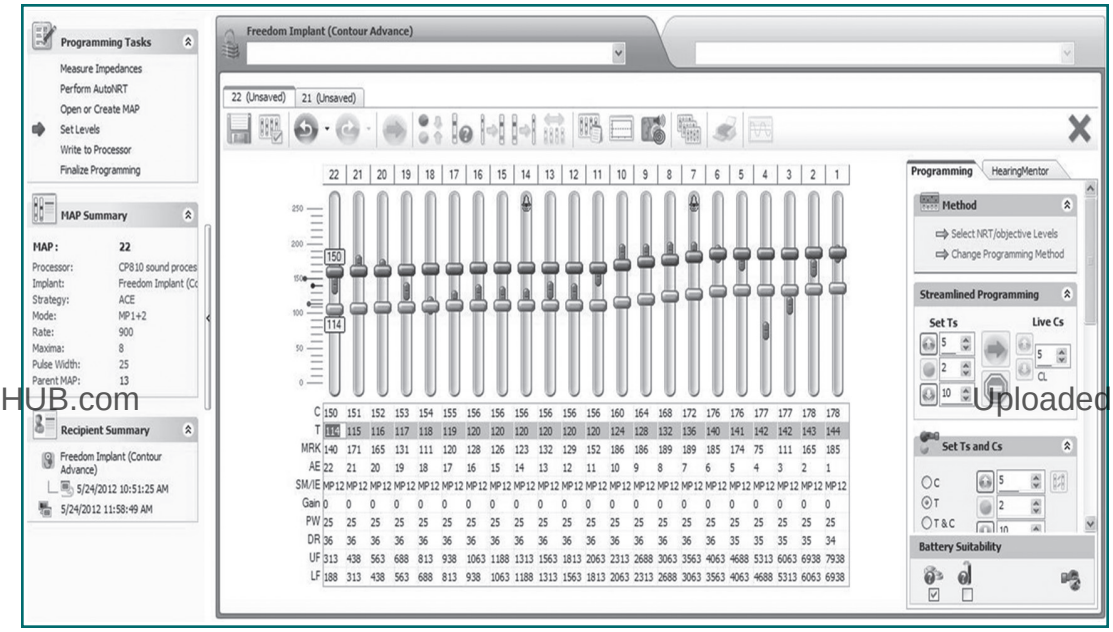

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**FIGURE 11–19.** Case 8: Cochlear implant program upon arrival to clinic. *Note.* Provided courtesy of Cochlear™ Ltd.

frequently removed the transmitting coil when stimulation levels reached the point required to elicit an eCAP response. However, an iPad tablet was used to distract the child so that NRT could be successfully measured across the array.

Figure 11–20 provides a comparison of the child's NRT responses relative to the stimulation levels he was using in his program. At first glance, it can be seen that his T levels approach or exceed his NRT responses for several electrodes (see electrodes 14–18 as well as 3 and 4). Furthermore, several of the electrodes possess C-level stimulation that is substantially higher than the NRT response. For instance, the C level of electrode 4 is 98 clinical units higher than the NRT response, and the C levels for electrodes 21 and 3 are also substantially higher than the NRT threshold. Although some well-performing implant recipients with appropriate cochlear implant programs possess C levels that are markedly higher than their NRT responses, the clinician should view this type of disparity between NRT and C levels with some concern, particularly in light of reports that the child is resistant to use the cochlear implant.

At this point, it is important to revisit the mother's report that the child would not even use the device with the volume control setting at a minimum position of "1." However, it was noted that he will wear the device when it is powered off. A quick inspection of the volume control range revealed that the range was set to the default of 20% of the electrical dynamic range. As a result, even when the volume control was set to the minimum position, the child was still receiving a substantial amount of stimulation. For example, electrode 3 possessed T levels, C levels, and NRT of 142, 177, and 75, respectively, which provided an electrical dynamic range of 35 clinical units (C level of 177 − T level of 142 = 35 clinical units). Decreasing the volume control from a maximum to a minimum setting resulted in a seven clinical unit decrease in C-level stimulation (electrical dynamic range of 35 clinical units * 20% = 7 clinical units). Consequently, when the volume control range was set to the minimum position, the child received C-level stimulation of 170 clinical units for inputs of 65 dB SPL and higher (average conversational level speech and higher). This amount



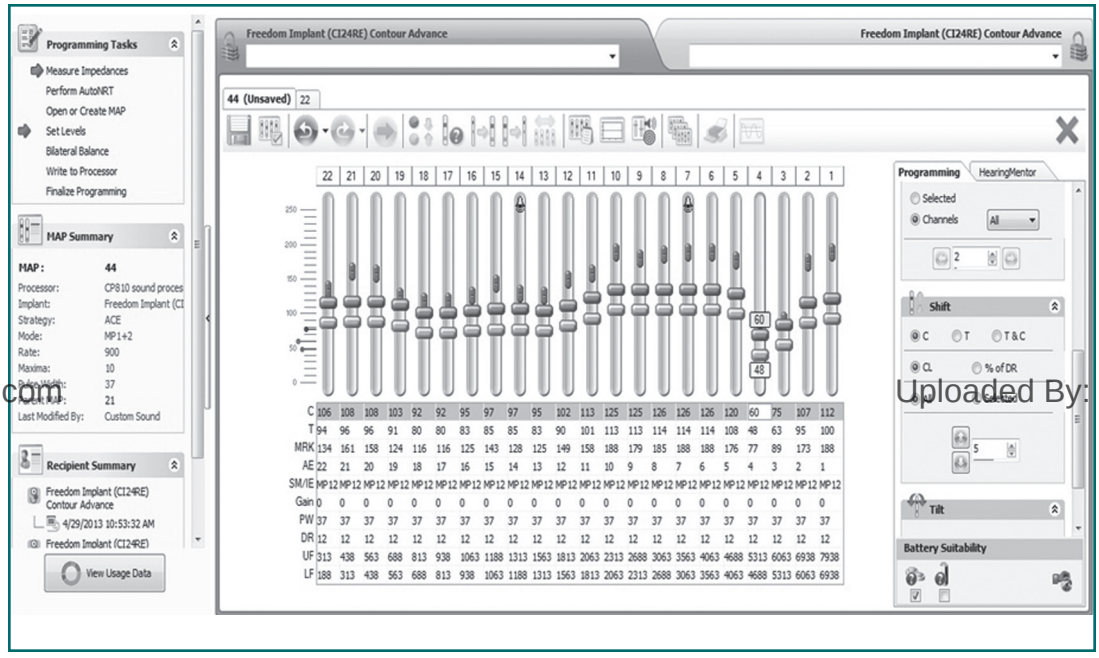

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**FIGURE 11–20.** Case 8: Comparison of the child's neural response telemetry (NRT) responses relative to the stimulation levels in his program. *Note.* Provided courtesy of Cochlear™ Ltd.

of C-level stimulation exceeded his NRT response by 95 clinical units. Additionally, he received a T-level stimulation of 142 clinical units, which meant that he would receive continuous stimulation at a level that was 47 clinical units above his NRT response.

Remember, the eCAP threshold should always be obtained at a level that is higher than the recipient's behavioral threshold for a given electrode. When T levels are set too high, the recipient may experience a constant buzzing or roaring sound. If T levels are set excessively high, this continuous noise may even be uncomfortable for the recipient.

Because this patient could not complete behavioral T-level measurements, stimulation levels were adjusted based on a combination of the NRT responses and the recipient's behavioral responses to speech. His T-level profile was arbitrarily set to roughly match that of the NRT profile (Figure 11–21). Then, C levels were set to hug (typically within 10–12 clinical units of one another) the T-level profile (see Figure 11–21). The volume control range was switched from 20% to 100%, and the volume control setting was placed at the minimum position of "1." The T levels and C levels were globally decreased to a level that was well below the NRT responses (i.e., T levels were 30 clinical units or lower than NRT responses). The objective was to set stimulation levels so that the child would not detect any stimulation upon activation of the processor. Next, the processor was activated in live speech mode with the volume control at the minimum setting. The volume control setting was gradually increased to the maximum setting, and the clinician ensured that the child showed no signs of aversiveness to the signal from the implant. In fact, the child showed no signs of detection of a signal from the implant, which is ideal when creating an initial program. Furthermore, the T levels and C levels were globally increased rather slowly in live speech mode until the child showed signs that the signal from the implant was audible. At that point, T levels, which were set 12 clinical units below C levels were left in place, and the C levels were gradually increased. The typical goal with this type if case is to achieve an electrical dynamic range of 30 to



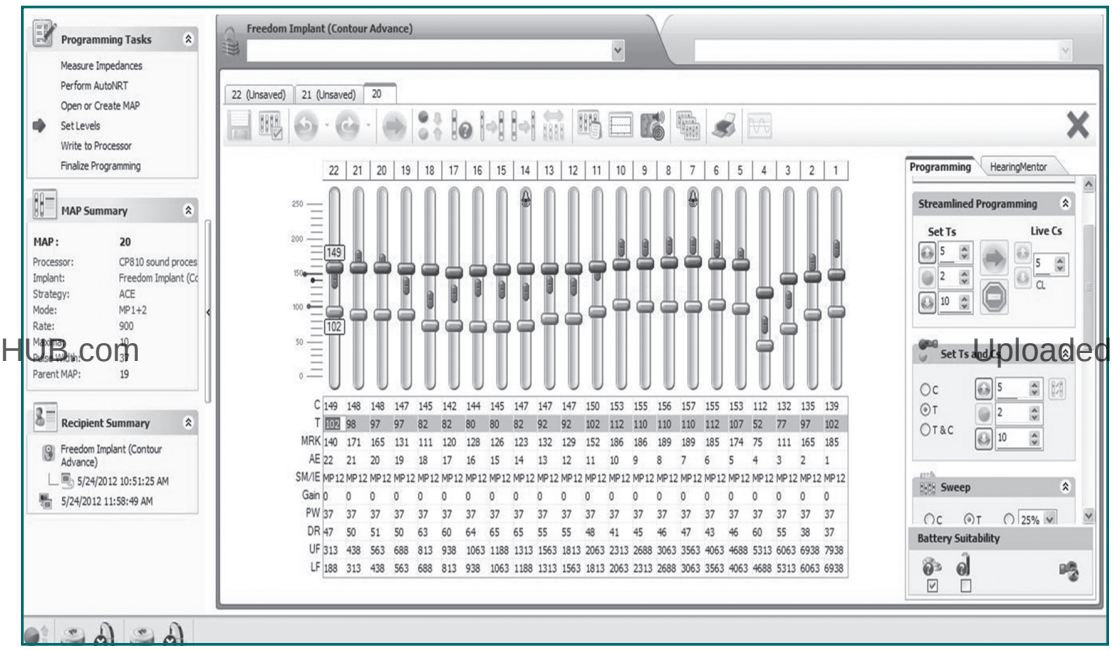

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**FIGURE 11–21.** Case 8: T levels set to approximate the NRT profile and C levels set within 10 to 12 clinical units from the T levels. *Note.* Provided courtesy of Cochlear™ Ltd.

40 clinical units, as most Nucleus users possess electrical dynamic ranges between 25 to 60 clinical units; although, notable exceptions for excellent performers do exist. Most importantly, the clinician ensured that the child showed no signs of discomfort as stimulation levels were increased. Additionally, as C levels were increased, the overall C-level profile was flattened relative to the T-level profile. Typically, the upper-stimulation level profile possesses less variation across the array relative to the T-level profile (Botros, Banna, & Maruthurkkara, 2013).

Eventually, a program with an electrical dynamic range varying between 30 to 50 clinical units across the electrode array was created (Figure 11–22). The clinician ensured that this program was comfortable for the child. Also, the clinician counseled the family to decrease the volume control setting to the minimum position before placing the device on the child's head, particularly if the child has gone several hours without using the device, as might be the case after awakening first thing in the morning. The family placed the device on the child's head with the volume control at the minimum setting and gradually increased the volume control setting over the course of a few seconds. The child showed no signs of intolerance of or discomfort with the implant with the processor set to the maximum volume control setting. Over time, the child gradually acclimated to the point at which there was no problem with starting with the maximum volume control setting.

The family returned 1 month after the initial appointment and further changes were made to C levels. As shown in Figure 11–22, the C-level profile was relatively flat in comparison to the T-level profile. Also, most electrodes possessed an electrical dynamic range between 40 to 60 clinical units. The family reported that the child consistently responded to low-level speech and environmental sounds while using the implant. They also noted that he wore the sound processor at the recommended volume control setting without any resistance. An informal assessment in the clinic revealed that he would turn toward all six Ling sounds when presented from 15 feet away. The child's auditory-verbal therapist began practicing on a conditioned play response in therapy, so the T-level measurements could be refined. Due to the improvement in use of the right cochlear implant, the child was scheduled to receive a cochlear implant for the left ear.




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**FIGURE 11–22.** Case 8: Program with an electrical dynamic range varying between 30 to 50 clinical units across the electrode array. *Note.* Provided courtesy of Cochlear™ Ltd.

## Take-Home Points

1. The clinician should suspect a problem with the CI program's stimulation levels, the external sound processor or cable/coil, or the cochlear implant when a child persistently refuses to use his or her cochlear implant.
2. Objective measures, such the eCAP, may be used in conjunction with behavioral responses to speech and environmental sounds to create an adequate program when a child will not provide valid and reliable behavioral responses to programming stimuli.
3. The volume control range of a pediatric recipient's sound processor should be set to possess a wide range of flexibility on the minimum side of the array, so that the child's caregivers may reduce stimulation as needed. Likewise, the authors typically limit the maximum point of the volume control range to a level we consider to be ideal, so the volume control will not be inadvertently set to the level that may be uncomfortable for the child. If we believe that further increases may be beneficial, then we will reschedule the child to return to the clinic for additional programming so stimulation levels may be increased under the supervision of the clinician.
4. Programming audiologists should work with the auditory-verbal therapists of young children to ensure that the programs are providing good access to sounds across the speech frequency range. auditory-verbal therapists may also assist with developing conditioned responses in therapy sessions with the goal that they may be carried over to programming sessions.

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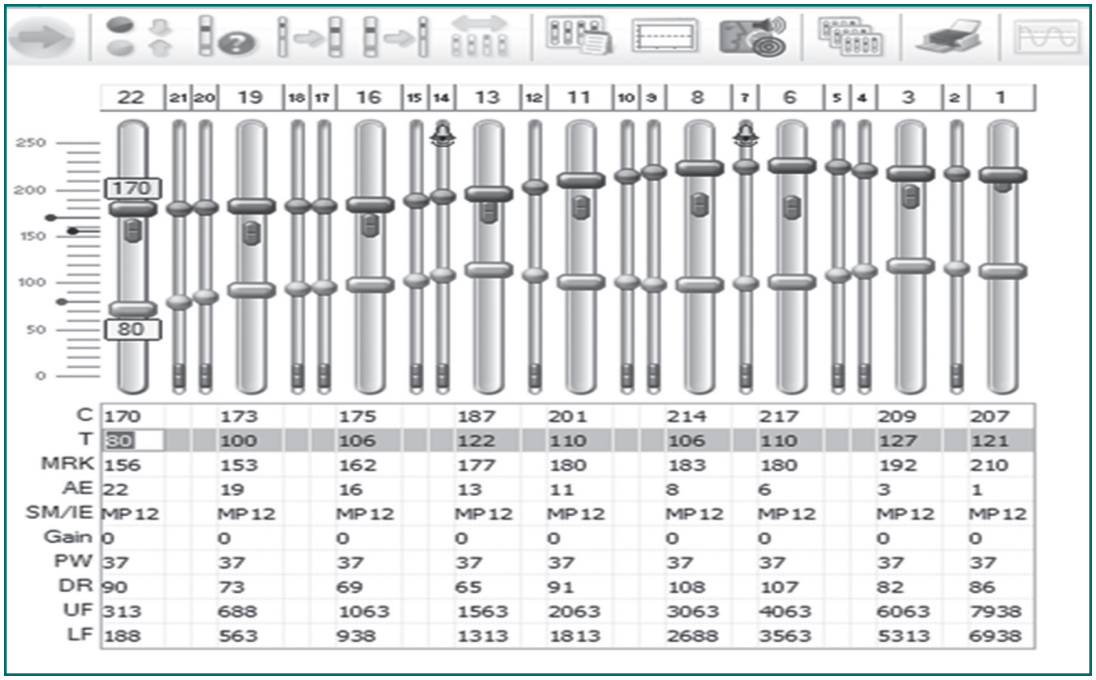
## CASE STUDY 9: ADDRESSING SPEECH-RECOGNITION DIFFICULTIES THAT MAY BE ASSOCIATED WITH EXCESSIVE STIMULATION

This 6-year-old girl received a Nucleus Freedom cochlear implant with Contour Advance internal device for her right ear when she was 1 year old and for her left ear when she was 16 months old. She was diagnosed during the first month of her life with bilateral profound sensorineural hearing loss of unknown etiology. She participated in auditory-verbal therapy, but her family often had difficulty with consistent attendance to therapy and audiology appointments. However, she received good benefit from her cochlear implants as evidenced by speech and spoken-language abilities that were consistent with typical hearing children of the same age.

She began the first grade when she was 6 years old. Her auditory-verbal therapist expressed concern regarding an increase in articulation errors and her difficulty understanding speech. Her last audiologic assessment and programming occurred 6 weeks prior to the onset of her difficulties.

On the day of her troubleshooting session, she scored 64% correct on the Lexical Neighborhood Test with use of the right implant alone. On previous occasions, she usually scored between 80 to 92% correct on the same test. Electrode impedances were within normal limits and were stable over the past several years. All of her external equipment appeared to be functioning normally. An analysis of her programs suggested that her stimulation levels may have been too high. Her program for her right ear is shown in Figure 11–23. As shown, her electrical dynamic range on most channels approached 90 clinical units, whereas some channels possessed an electrical dynamic range near 110 clinical units. The typical electrical dynamic range of a recipient of a Cochlear Nucleus Freedom (CI24RE), CI24, and CI5112 implant is between 25 to 60 clinical units. Additionally, her C levels on some of the more basal channels exceeded her eCAP threshold by 30 to almost 50 clinical units. Although the C levels of some high-performing recipients are significantly higher than their eCAP thresholds, studies suggest that the typical difference is ± 20 clinical units (Botros et al., 2013).

A review of the child's programs over time revealed that her upper-stimulation levels were slowly increased over the course of several years. More specifically, her C levels were typically increased by about 5 to 10 clinical units at each of her programming sessions. As a result, the C levels and electrical dynamic range were substantially




| | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| C | 170 | 173 | 175 | 187 | 201 | 214 | 217 | 209 | 207 |
| T | 80 | 100 | 106 | 122 | 110 | 106 | 110 | 127 | 121 |
| MRK | 156 | 153 | 162 | 177 | 180 | 183 | 180 | 192 | 210 |
| AE | 22 | 19 | 16 | 13 | 11 | 8 | 6 | 3 | 1 |
| SM/IE | MP12 | MP12 | MP12 | MP12 | MP12 | MP12 | MP12 | MP12 | MP12 |
| Gain | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 |
| PW | 37 | 37 | 37 | 37 | 37 | 37 | 37 | 37 | 37 |
| DR | 90 | 73 | 69 | 65 | 91 | 108 | 107 | 82 | 86 |
| UF | 313 | 688 | 1063 | 1563 | 2063 | 3063 | 4063 | 6063 | 7938 |
| LF | 188 | 563 | 938 | 1313 | 1813 | 2688 | 3563 | 5313 | 6938 |

**FIGURE 11–23.** Case 9: Program for her right ear. *Note.* Provided courtesy of Cochlear™ Ltd.

larger than most high-performing recipients. Indeed, at her last programming session 6 weeks earlier, her C levels were increased by 6 to 10 clinical units on most electrodes. It should be noted that she provided reliable T-level responses via VRA and CPA over the past several years, and her T levels remained largely unchanged. Zwolan and Overstreet (2005) have previously shown that young children's upper-stimulation levels have a tendency to increase gradually over the first several years of their implant use. They attributed this increase to the clinician's inclination to increase upper-stimulation levels by a small amount at each session in order to make the session seem like it was productive. This is more likely to occur with young children because they are often unable to provide reliable feedback pertaining to loudness judgments. In contrast, postlingually deafened adult recipients have a vast experience and understanding of loudness judgments, and as a result, they were very capable of informing clinicians of when excessive upper-stimulation levels resulted in discomfort or in a degradation of sound quality or speech clarity.

The clinician attempted to obtain eSRT measurements, but the response could not be obtained with the admittance probe in either ear. We also attempted to conduct loudness-balancing measures, but the recipient did not fully understand the task. T levels were measured and were similar to previous measures. The C-level profile was flattened or smoothed out, and C levels were adjusted globally to levels that resulted in an electrical dynamic range that was more similar to the typical high-performing recipient (Figure 11–24). Also, her NRT responses typically fell within her electrical dynamic range after her C levels were revised. She did complain that the new program resulted in all speech and environmental sounds being "too soft." She was asked to use the new program to see if she could adjust to the lower loudness it provided. Of note, eCAP (i.e., NRT) were measured in the AutoNRT module and were found to be significantly higher than when last measured. Use of excessive stimulation levels may cause an increase in eCAP thresholds over time, presumably due to neural fatigue and/or neural damage.

She was seen for assessment 4 weeks after the changes were made. Her auditory-verbal therapist noted a substantial improvement in her articulation and in her speech recognition. Also, she

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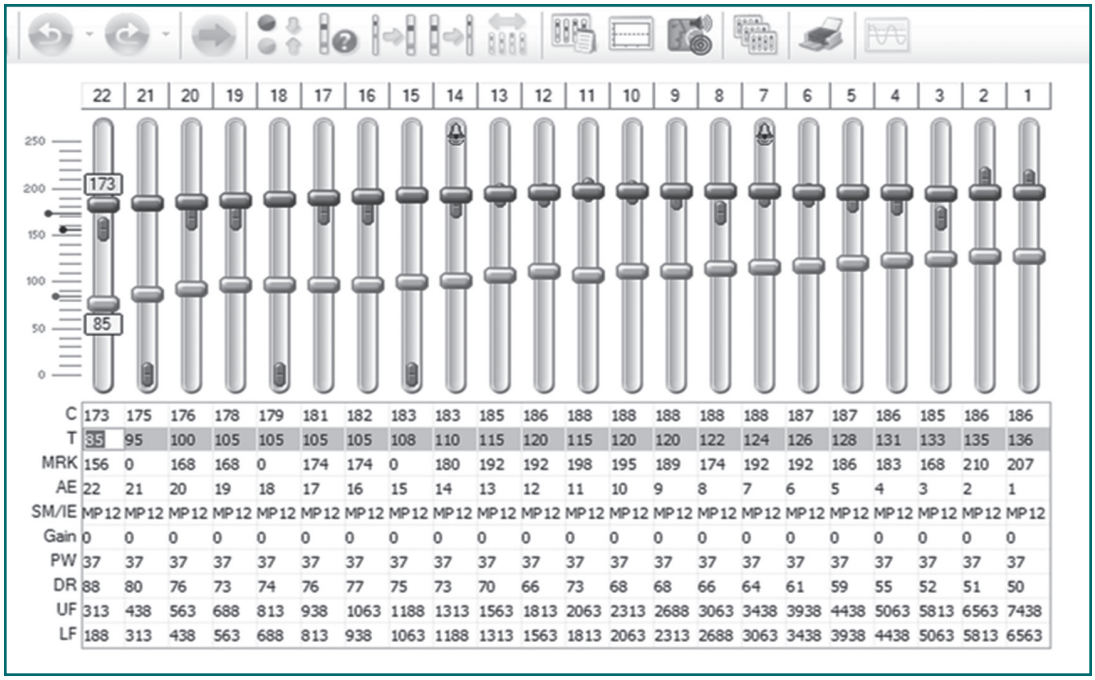



**FIGURE 11–24.** Case 9: C-level profile flattened and C levels adjusted globally to make electrical dynamic range similar to the typical recipient. *Note.* Provided courtesy of Cochlear™ Ltd.

scored 92% correct on the Lexical Neighborhood Test with use of the right implant alone. Furthermore, the recipient reported that the loudness of the signal she received as satisfactory. Finally, her eCAP was remeasured in the NRT module, and her NRT thresholds actually decreased from her responses measured at the previous session. This finding further suggests that the stimulation she was receiving from her previous program was excessive and possibly leading to neural fatigue. Finally, the child's auditory-verbal therapist began working on establishing the concepts of loudness judgments and the difference between pitch and loudness. She also began to practice psychophysical loudness scaling and loudness balancing in the child's therapy sessions with the goal being for the child to successfully complete these tasks in programming sessions.

## Take-Home Points

1. Optimizing upper-stimulation levels is a critically important step in the provision of a comfortable signal that provides good speech understanding and sound quality. One of the most difficult components of creating an optimal program for young children is the determination of the most appropriate upper-stimulation levels.
2. Clinicians should use a variety of objective measures to determine where upper-stimulation levels should be set for recipients who cannot provide feedback about the signal they experience, but the eSRT is the ideal objective measure to estimate appropriate upper-stimulation levels.
3. When eSRTs cannot be obtained, the clinician should rely on a battery of behavioral and objective measures and information to guide in the determination of upper-stimulation levels. It is helpful to have an understanding of what constitutes a typical electrical dynamic range and upper-stimulation levels for high-performing recipients of a given implant model and electrode array. Of course, some high-performing recipients will possess atypical electrical dynamic ranges and upper-stimulation levels. With that said, the clinician should always be suspicious of electrical

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dynamic ranges or upper-stimulation levels that are substantially different from the normal range.

4. Clinicians should also judge the appropriateness of a recipient's program on the basis of the user's hearing performance. For young children, it is imperative to receive feedback from the child's family and auditory-verbal therapist to ensure that satisfactory progress is being achieved. Also, the audiologist should regularly administer speech-recognition assessments to ensure good open-set speech understanding for a linguistically appropriate set of speech materials.
5. The clinician should partner with the child's family and auditory-verbal therapist to practice the concepts of loudness normalization and loudness balancing. Once a child can master those tasks, it becomes much easier to create an excellent program, even when objective measures, such as the eSRT, cannot be obtained to determine upper-stimulation levels.

## CASE STUDY 10: UTILIZING BEHAVIORAL INFORMATION TO CREATE A COCHLEAR IMPLANT PROGRAM AND ADDRESS RECIPIENT DIFFICULTIES

A 9-year-old girl was referred for audiologic assessment and programming after she experienced a decrease in performance following another programming session. This child was diagnosed with severe to profound hearing sensorineural hearing loss shortly after birth, and she received a cochlear implant for her right ear at her first birthday. She continued hearing aid use for the left ear, but received a cochlear implant for that ear when she was 5 years old. She willingly used both cochlear implants during all waking hours, although she did favor her right ear implant, presumably because she received that implant at a much earlier age than for the left ear. When she was 8 years old, she scored 92% correct on CNC monosyllabic word recognition with use of the right cochlear implant and 68% correct with use of the left cochlear implant.

About 1 year later (at 9 years of age), she was seen for audiologic assessment and cochlear implant programming. Her mother was uncertain of the methods used to determine programming changes that were made to the right ear implant, but she did report that the child was reluctant to use the device following that programming session. Additionally, the mother reported that the child consistently complained that loud sounds were uncomfortable and that she heard a constant noise from the implant. Furthermore, she noted that the child seemed to be having considerable difficulty understanding speech, particularly in the presence of competing noise. CNC word recognition assessment indicated a score of 64% correct for the right ear. No changes were made to the left ear implant, and she continued to willingly use that device without problems.

The child's current program for the right ear was compared to the program she was using satisfactorily prior to her previous programming session (see Figure 11–25 for current program causing difficulty). It was determined that M levels had increased by approximately 130 charge units across the electrode array. Also, T levels were increased across the array between the two sessions, and the T levels were quite erratic across the array. Additionally, the automatic gain control (AGC) input processing had been disabled.

At the present appointment, numerous programming changes were made to address the recipient's difficulties. First, T levels were measured using CPA. Although the child was definitely capable of completing T-level measurements via a conventional audiometric approach, many children at this age are more willing to provide threshold level responses and maintain attention throughout the programming session when measures are completed via the use of engaging conditioned-play games. Based on the child's responses, T levels were decreased significantly across the electrode array.

Second, we attempted eSRT measures but were unable to successfully obtain to response. As a result, the child was allowed to operate the programming keyboard to globally adjust her M levels in live speech mode to her most preferred listening level. Then, loudness balancing was completed across the electrode array. After loudness

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balancing, the child was once again asked to globally adjust her M levels to her preferred listening level. The new cochlear implant program established with her behavioral responses is shown in Figure 11–26.

Next, ClearVoice input processing was enabled, and the child once again adjusted her M levels globally to her preferred listening level while listening to speech and a low-level noise (50 dBA). The microphone mixing of her processor was also

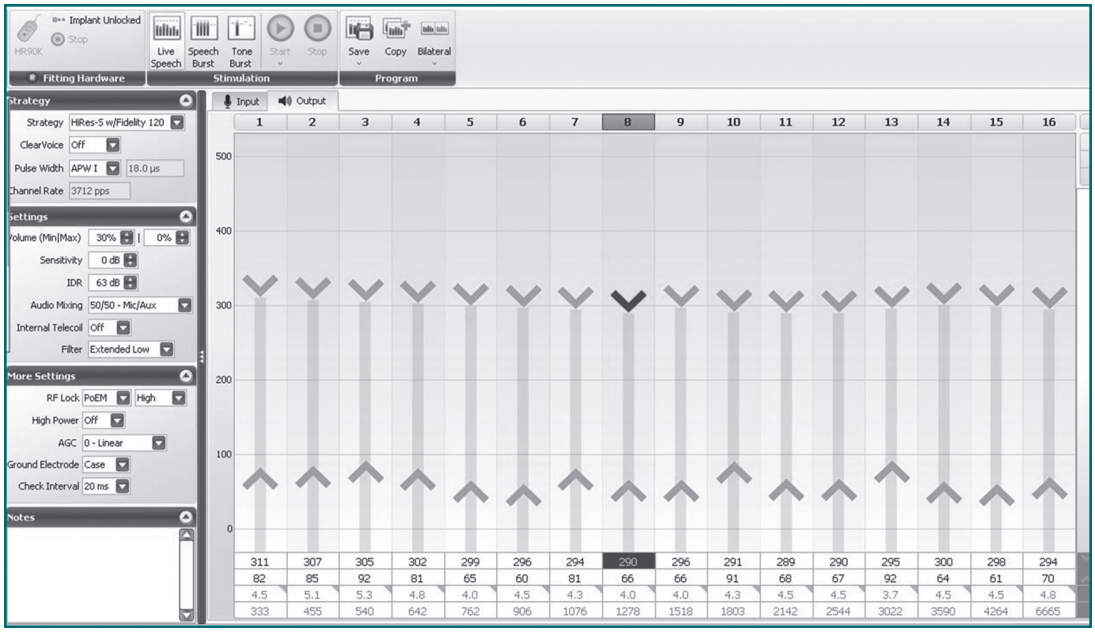
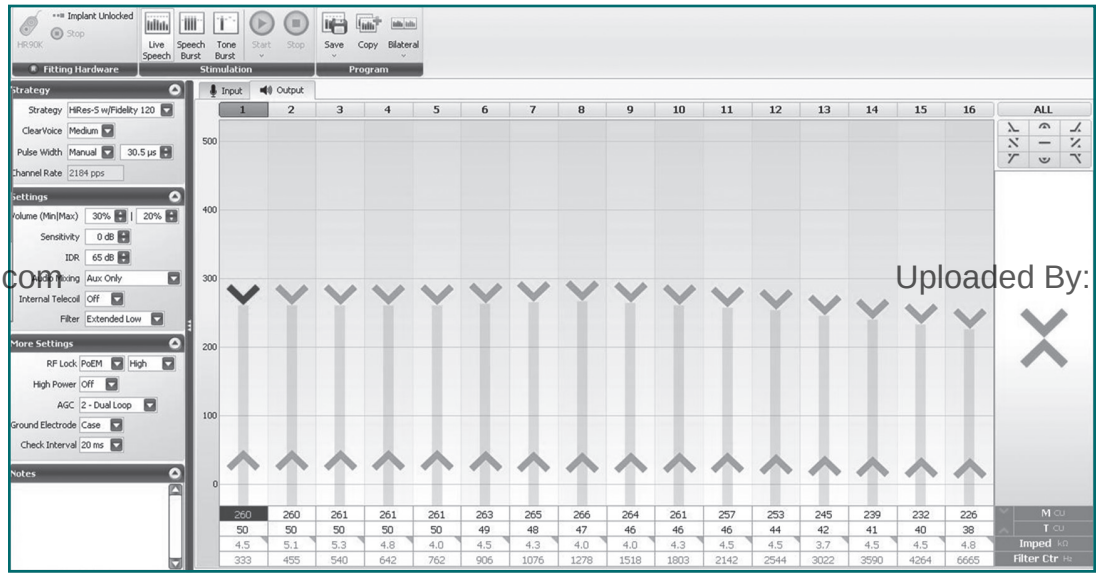



**FIGURE 11–25.** Case 10: Child's current program. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–26.** Case 10: New cochlear implant program established with her behavioral responses. *Note.* Provided courtesy of Cochlear™ Ltd.

set so that the T-Mic was the only microphone that was active (i.e., the on-board processor microphone was disabled). Once again, we ensured that the recipient was satisfied with the loudness and sound quality of the signal she received. Finally, the dual-action AGC input processing was enabled (i.e., AGC-2).

The child was satisfied immediately with the new program that was created for her right ear. Her mother reported that she resumed bilateral cochlear implant use during all waking hours. She returned to the clinic within 2 months of the sound processor adjustments and scored 92% correct with the right implant alone.

### Take-Home Points

1. Excessive T levels will elicit a constant buzzing, static, frying, or hissing sound. This undesirable percept will typically disappear once T levels are decreased.
2. Upper-stimulation levels have a large effect on sound quality, speech recognition, and the user's ability to monitor his or her voice.
3. The clinician should attempt to use a battery of measures to determine the optimal upper-stimulation levels a child should use. However, when objective measures, such as the eSRT, cannot be effectively obtained in older children, then the clinician must strive to create an optimal program using behavioral measures.
4. Stimulation levels were set appropriately based on the patient's subjective responses. Overall, her T levels and C levels decreased considerably from her previous program.
5. The recipient's microphone mode was set to "T-Mic only," which has been shown to improve speech recognition in noise when compared to the processor microphone. Also, AGC-2 input processing was enabled in an attempt address the recipient's goal of difficulty listening in noisy conditions.
6. ClearVoice and AGC-2 input processing were also activated to improve speech recognition and sound quality in noise. The clinicians selected these strategies because previous research has suggested that these technologies may be beneficial in noise.

## ELECTRODE IMPEDANCE CASE STUDIES

The following four case studies illustrate a variety of examples in which clinicians altered management strategies on the basis of electrode impedance results. The reader should refer to Chapter 2 for a description of electrode impedance measurement and for a fundamental description of normal and abnormal electrode impedance results.

## CASE STUDY 11: SAWTOOTH ELECTRODE IMPEDANCE PATTERN MANAGED WITH PROGRAMMING

This 10-year-old boy passed his newborn hearing screening but was diagnosed with bilateral profound sensorineural hearing loss at 15 months of age. He received a Cochlear Nucleus CI24R cochlear implant for his right ear when he was 18 months old and a Cochlear Nucleus CI24R implant for his left ear when he was 3 years old. He excelled with his cochlear implants as evidenced by age-appropriate scores on standardized assessments of speech and language as well as good scores on word-recognition assessment. When he was 9 years old, he scored 84% correct with use of the right cochlear implant and 80% correct with use of the left cochlear implant on CNC monosyllabic word recognition assessment.

When he was 10 years old, his parents reported that he was experiencing more difficulty understanding speech, particularly in the presence of competing noise. Speech recognition assessment indicated a significant decrease in word recognition for the right ear. Specifically, he scored 60% correct with use of the right implant and 84% with use of the left implant on CNC word recognition assessment. Electrode impedance assessment indicated a sawtooth (zig-zag) pattern associated with the right implant (Figure 11–27) with many of the even-numbered electrode contacts.

His audiologist disabled all even-numbered electrodes between contacts 8 through 20, and the child immediately reported better sound quality. The new cochlear implant program is shown in

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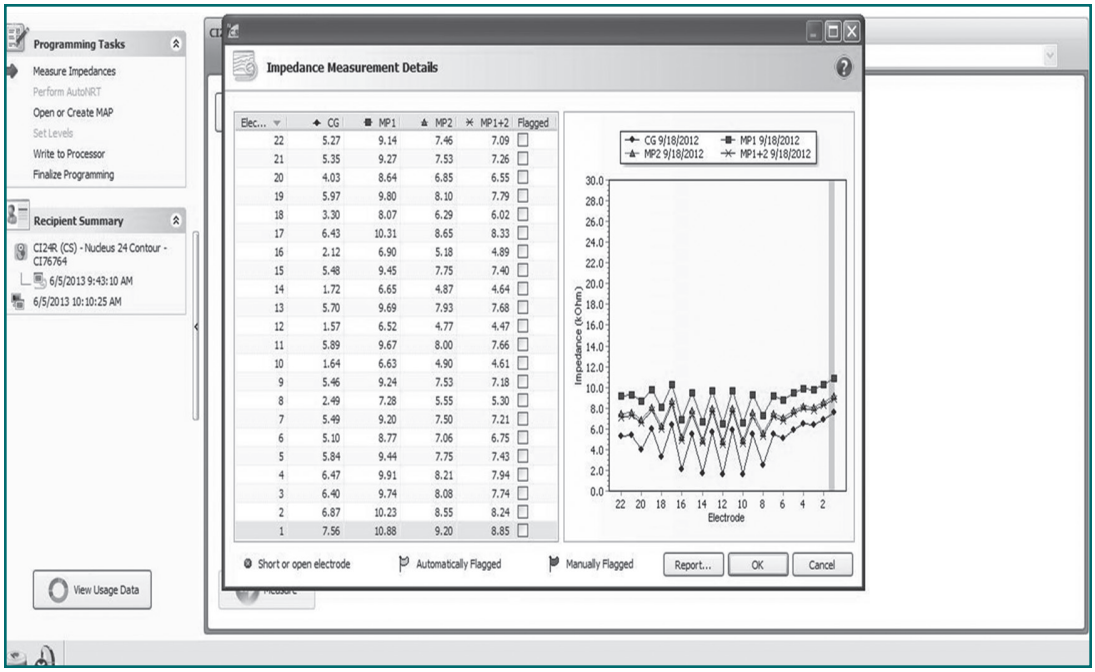
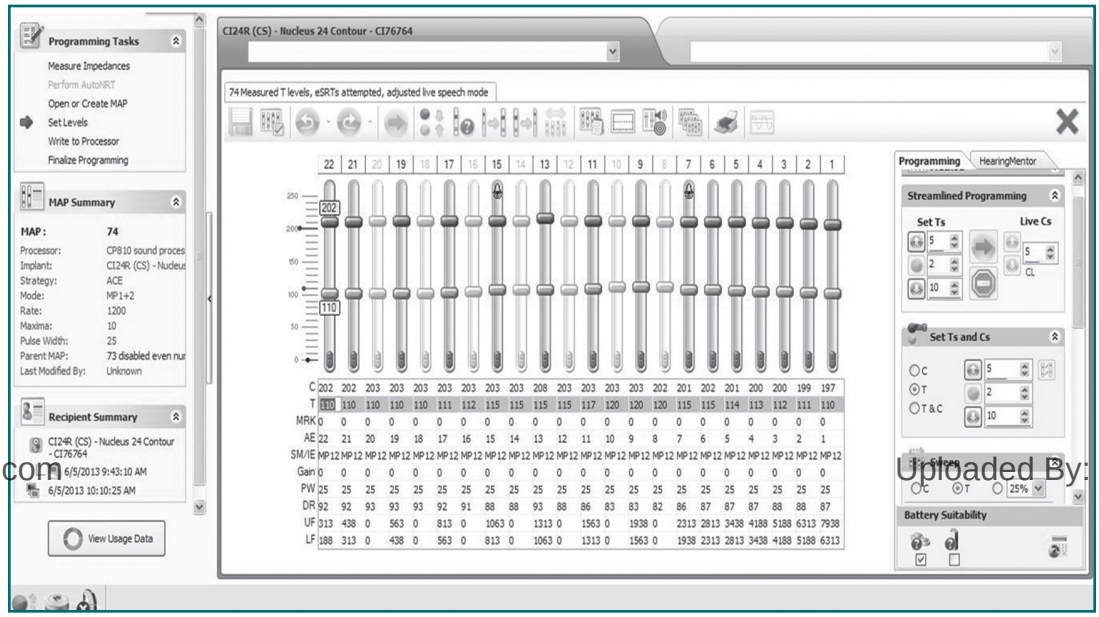



**FIGURE 11–27.** Case 11: Electrode impedance results indicate a sawtooth (zig-zag) pattern in the right implant. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–28.** Case 11: New cochlear implant program with several even-numbered electrodes disabled. *Note.* Provided courtesy of Cochlear™ Ltd.

Figure 11–28. Word-recognition assessment (CNC words) conducted 1 month later showed a substantial improvement in performance (80% correct with use of the right implant alone). The child is seen for quarterly checkups to ensure satisfactory performance and implant function.

## Take-Home Points

1. An atypical sawtooth pattern in Nucleus implants may be attributed to a tear in the silicone insulation on one side of the implant. This fault may cause partial shorting on even-numbered or odd-numbered electrode contacts.
2. If this problem arises, one management strategy is to disable electrodes that are within the sawtooth configuration and contain lower impedances than their neighbors. In this particular case, this strategy significantly improved the recipient's performance. It should be noted that this type of improvement does not always occur in cases with similar sawtooth impedance findings. Some recipients do not report satisfactory sound quality and/or do not experience an improvement in speech recognition when affected electrodes are disabled. Implant replacement may be considered for these recipients.

## CASE STUDY 12: FLAT LOW COMMON GROUND IMPEDANCES WITH PERFORMANCE DECREASE

This 5-year-old male was diagnosed with hearing loss at birth and underwent simultaneous bilateral cochlear implant surgery (Nucleus Freedom with Contour Advance) at 11 months of age. By all accounts, he was a very successful implant user. His scores on standardized speech and language assessments exceeded the norm for normal-hearing children his age. At four-and-one-half years of age, he scored 92% correct, 88% correct, and 96% correct on Lexical Neighborhood Test (LNT) monosyllabic word recognition testing in the right ear, left ear, and bilateral conditions, respectively.

However, shortly after an audiologic assessment at age 5 years, the child's parents reported that he was experiencing difficulty understanding speech. They noted that his teachers also had expressed concern that he was struggling to hear in the classroom. Audiologic assessment indicated a significant deterioration in word recognition for the right ear. At that session, he scored 96% correct on the LNT word recognition test with use of the left implant alone but only 52% with use of the right implant. All of his external equipment was replaced, but his difficulties persisted. His eCAP (NRT) and eSRT responses were present at levels consistent with previous assessments. Also, his T levels and C levels remained largely unchanged.

A retrospective review of his electrode impedances revealed a potential issue. Electrode impedances appeared to be unremarkable on the second day of his initial activation (Figure 11–29) and again at the 1-week and 1-month postactivation appointments (1 month shown in Figure 11–30). In contrast, electrode impedances measured at a session conducted 2 months after implant activation showed evidence of partial shorting on some odd electrodes (electrodes 9, 11, and 15) consistent with a possible hermeticity breach of the silicone insulation of the implant (Figure 11–31). Additionally, electrode impedance assessment conducted at 4 months postactivation was further indicative of partial electrode shorting with electrodes 5, 9, 11, 15, and 21 all possessing significantly lower electrode impedance than their even-numbered electrode contacts (Figure 11–32). The clinician managing the patient at the time did not consider the electrode impedances to be atypical.

At the next programming session (7 months postactivation), the morphology of the electrode impedance graph was once again unremarkable (Figure 11–33). However, a closer inspection of the common ground electrode impedance values revealed that they were slightly lower than the common ground impedance values obtained at the 1-month postactivation assessment. As shown in Figures 11–34, 11–35, and 11–36, the common ground electrode impedance values measured at later programming sessions continued to decrease. Specifically, common ground electrode impedances ranged from 4 to 6 kOhms at the 1-year postactivation assessment but had decreased to a range from to 2 to 3 kOhms at the time at which his difficulties began to surface.

The implant team made to decision to replace the child's suspect implant. He immediately reported that his new implant provided a better sound quality compared to the previous implant. LNT word recognition assessment conducted at 1-month postactivation of the replacement implant revealed a score of 96% correct with use

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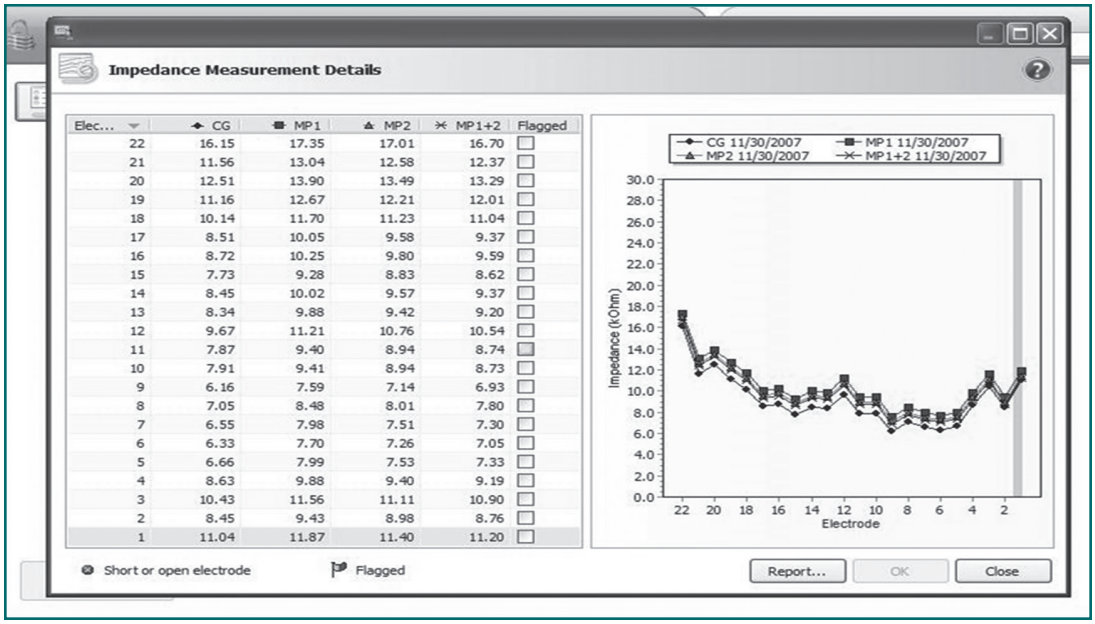
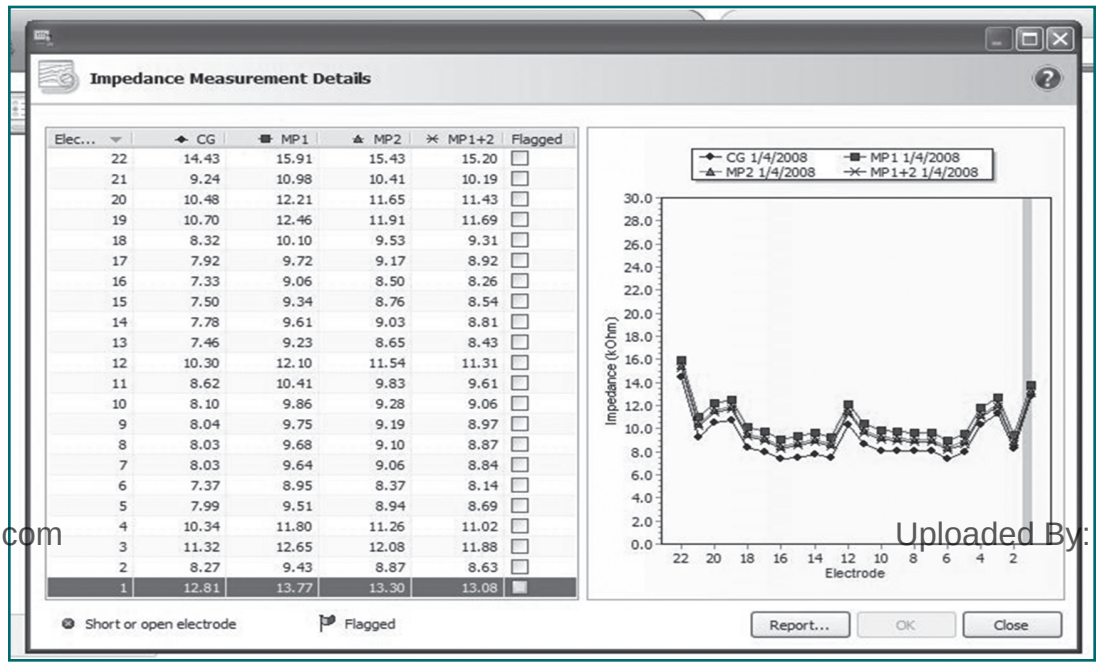



**FIGURE 11–29.** Case 12: Electrode impedances on the second day of his initial activation *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–30.** Case 12: Electrode impedances 1-month postactivation. *Note.* Provided courtesy of Cochlear™ Ltd.

of that implant alone. His parents and teachers also both reported that they observed an obvious improvement in his listening abilities at home, school, and social situations. Finally, a postrevision analysis of the suspect device confirmed a tear in the insulation of the implant case.

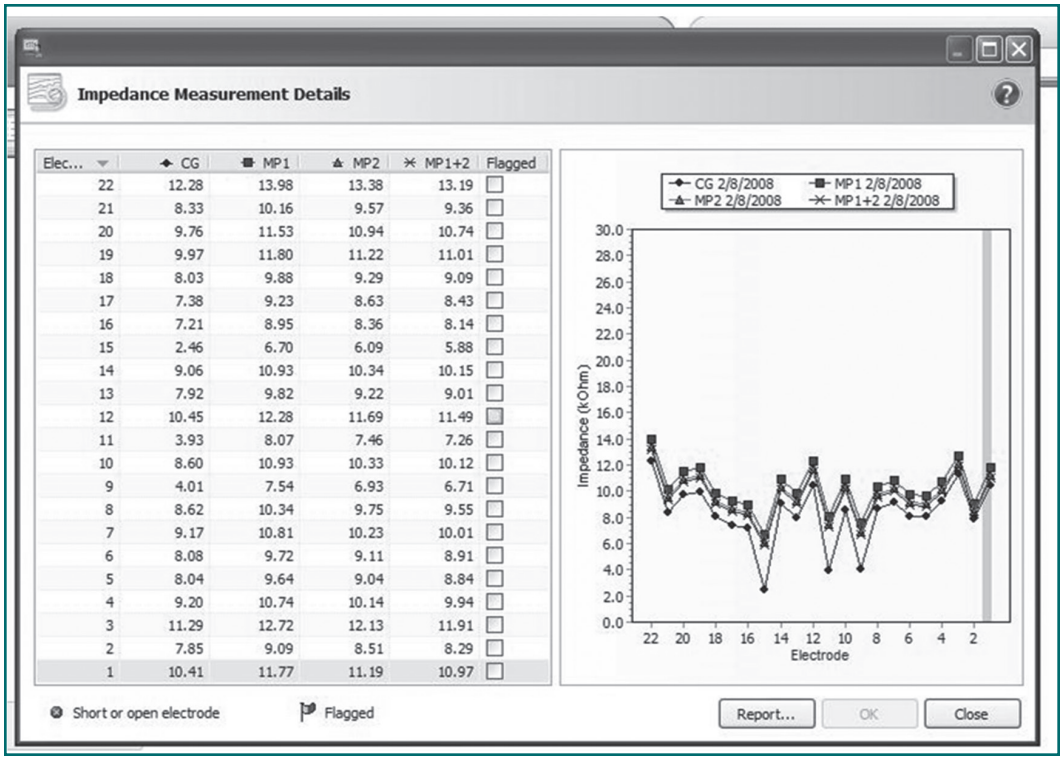
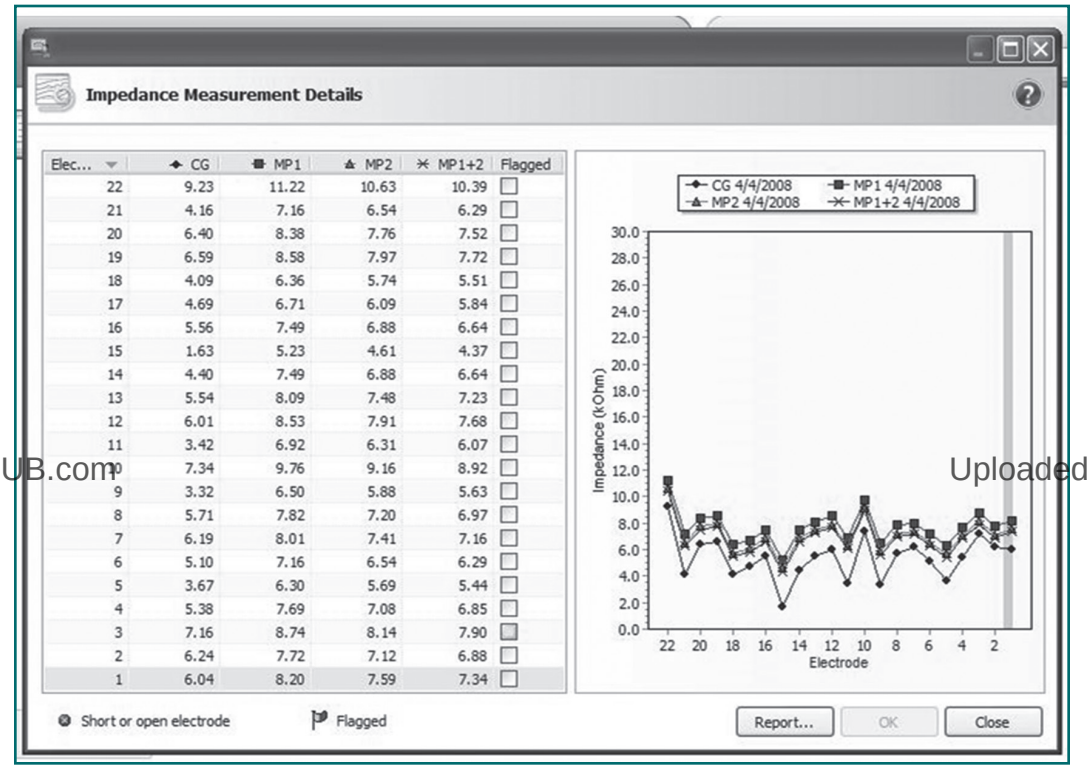



**FIGURE 11–31.** Case 12: Electrode impedances 2 months postactivation *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–32.** Case 12: Electrode impedances 4 months postactivation. *Note.* Provided courtesy of Cochlear™ Ltd.

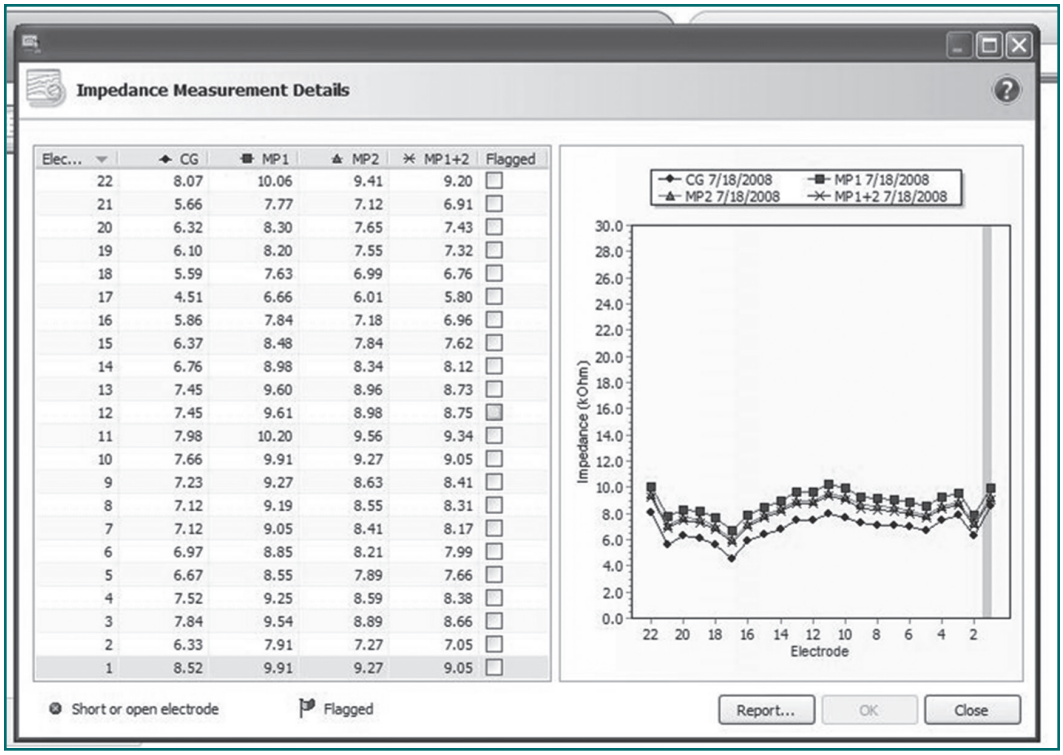
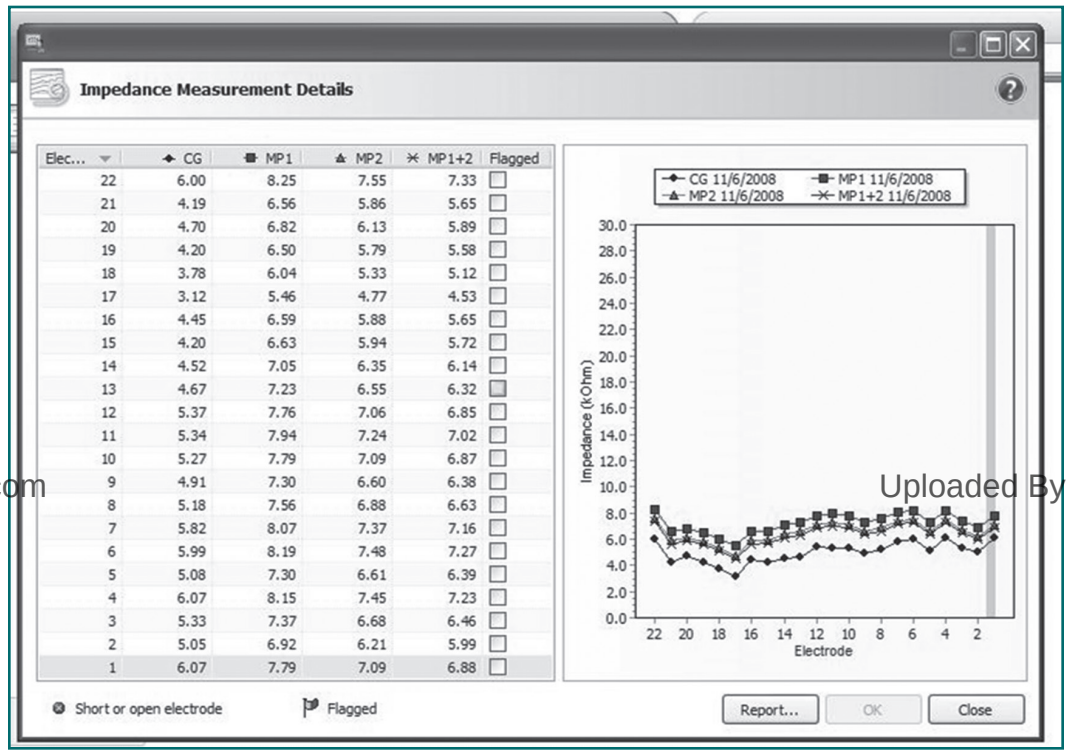



**FIGURE 11–33.** Case 12: Electrode impedances 7 months postactivation. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–34.** Case 12: Electrode impedance results measured at 11 months postactivation. Note that common ground electrode impedance values are lower than what was measured at the 7-month postactivation session shown in Figure 11–33. *Note.* Provided courtesy of Cochlear™ Ltd.

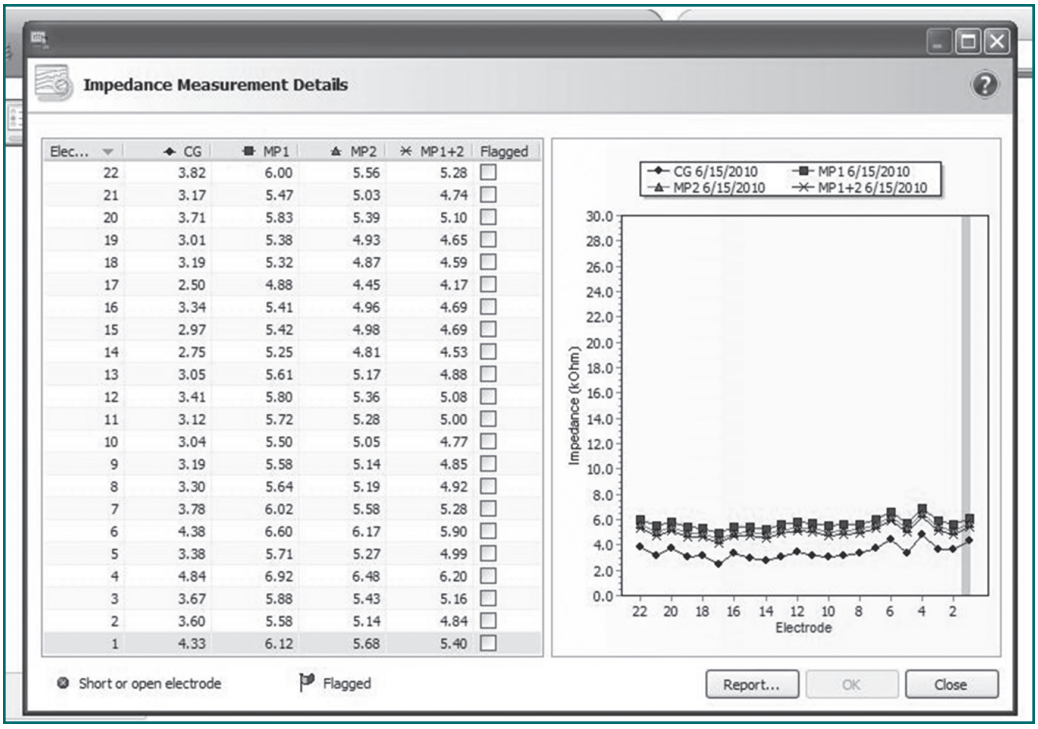
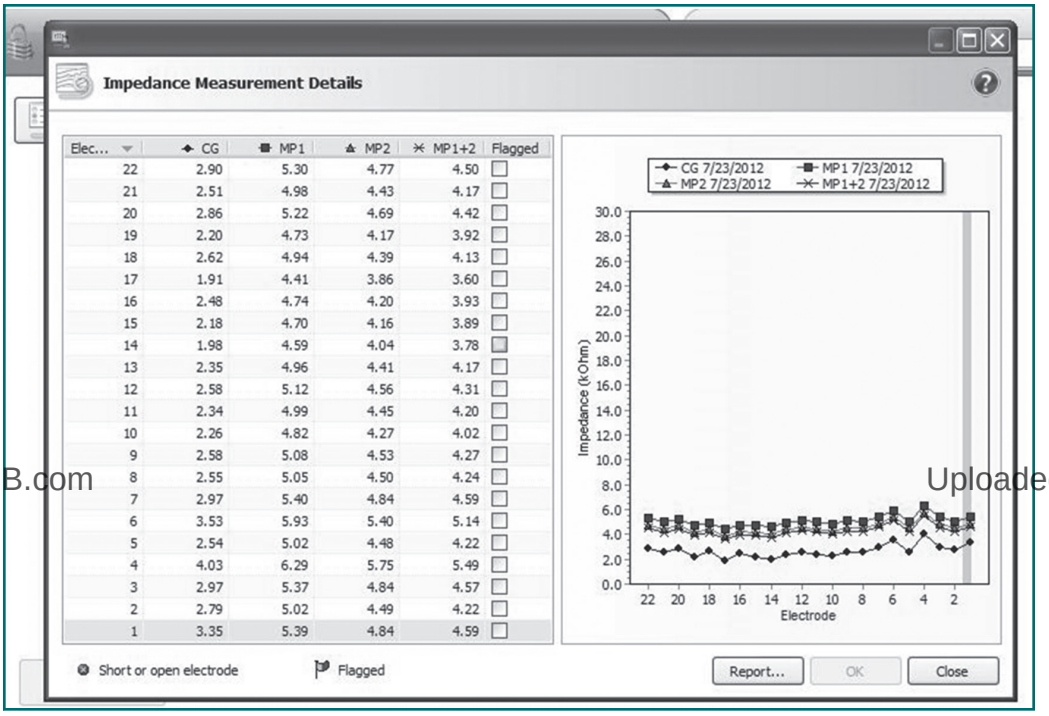



**FIGURE 11–35.** Case 12: Electrode impedance results measured at approximately two-and-one-half years postactivation. Note that common ground electrode impedance values continue to decrease relative to previous measurements. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–36.** Case 12 Electrode impedance results measured at approximately four-and-one-half years postactivation. Note that common ground electrode impedance values have continued to decrease from pervious measures and are unusually low and flat across the electrode array. *Note.* Provided courtesy of Cochlear™ Ltd.

## Take-Home Points

1. Electrode impedances generally decrease significantly throughout the first few weeks of implant use. However, after impedances stabilize, a continued and considerable decrease in the electrode impedance may be indicative of partial shorting secondary to body fluid ingress through a tear in the silicone insulation of the cochlear implant. Clinicians should monitor electrode impedances over time to identify unusual impedance changes.
2. It is important to periodically evaluate speech recognition using a challenging metric (i.e., monosyllabic words), so the effect of a change in implant function or integrity can be captured in the recipient's performance.

## CASE STUDY 13: PHYSIOLOGIC-RELATED ELECTRODE IMPEDANCE CHANGES

This adult male received a Cochlear Nucleus Freedom cochlear implant with Contour Advance electrode array when he was 75 years old. He had a 20-year progressive hearing loss (audiogram in Figure 11–37), and he noted that he was unable to

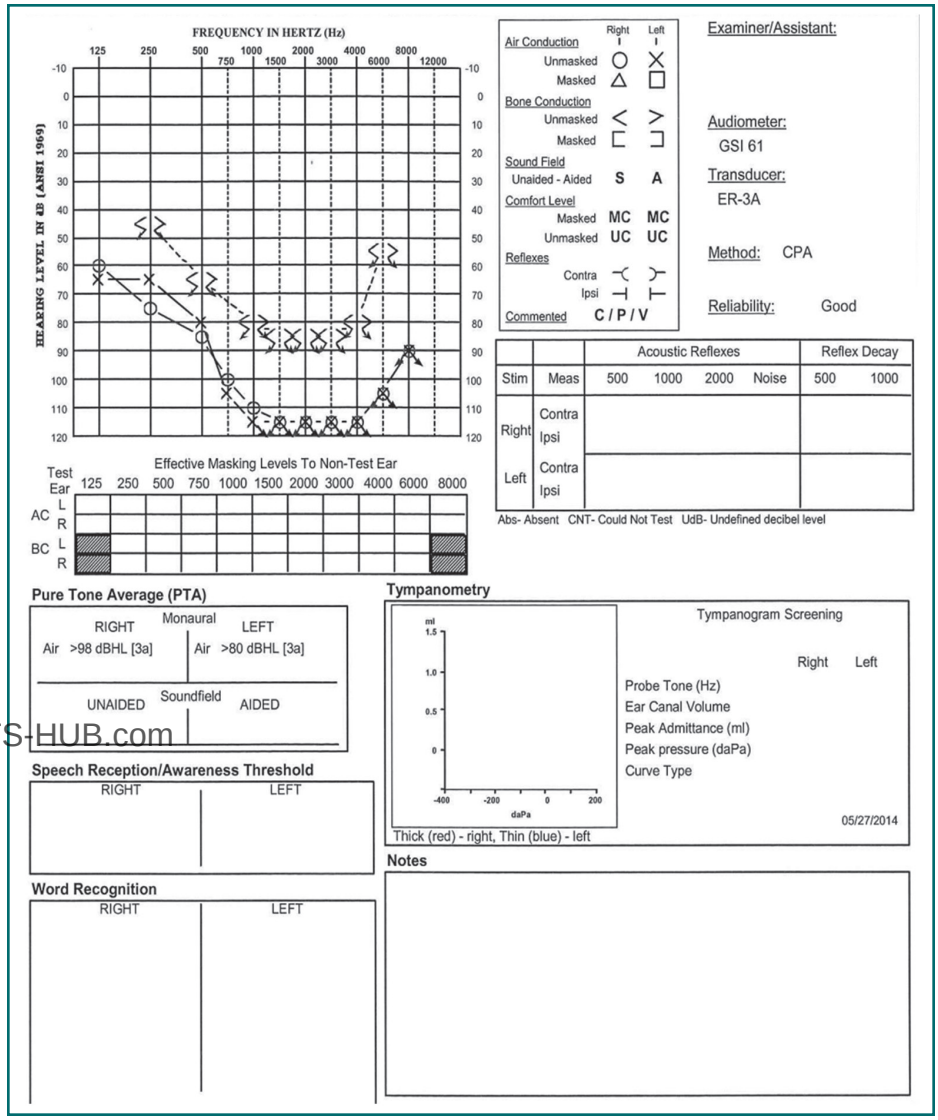
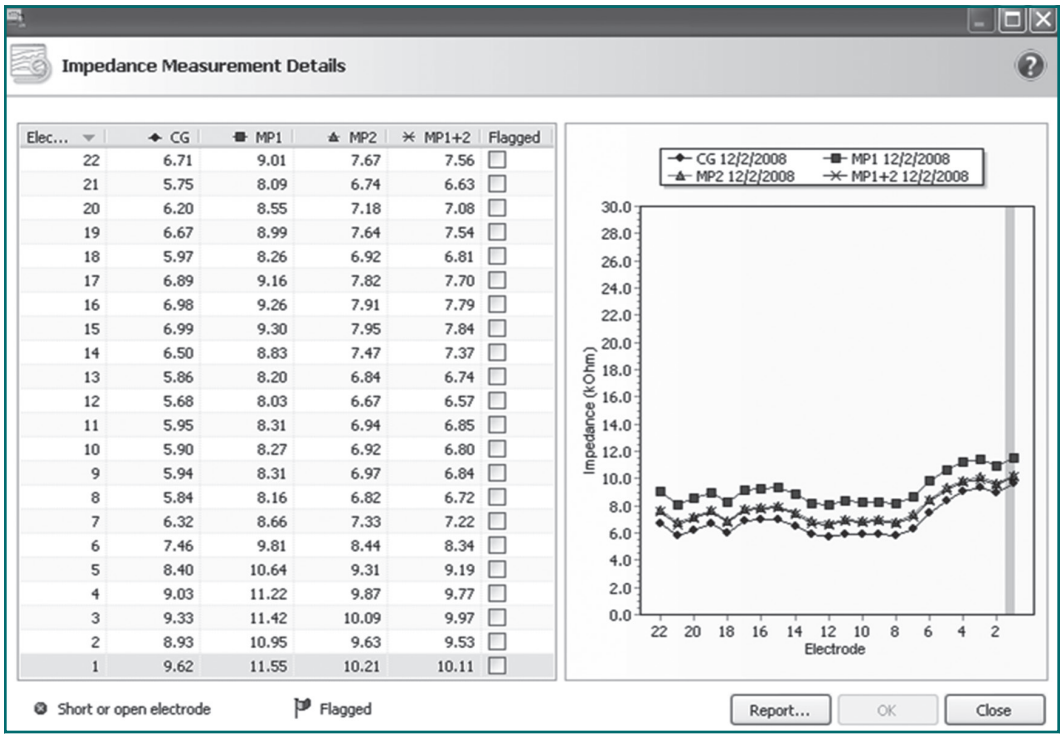



**FIGURE 11–37.** Case 13: Preimplant audiogram.

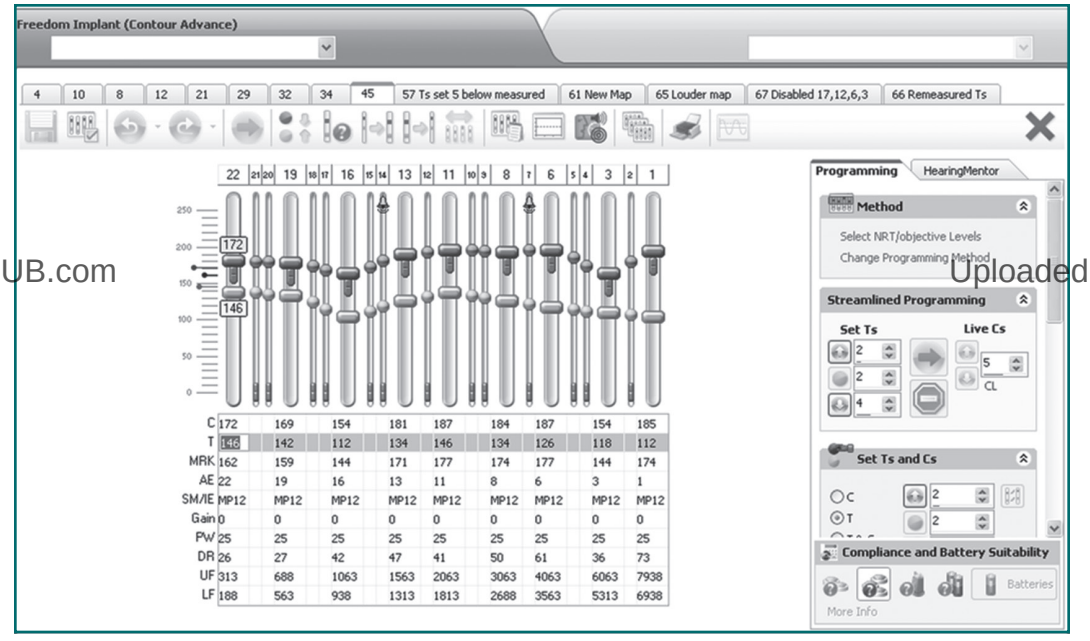
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use the telephone for 3 to 4 years prior to implantation. His cochlear implant surgery was unremarkable. At his 1-year postactivation appointment, he scored 99% correct on Hearing in Noise Test (HINT) sentences in quiet (60 dBA) and 88% on CNC word recognition assessment (60 dBA). Figures 11–38 and 11–39 show his cochlear implant electrode impedances and program, respectively.




| Elec... | CG | MP1 | MP2 | MP1+2 | Flagged |
|---|---|---|---|---|---|
| 22 | 6.71 | 9.01 | 7.67 | 7.56 | |
| 21 | 5.75 | 8.09 | 6.74 | 6.63 | |
| 20 | 6.20 | 8.55 | 7.18 | 7.08 | |
| 19 | 6.67 | 8.99 | 7.64 | 7.54 | |
| 18 | 5.97 | 8.26 | 6.92 | 6.81 | |
| 17 | 6.89 | 9.16 | 7.82 | 7.70 | |
| 16 | 6.98 | 9.26 | 7.91 | 7.79 | |
| 15 | 6.99 | 9.30 | 7.95 | 7.84 | |
| 14 | 6.50 | 8.83 | 7.47 | 7.37 | |
| 13 | 5.86 | 8.20 | 6.84 | 6.74 | |
| 12 | 5.68 | 8.03 | 6.67 | 6.57 | |
| 11 | 5.95 | 8.31 | 6.94 | 6.85 | |
| 10 | 5.90 | 8.27 | 6.92 | 6.80 | |
| 9 | 5.94 | 8.31 | 6.97 | 6.84 | |
| 8 | 5.84 | 8.16 | 6.82 | 6.72 | |
| 7 | 6.32 | 8.66 | 7.33 | 7.22 | |
| 6 | 7.46 | 9.81 | 8.44 | 8.34 | |
| 5 | 8.40 | 10.64 | 9.31 | 9.19 | |
| 4 | 9.03 | 11.22 | 9.87 | 9.77 | |
| 3 | 9.33 | 11.42 | 10.09 | 9.97 | |
| 2 | 8.93 | 10.95 | 9.63 | 9.53 | |
| 1 | 9.62 | 11.55 | 10.21 | 10.11 | |

**FIGURE 11–38.** Case 13: Electrode impedances. *Note.* Provided courtesy of Cochlear™ Ltd.

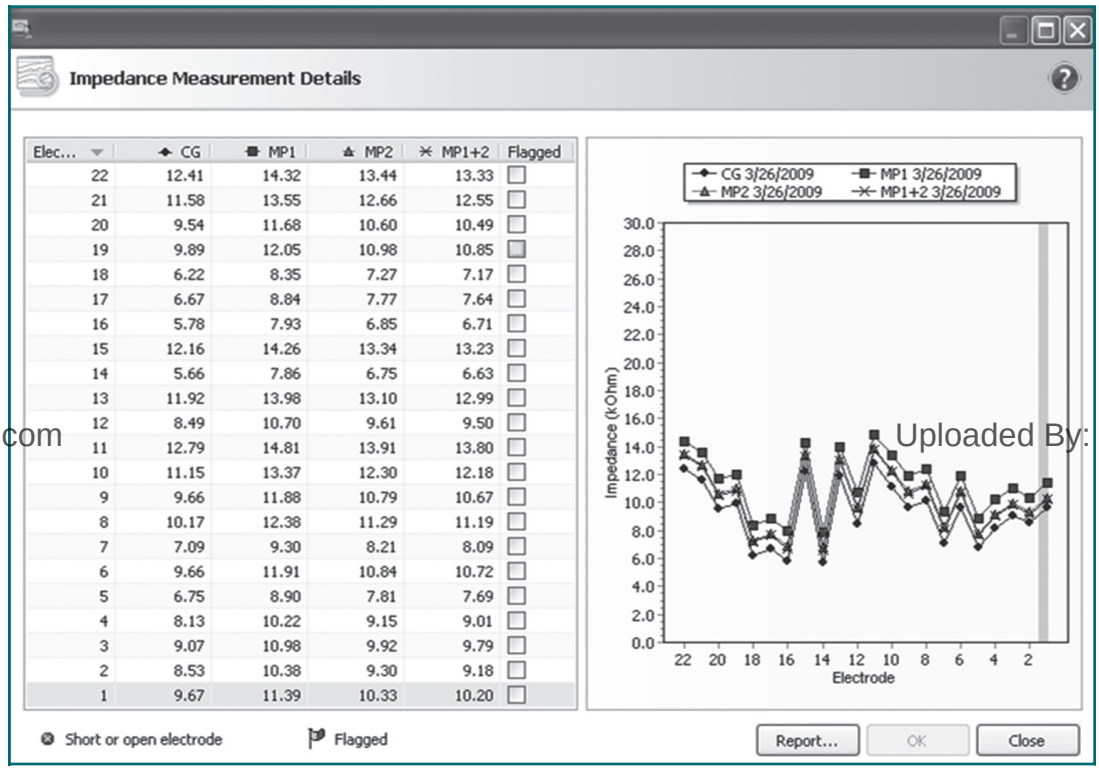



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**FIGURE 11–39.** Case 13: Cochlear implant program. *Note.* Provided courtesy of Cochlear™ Ltd.

At 23 months postactivation, he contacted the clinic to report a sudden deterioration in sound quality and speech recognition. He scored 63% correct on HINT sentence testing in quiet, and CNC word recognition was not completed because the patient became upset because it was too difficult. Electrode impedances had changed considerably (Figure 11–40), and his stimulation levels also substantially increased. He was seen several times for assessment and programming over the next several weeks. His electrode impedance findings continued to possess an unusual morphology (i.e., electrode impedances varied substantially across the array), and his performance continued to remain poor. Additionally, his eSRT and eCAP responses, which were formally present and robust across the electrode array, were no longer present for any electrode prior to reaching his loudness discomfort threshold. A representative from the implant manufacturer conducted a device integrity assessment, which was inconclusive. The implant team decided to replace the cochlear implant. Prior to replacement, he scored 40% correct on CNC word recognition assessment, and his electrode impedances possessed an atypical pattern (Figure 11–41).

The patient underwent an unremarkable cochlear implant replacement surgery. His Nucleus Freedom implant with a Contour Advance electrode array was activated 2 weeks after surgery, and typical electrode impedances were measured (Figure 11–42). He reported that sound quality was improved with the new implant. However, when he returned for a 1-week, postactivation checkup, his electrode impedances possessed an atypical pattern (Figure 11–43). He reported that he was struggling to understand speech with his new implant, and sound quality was poor. After 4 months of use of the replacement implant, his electrode impedances were unusual (Figure 11–44), and his word recognition was poor (50% correct on CNC words at 60 dBA). Furthermore, his stimulation levels required for adequate loudness progressively increased.




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**FIGURE 11–40.** Case 13: Electrode impedances at 23-months postactivation. *Note.* Provided courtesy of Cochlear™ Ltd.

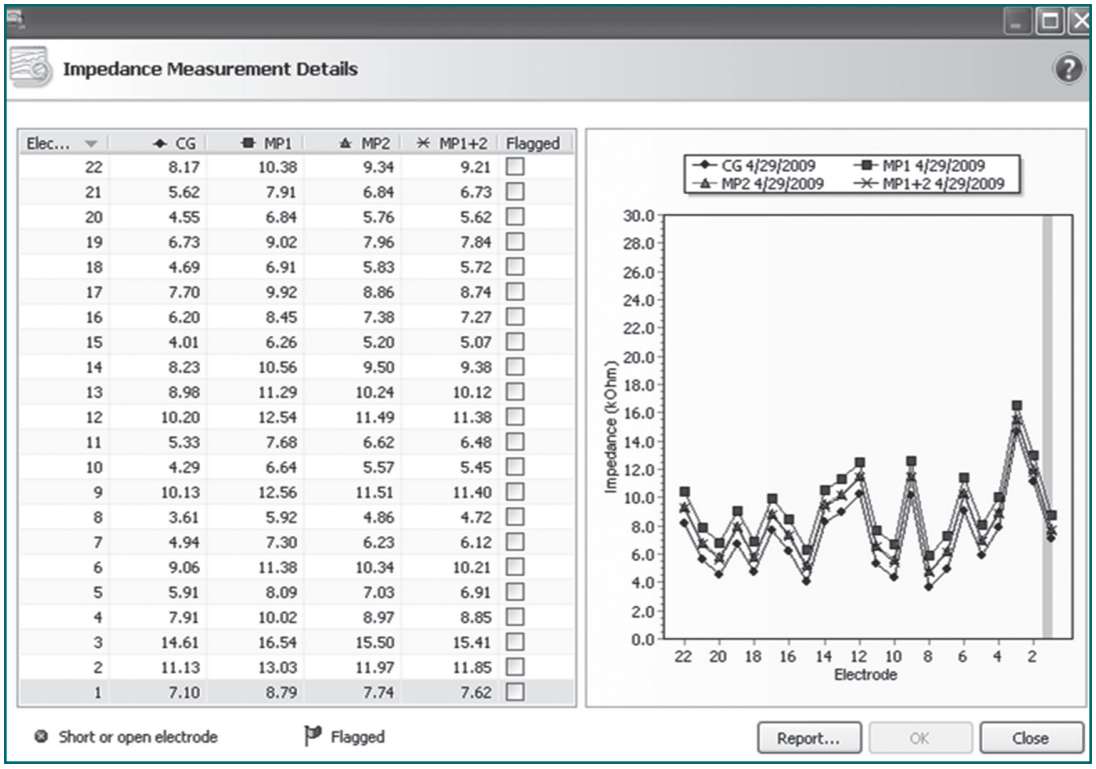
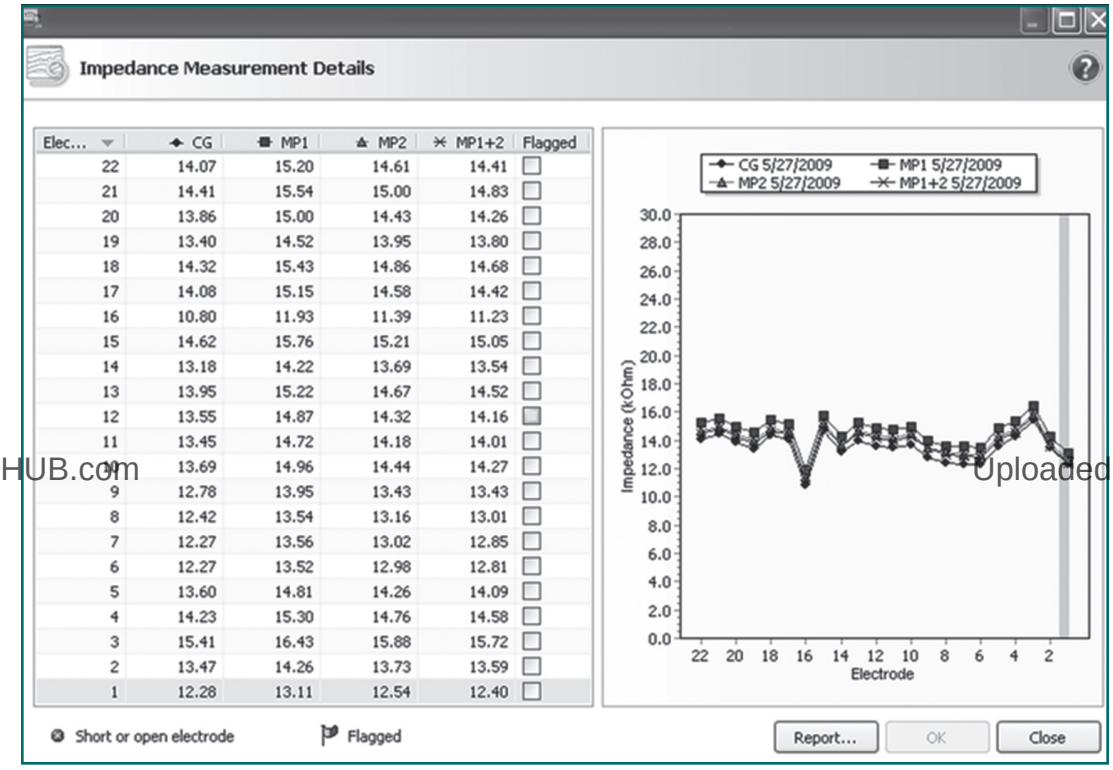



**FIGURE 11–41.** Case 13: Electrode impedances prior to replacement surgery. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–42.** Case 13: Typical electrode impedances measured after cochlear implant replacement surgery. *Note.* Provided courtesy of Cochlear™ Ltd.

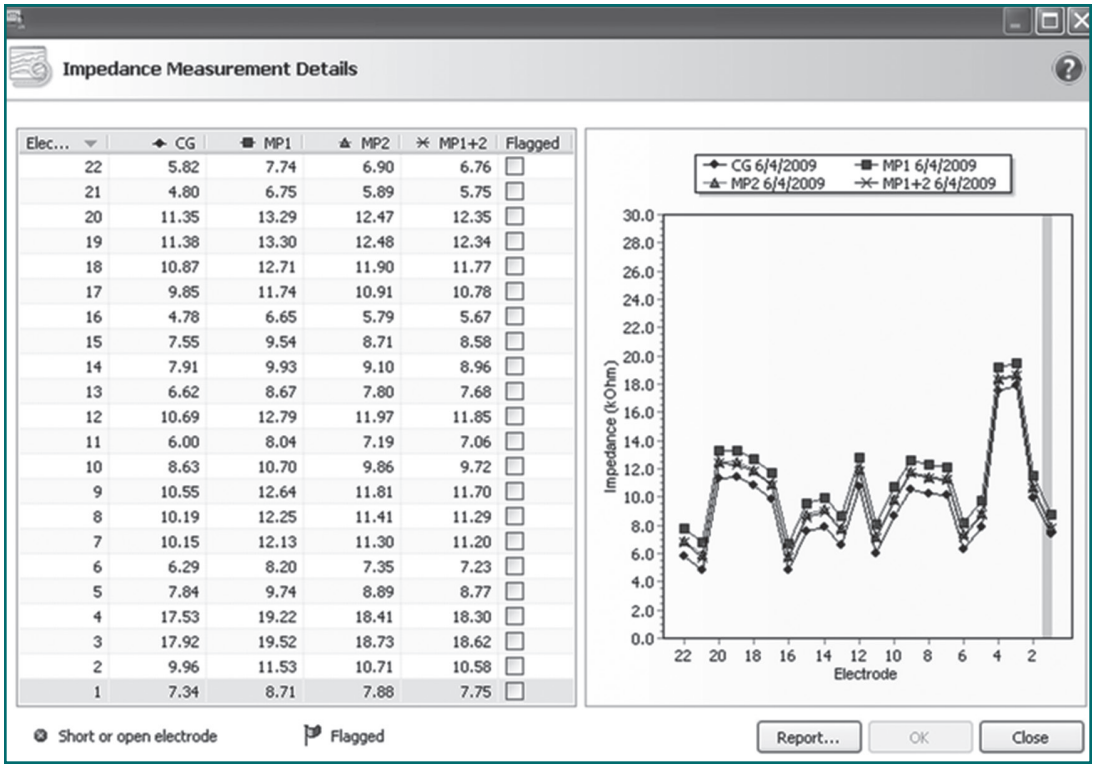
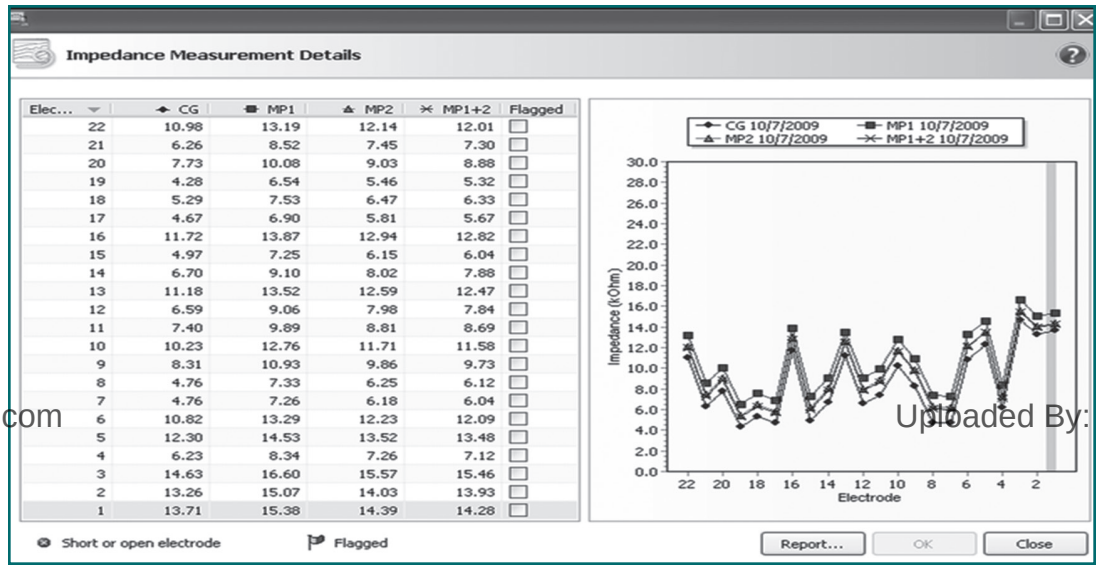



**FIGURE 11–43.** Case 13: Atypical electrode impedances at 1-week postactivation. *Note.* Provided courtesy of Cochlear™ Ltd.



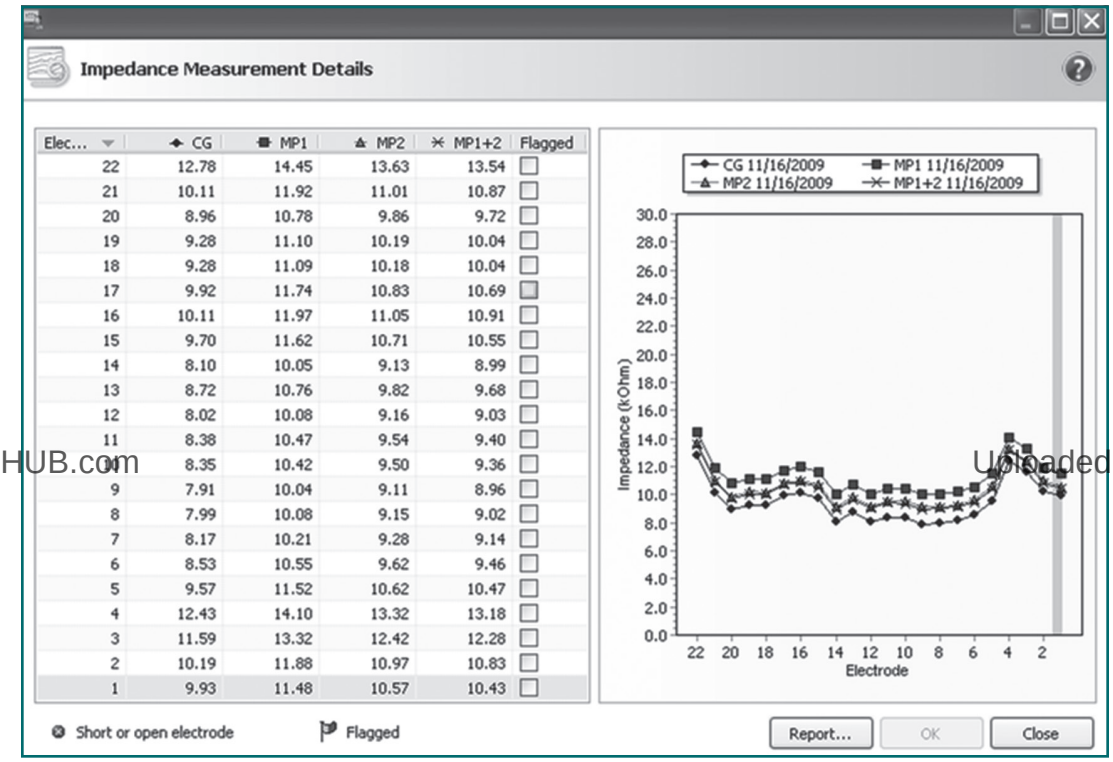

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**FIGURE 11–44.** Case 13: Atypical electrode impedances at 4 months postactivation. *Note.* Provided courtesy of Cochlear™ Ltd.

Given his poor performance, the implant team conducted a comprehensive review of his records. It was noted that the electrode impedances of his second cochlear implant were typical at the initial activation of the implant. The implant team discussed the possibility that the electrical stimulation from the implant may have been causing inflammation of the cochlear tissues, which would result in a change in electrode impedances. Based on that theory, the patient was asked to temporarily suspend the use of his cochlear implant and return to the clinic in 1 week for evaluation.

After 1 week of nonuse, electrode impedances were measured, and they returned to typical values (Figure 11–45). He resumed cochlear implant use and returned to the clinic for assessment. Once again, his electrode impedance values varied considerably across the array (Figure 11–46). Considering the fact that his impedances appeared to become erratic with use of the implant, the implant team recommended the provision of an anti-inflammatory steroid (prednisone) in an effort to address the inflammation that appeared to arise with electrical stimulation. After 3 days of steroid treatment, electrode impedances were measured and had returned to typical values (Figure 11–47). The patient remained on a low dosage of steroids for 3 months, and electrode impedances remained stable.

Unfortunately, his speech recognition did not return to the levels he achieved prior to his initial report of deterioration in sound quality and performance (CNC word recognition ranged from 36 to 60% correct across several appointments). The decision was made to implant the opposite ear. At the time of this writing, he scored 88%, 84%, and 52% on CNC word recognition assessment in the bilateral, left ear (second implanted ear), and right ear conditions, respectively. It should be noted that a wider pulse width (e.g., 50 microseconds) was selected for his programs to avoid the use of high current levels. This recipient is seen every 2 months to monitor performance, electrode impedances, and physiologic responses.




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**FIGURE 11–45.** Case 13: Electrode impedances returned to normal after 1 week of nonuse. *Note.* Provided courtesy of Cochlear™ Ltd.

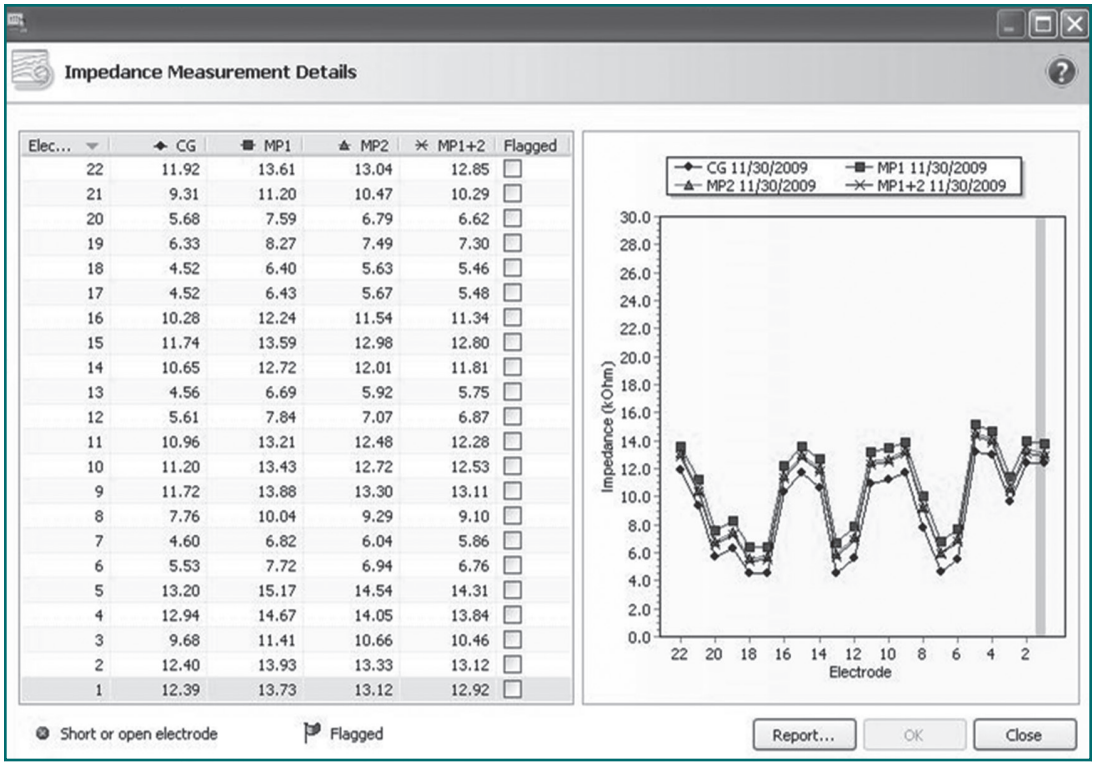
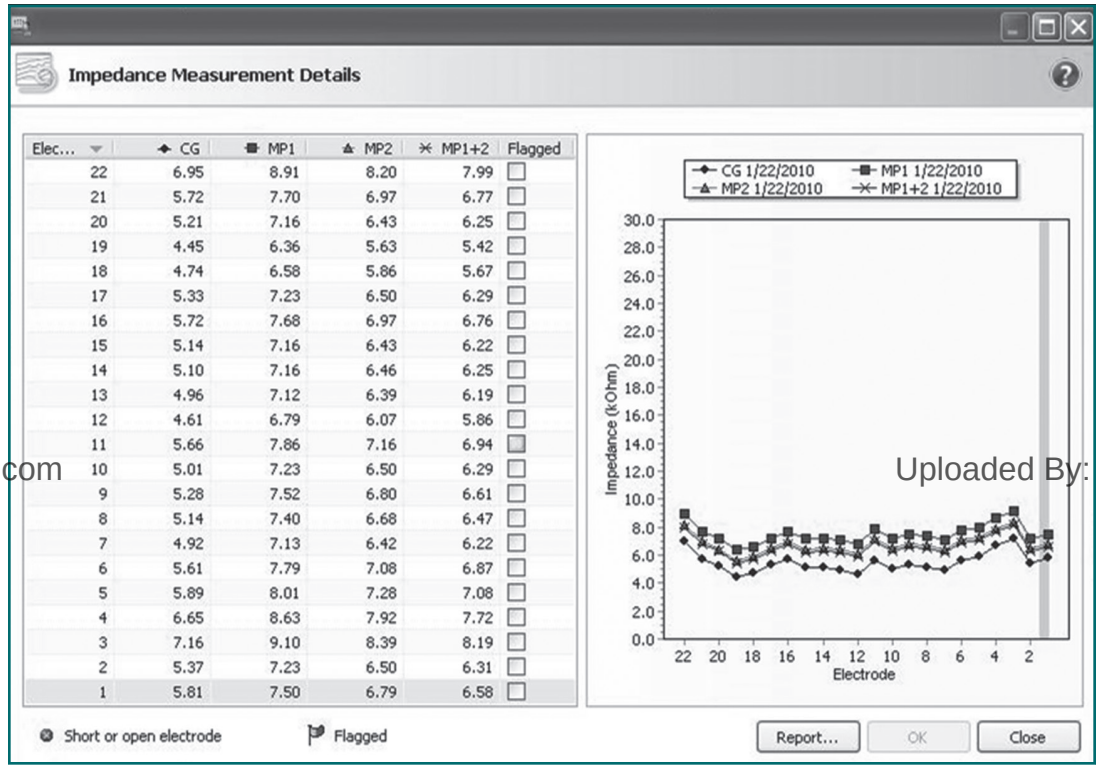



**FIGURE 11–46.** Case 13: Electrode impedance varied considerably across the array after resuming implant use. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–47.** Case 13: Electrode impedances returned to typical values after 3 days of steroid treatment. *Note.* Provided courtesy of Cochlear™ Ltd.

## Take-Home Points

1. Electrode impedances may occasionally change for physiologic reasons rather than a device-related problem. The physiologic reason for the change in this case was not entirely clear, although the recipient did begin receiving testosterone supplements immediately prior to the change in his electrode impedances. Additionally, he has rheumatoid arthritis and indications of a possible autoimmune disorder. It is possible that electrode impedances are known to occasionally change with hormonal fluctuation (i.e., menopause, puberty), viral episodes, overstimulation from the implant, and autoimmune disease.
2. If a recipient complains of a change in sound quality and/or performance and electrode impedances have changed significantly from previous measurements, then it may be prudent to consider medical intervention (i.e., anti-inflammatory steroid, antibiotics) and a period of nonuse of the implant. The clinician should avoid increasing current levels in an attempt to improve performance as doing so may lead to damage to the cochlear structures and/or peripheral auditory neural elements.
3. It is important to routinely obtain electrophysiologic measures, such as the eCAP and eSRT, so a physiologic baseline is available for review when changes in performance or sound quality occur. If eCAP and/or eSRT responses diminish, the clinician should take steps to attempt to prevent damage to the peripheral auditory system. These steps may include a recommendation for pharmacological treatment, a recommendation of nonuse of the implant, and a reduction in stimulation levels until the condition subsides.
4. Routine follow-up is important to ensure the incident does not reoccur.

## CASE STUDY 14: IMPEDANCE CHANGES SECONDARY TO EXCESSIVE STIMULATION

This 51-year-old received bilateral cochlear implants after a hereditary hearing loss had progressed to the moderately severe to profound hearing loss range. His right ear was implanted 9 years ago with an Advanced Bionics HiRes 90 K implant with a 1j electrode array. He used the Advanced Bionics Harmony sound processor, and by all accounts, he was a successful cochlear implant recipient. He continued to use a high-power hearing aid for his left ear. He was able to converse over the telephone, and he held a job as a school administrator, which required him the successfully communicate in adverse listening situations. Previous records indicated that he scored 86% correct on CNC word recognition assessment in quiet with use of his cochlear implant alone.

After using his cochlear implant for 6 years, he transferred to our clinic with a complaint of a sudden decrease in speech recognition and poor sound quality. He noted that the problem arose 2 days prior to his arrival at our clinic and that he was no longer able to communicate over the telephone with his right ear cochlear implant. He scored 44% correct on CNC word recognition assessment with use of the right cochlear implant alone. He reported that his family had a history of progressive hearing loss with an onset in early adulthood. He also noted that a previous otologic evaluation indicated normal cochlear anatomy. His electronic Sound Wave file was obtained in order to review his previous programs, NRI results, and electrode impedances.

A review of his cochlear implant program revealed an unusual M-level configuration with a significant emphasis toward stimulation on low-frequency channels and exceptionally high M levels over most of the electrode array (Figure 11–48). It should be noted that M levels for Advanced Bionics recipients are typically in the 100 to 200 charge unit range with some high-performing users having M levels between 200 to 300 charge units. It also should be noted that in exceptional cases, some high-performing Advanced Bionics implant users will have M levels in excess of 300 charge units, but this is certainly the exception rather than the rule. When queried, he was uncertain of the methods his previous clinician used to set his stimulation levels, and there were no notes in his Sound Wave file that indicated the rationale for how the M levels were set. A chronological review of his programs indicated that his M levels had been gradually increased over the course of the 6 years that he had been using his implant.

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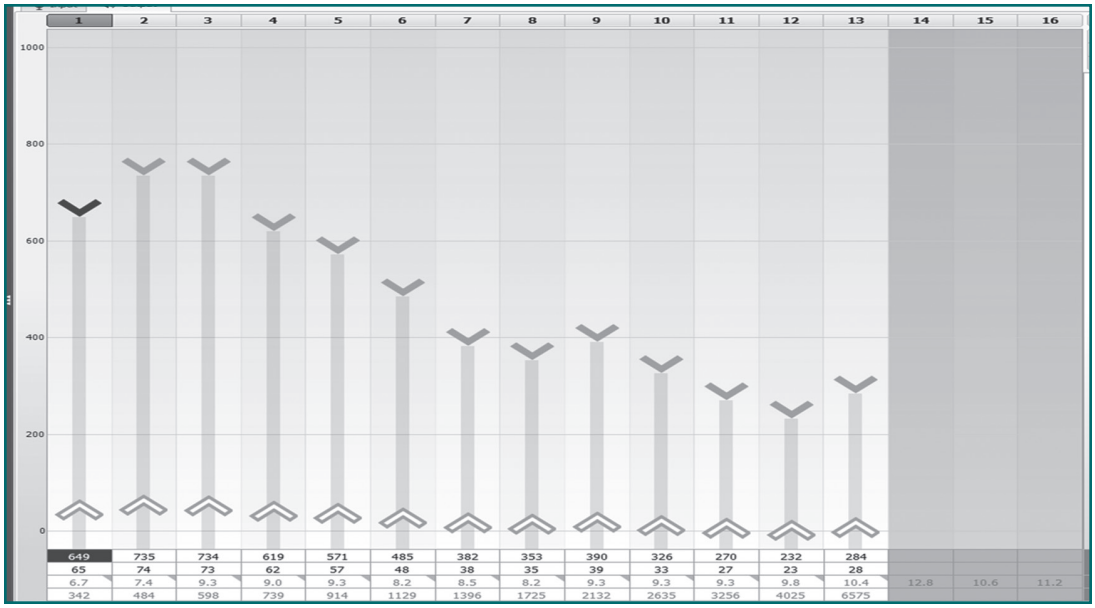



**FIGURE 11–48.** Case 14: M-level configuration high stimulation levels and emphasis toward low-frequency channels. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

Electrode impedance assessment showed no shorted or open electrode circuits. However, when the results of the assessment were compared to electrode impedance results from previous sessions, it was determined that the electrode impedance values of the most apical channels had increased significantly (Figures 11–49 and 11–50). Additionally, the eCAP was measured in the NRI platform, but no responses could be obtained. This was also in contrast to previous sessions in which a robust NRI response was present. Indeed, chronic overstimulation may lead to fatigue and/or damage to the peripheral auditory system, and a loss of physiologic responses (e.g., eCAP, eSRT) may be indicative of damage to or a change in the function of the peripheral auditory system.

It was theorized that this recipient may have suffered from inflammation secondary to excessive electric stimulation of the cochlea. We balanced his M levels in loudness two channels at a time, beginning at the basal end of the array and finishing at the most apical electrodes. This resulted in a flatter M level configuration and a reduction in M-level stimulation for the low-frequency channels (Figure 11–51). The recipient reported that the overall loudness of conversational level speech was insufficient, but he was counseled to attempt to acclimate to the new program. Additionally, he was asked to refrain from implant use except in situations, such as important work meetings, in which it was imperative to use the implant to communicate. Furthermore, he was referred to an otologist on that same day and was placed on a steriod treatment (prednisone) to reduce inflammation.

He returned to our clinic 1 week later. Electrode impedances were similar to previous levels (Figure 11–52). He noted that he had adjusted to the loudness of the M levels that were set the week before. He also noted that he was, once again, able to converse over the mobile telephone when he needed to do so for work purposes. He scored 78% on CNC word assessment. Furthermore, robust NRI responses were once again present. He was seen weekly for the next month to track his progress and his electrode impedances. Both remained stable. He was encouraged to promptly contact our clinic if the difficulties arose again.

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## Take-Home Points

1. Clinicians should be cautious about providing upper-level stimulation levels that greatly

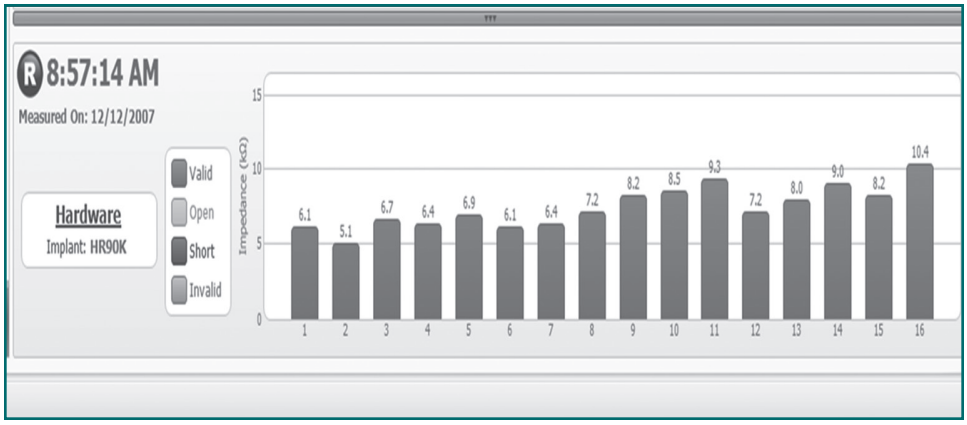



**FIGURE 11–49.** Case 14: Initial electrode impedance measurement. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

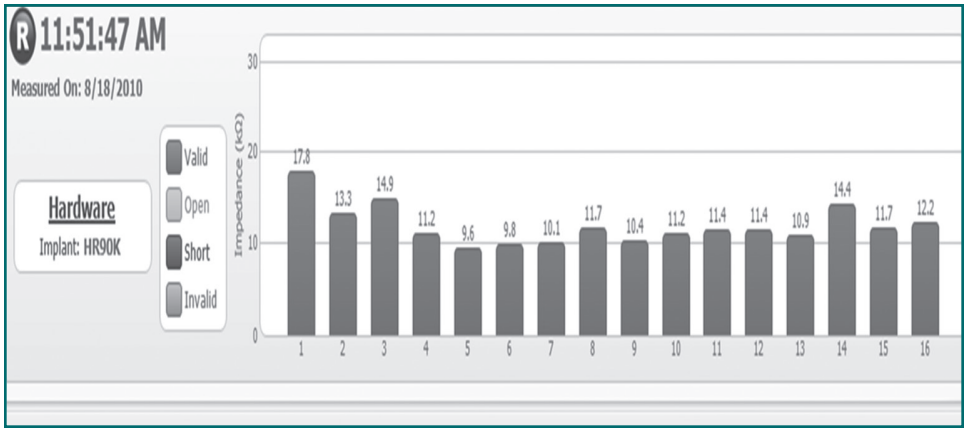



**FIGURE 11–50.** Case 14: Relative to previous measurement (Figure 11–49), electrode impedance values on most apical channels had increased. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

exceed levels appropriate for the typical users. Indeed, some recipients, such as those with a history of bacterial meningitis and/or cochlear abnormalities, will require higher upper-stimulation levels to achieve optimal performance. In all cases, the appropriateness of upper-stimulation levels should be validated by objective measures and assessment of recipient performance.

2. Chronic provision of excessive electrical stimulation may cause cochlear inflammation. This may manifest in the form of a decrease in sound quality and hearing performance, a change in electrode impedance, an onset of tinnitus, and a change in objective physiologic responses (e.g., eCAP, eSRT).
3. When the aforementioned evidence of cochlear inflammation potentially exists, the clinician should react promptly in an attempt to prevent long-term damage. Intervention strategies may include but are not limited to (a) a reduction in stimulation levels, (b) an increase in stimulus pulse width, (c) a recommendation for medical management (e.g., steroids,

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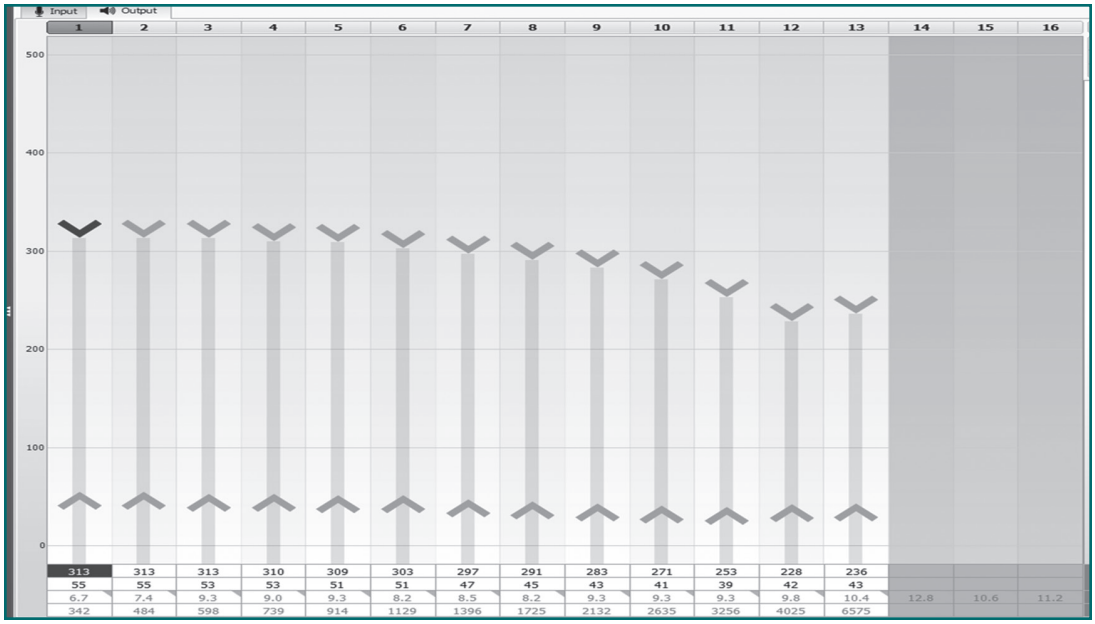



**FIGURE 11–51.** Case 14: Flatter M-level configuration after loudness balancing procedure. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics.com

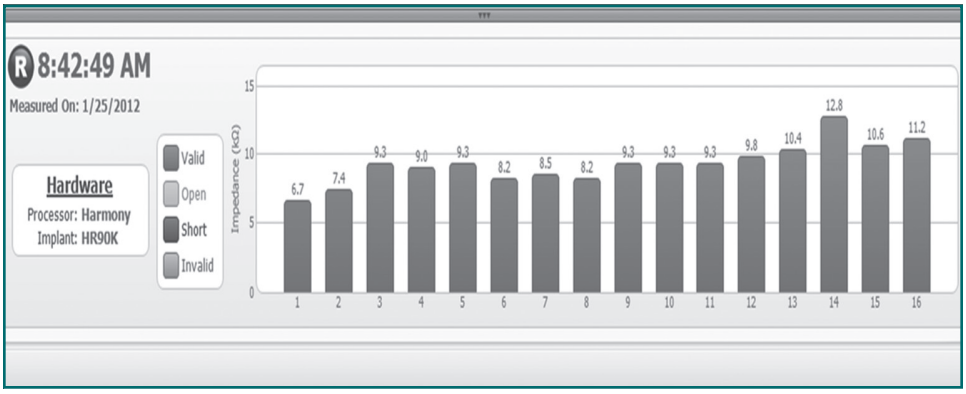



**FIGURE 11–52.** Case 14: Electrode impedances were similar to previous levels. *Note.* Image provided courtesy of Advanced Bionics. http://www.AdvancedBionics .com

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antibiotics), and (d) a period of nonuse of the implant.

4. Clinicians should routinely conduct assessments of recipient performance (e.g., speech recognition) and objective responses (e.g., eCAP, eSRT, electrode impedances) in order to have a baseline of performance to which to occur if a recipient reports to the clinic with acute difficulties.
5. In cases, such as the one described here, clinicians should provide vigilant monitoring to ensure that the problem does not reoccur. The recipient should be instructed to contact the clinic immediately if the problem resurfaces.

## CASE STUDY 15: PROGRAMMING FOR A TEENAGE RECIPIENT WHO HAD SIGNIFICANT RESIDUAL HEARING AFTER IMPLANTATION

This patient was referred to our center at the age of 12 by his school speech-language pathologist who had significant concerns regarding his hearing sensitivity and speech development. His mother reported multiple difficulties at birth including respiratory problems, seizures, oxygen, placenta abruption, and approximately a 1-month stay in the NICU. He was given a 0 APGAR score at birth and was diagnosed with cerebral palsy. It was unknown if this child had a hearing screening at birth, but his mother denied any history of middle ear dysfunction or a family history of hearing loss. Both his mother and speech-language pathologist reported that he responds to speech but does not seem to understand what was said. Additionally, he has been placed in a special education self-contained classroom at school.

At his initial diagnosis at age 12, he was diagnosed with normal hearing through 500 Hz and sloping, moderate-to-profound sensorineural hearing loss in the mid and high frequencies (Figure 11–53). Although cognitively delayed, this patient was very reliable in the soundbooth. He was fitted with binaural Phonak Naída III SP with integrated Phonak FM receivers and a Phonak Inspiro transmitter. Aided thresholds were obtained in the normal sloping to mild hearing loss range, but his speech delays largely remained. He was monitored closely by his audiologist and speech-language pathologist, who both expressed significant concern that he continued to struggle with auditory, speech, and language issues, which were believed to be associated with limited audibility of high-frequency speech sounds.

After he continued to exhibit substantial speech and auditory difficulties following 2 years of hearing aid use, he returned to the clinic for an assessment to determine whether he was a candidate for a cochlear implant. Aided word recognition testing was conducted in three hearing aid conditions using the HINT sentences and CNC words at 60 dBA: right, left, and bilateral. Results on the HINT in the aforementioned hearing aid conditions were 22%, 17%, and 14% and on the CNC words were 36%, 18%, and 44%.

Based on his real-world and clinically measured difficulty understanding speech with hearing aids, the cochlear implant team recommended an implant for the left ear. The patient's family was receptive to this recommendation, and he received

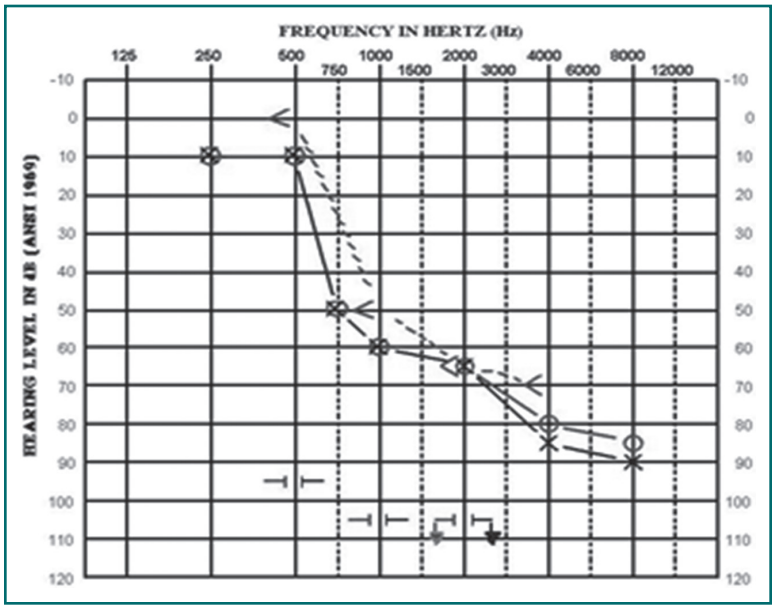



**FIGURE 11–53.** Case 15: Preimplant audiogram.

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a Cochlear Nucleus CI422 implant and a Nucleus 5 sound processor for the left ear when he was 14 years old. He also received a Nucleus CI422 cochlear implant for the right ear when he was 15 years old. Soft surgical techniques were employed during both surgical procedures, and there were no surgical complications for either surgery.

As shown in Figures 11–54 and 11–55, this patient maintained most of his low-frequency residual hearing after his cochlear implant surgery.

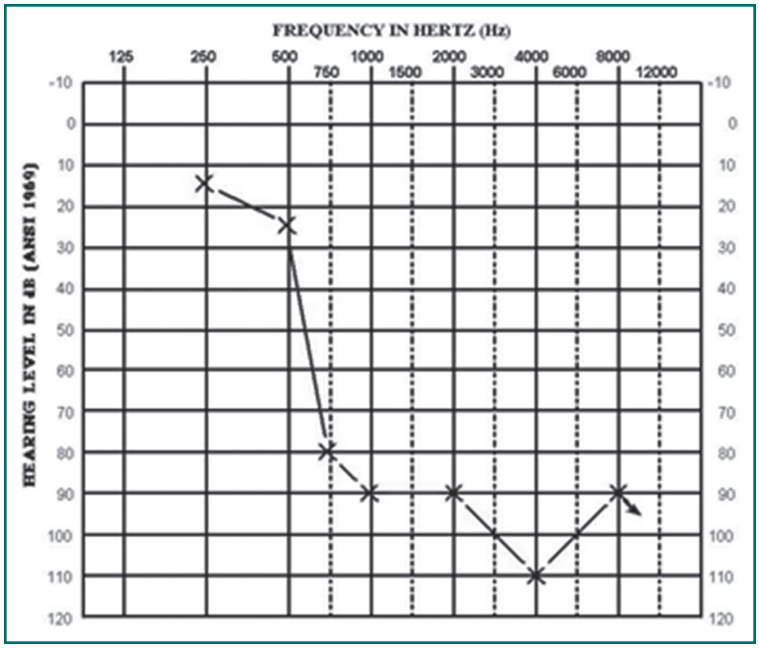



**FIGURE 11–54.** Case 15: Postimplant left audiogram.

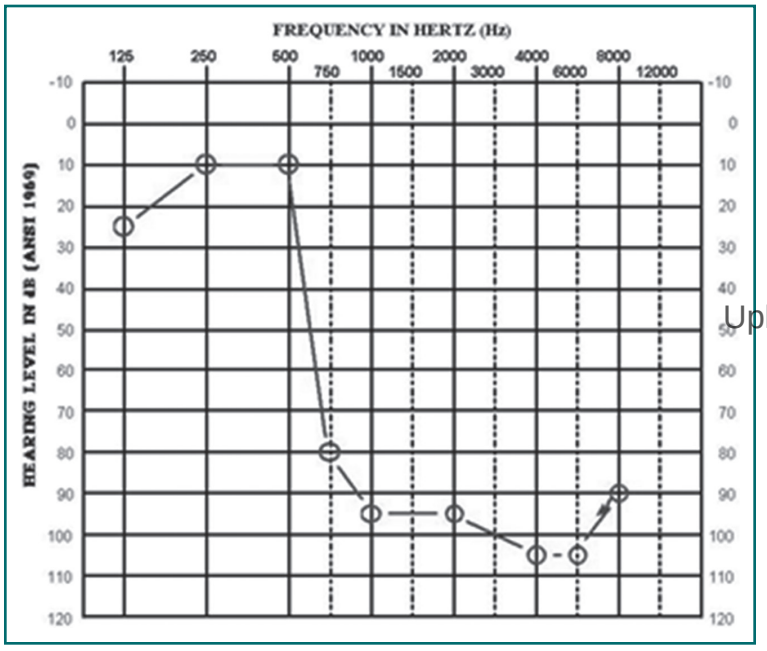



**FIGURE 11–55.** Case 15: Postimplant right audiogram.

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Considering the fact that he has reasonably good access to low-frequency sounds via acoustic hearing, inputs below 500 Hz were allocated to receive acoustical stimulation, whereas inputs above 500 Hz were slated to receive electrical stimulation (Figures 11–56 and 11–57).

This patient has been followed every 3 to 6 months since initial activation. He continues to exhibit preservation of his low-frequency residual hearing. At 9 months postactivation of the left cochlear implant and 6 months postactivation of the right cochlear implant, this patient was able

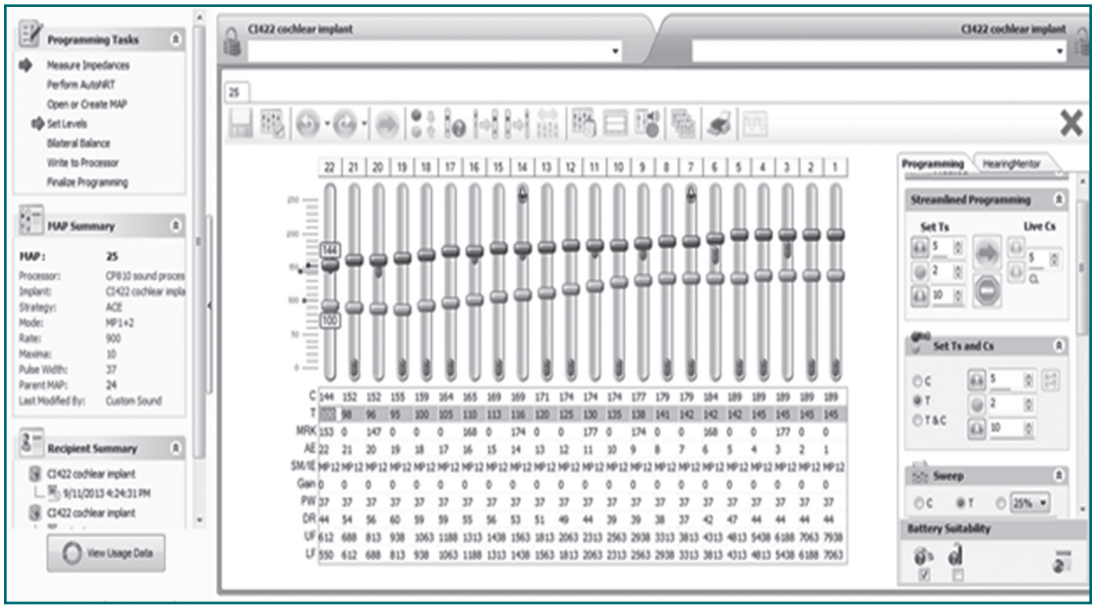
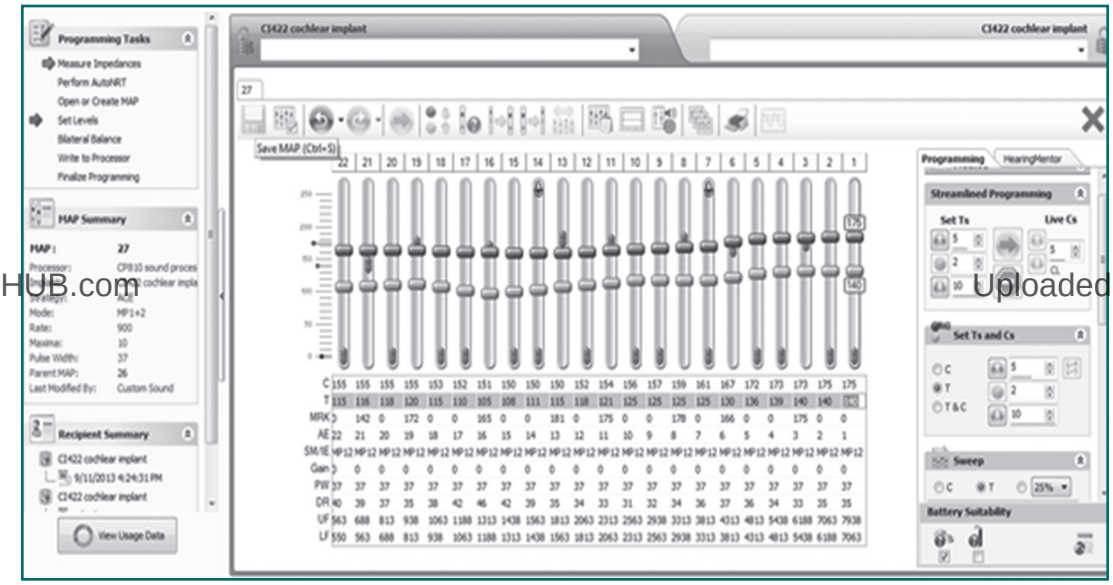



**FIGURE 11–56.** Case 15: Stimulation levels for left implant. *Note.* Provided courtesy of Cochlear™ Ltd.




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**FIGURE 11–57.** Case 15: Stimulation levels for right implant. *Note.* Provided courtesy of Cochlear™ Ltd.

to correctly identify 76% and 84% of CNC words presented at 60 dBA, respectively. After a few months of cochlear implant use, a personal FM system was fit (Phonak Insprio transmitter with ML14i integrated receivers). The FM system is used during all academic classes and at home. Currently, this patient remains in a self-contained classroom. At a recent Individualized Education Plan (IEP) meeting, his teachers reported that he made significant progress since receiving his cochlear implants. Further audiological testing will be completed to ensure that this patient retains his residual hearing.

## KEY CONCEPTS

- Cochlear implant programming is best accomplished by using a combination of subjective and objective measures.
- The provision of appropriate stimulation levels is most likely the most important component of an optimal program for a cochlear implant user.
- Stimulation levels should be verified by the use of objective measures.
- The clinician should be concerned when upper-stimulation levels are erratic (i.e., vary substantially across the array) or are markedly different from levels that are typical for a given make and model of implant/electrode array.
- When possible, upper-stimulation levels should be balanced in loudness across the electrode array.
- Insufficient T levels will compromise access to low-level sounds.
- The clinician should routinely conduct sound-field detection threshold assessment and speech recognition assessment to ensure a recipient's satisfactory performance. The clinician should generally possess high expectations for recipients who have favorable preimplant histories and use contemporary cochlear implant technology.
- Electrode impedance assessment should be conducted routinely, and the clinician should closely inspect impedance results in order to ensure the recipient's implant is functioning appropriately.
- The clinician should work with a team of professionals to provide optimal evaluation and support of cochlear implant candidates and recipients.
- When an implant recipient is inexplicably performing poorly, then the clinician should consider contacting the implant manufacturer in order to request additional assessment of the integrity of the implant. Cochlear implant replacement is warranted when assessment indicates faulty implant function or when a recipient performs poorly without a reasonable explanation.
- A cochlear implant recipient's performance is dependent upon the recipient's physiologic function, the integrity of the cochlear implant and sound processor, and the quality of the CI sound processor program (i.e., MAP) created by the clinician.

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