# A Uniform Graphical Representation of Intensity Coding in Current-Generation Cochlear Implant Systems

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**Objectives:** Understanding and predicting the impact of MAP changes on the electrical current delivered at the level of cochlear implant (CI) electrodes is challenging. However, it is an important prerequisite for effectively programming these devices in clinical practice. This article describes a graphical representation to illustrate the intensity-coding behavior of four CI systems (Cochlear, MED-EL, Advanced Bionics, and Neurelec). Design: For this the authors have broken down the intensity coding into two separate transformations: (1) from broadband acoustical input to band limited channel amplitude and (2) the mapping function within a single channel. These functions have been synthesized and presented in a uniform plot across brands.

**Results:** The plot describes the output of a CI channel in response to different input signals. This has been incorporated in an interactive software application that illustrates the different stages of intensity coding and the impact of the relevant fitting parameters for each CI brand.

**Conclusions:** The plot provides the clinician with an assistive tool to better understand and predict the behavior of CIs, which may lead to more knowledgeable interpretation and CI programming.

**Key words:** Cochlear implant, Fitting, Intensity coding, Loudness growth, Optimization, Programming, Technical Summary.

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## INTRODUCTION

Cochlear implants (CIs) are now widely accepted as an effective treatment for profound deafness (Wilson 1991). After surgical implantation, the sound processor must be appropriately programmed and customized for the individual, which is commonly called fitting. The aim of fitting is to set a number of parameters to ensure that the electrical pattern generated by the internal device in response to sound, yields an optimal auditory percept (Cope & Totten 2003; Shapiro 2012). Several tuning parameters are available, and all their values together are commonly called the MAP.

The main focus in current-generation CI systems lies on compressing the wide range of intensities present in acoustical input signals into the limited range that is available for electrical stimulation. Hence most of the MAP parameters relate to the coding of intensity while only few relate to other sound-coding features, like the spectral mapping. A recent global survey on CI fitting practices has shown that in most cases fitting is restricted to setting the threshold of audibility for electrical stimulation, and a level of upper tolerance limit, for each electrode separately (Vaerenberg 2014). Those levels define the electrical dynamic range (EDR) of each electrode and will be referred to as EDR Minimum and EDR Maximum, respectively. Other MAP parameters may also affect the system's mapping of intensities, and adjusting them has been shown to produce better outcomes for individual recipients (Vaerenberg 2014). Yet, these additional parameters are left at default values in most cases (Vaerenberg 2014). The authors believe that the reasons for this are multiple. A major reason may lie in the intrinsic complexity of the CI system and its sound processing and the differences, often subtle, in the underlying technologies used by the different CI devices. This makes it difficult to predict the impact of a specific MAP parameter on the behavior of a given CI system. In addition, clinicians often program devices from different manufacturers while features/parameters with similar names across brands may be implemented differently. For instance, the channel gains in Cochlear's system are applied at the input of a channel, while in AB devices they are added to the channel's output. Although this has a similar effect on loudness in both systems, it may induce different effects on threshold, maximum stimulation level, among others, as will become apparent later in the article.

In this article we present a uniform graphical representation that illustrates the effects of parameter changes on the CI's output for all currently available CI systems. This representation has been incorporated in an interactive software application that allows a dynamic visualization of the effects of chosen MAP settings. We believe that such a comprehensive summary of the behavior of CI systems, represented in a uniform way across brands, may assist the audiologist in gaining more insight into the clinical behavior of these systems and in further optimizing the fitting process.

It is beyond the scope of this article to explain the meaning of all possible MAP parameters found in the various CI systems. For this information, the reader is referred to the manufacturer's user manuals and clinical guidelines, and to existing comprehensive overviews (Wolfe 2010; Shapiro 2012).

## MATERIALS AND METHODS

### Visualization of the Intensity-Coding Function

To visualize the input–output relation of a CI system, we established a three-axial graphical representation to reflect the three major stages that can be indentified in the signal-processing path of all current-generation systems (Fig. 1): (1) an acoustical stage where the broadband signal is captured by a microphone and preprocessed; (2) a digital stage where the signal has been digitized and its energy is distributed over a number of channels, and finally (3) an electrical stage where the energy in each channel is mapped to an electrode-activation level.
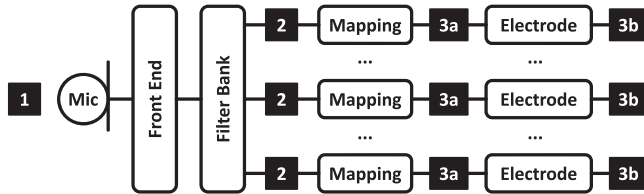
Fig. 1. Cochelar implant processing path block diagram showing the different stages at which signal levels are considered: (1) the intensity of the acoustical signal; (2) the amplitude of the band limited digital signal at the output of the filter bank; and (3) the magnitude of the electrical stimulation, either as displayed in the fitting software (3a) or as the equivalent amount of charge delivered by the electrode (3b).

Figure 2 depicts the acoustical stage at the right horizontal axis (marked as 1), the digital stage at the vertical axis (marked as 2), and the electrical stage at the left horizontal axis (marked as 3). At any of these stages the signal level is expressed in an appropriate unit. The acoustical input signal level is presented in broadband dB SPL (stage 1). The signal is then digitized and preprocessed (e.g., beam-forming, noise reduction, wind noise reduction, dereverberation, etc.), which often also includes the application of a preemphasis filter and a gain factor, before it is split into separate frequency bands. The resulting signal level at the output of the filter bank is no longer a direct representation of acoustical sound pressure. Instead, this signal level is expressed relative to a digital maximal value known as full scale (FS; 0 dB FS, i.e., the largest signal amplitude that can be expressed by the internals of the CI system, stage 2). It is essential to be aware that at this point the energy of the input signal is divided over multiple channels, which makes the dB FS represent a narrowband energy in a single channel. Finally, using the values of EDR Minimum and EDR Maximum, this energy is mapped to an electrode-activation level. That level may be expressed in a "clinical" unit that is displayed to the user of the fitting software (stage 3a) or it may be expressed in a unit of the equivalent charge per pulse phase at the channel's electrode contact (i.e., nanoCoulomb (nC), stage 3b).

Hence, the acoustical axis of Figure 2 displays the Input Sound Level ranging from 0 to 120 dB SPL; the digital axis displays the Channel Amplitude ranging from −100 to 0 dB FS; and the electrical axis displays the Charge Output ranging from 0 to 30 nC. As such the three axes in Figure 2 allow the visualization of two distinct transformations: on the right one can see the preprocessing that operates on the broadband signal, and on the left one can see the mapping function that is used to transform the narrowband energy within a single channel to an electrode-activation level. The combined graph is called

the Intensity Coding Function (ICF) plot, and the interactive software application that allows the dynamic visualization of these plots in function of chosen MAP settings is called Intensity Coding in CIs (ICCI). Both are explained in more detail further in the text.

## Scope, Constraints, and Disclaimer

The different CI systems currently available all have their particular signal-processing strategies and features, which makes it hard to produce a general model that covers them all. For this reason some constraints have been applied, which allow the graphs to have a well-defined scope in which they should be interpreted. First, the plots are only reflecting the current-generation (anno 2013) CI systems using their default sound-coding strategies. For Cochlear (Sydney, Australia), this is the CP810 processor with the ACE strategy and CI512 implant; for MED-EL (Innsbruck, Austria), this is the OPUS 2 processor with the FS4 strategy and CONCERTO or SONATA implant; for Advanced Bionics (AB, Stäfa, Switzerland) this is the Naída CI Q70 processor with the HiRes strategy and HiRes90k implant; for Neurelec (Vallauris, France), this is the Saphyr processor with the Digisonic SP implant. Second, a number of features have been excluded from the analysis, because they are either of little relevance to the essentials of intensity coding or too dynamic (dependent on temporal and spatial aspects of the input signal) to be visualized on a static graph. These features include: input mixing and beam-forming (the use of multiple microphones, optionally in combination with telecoil, aux, etc.), volume-control options that can be manipulated by the recipient, noise-reduction algorithms, temporal aspects of speech coding strategies, etc. These features have not been included as variable parameters in ICCI and their effect on the device behavior is ignored. The features that have been included are: EDR Minimum, EDR Maximum, Instantaneous Mapping Range (IMR), Input Gain, Output Gain, Input Compression, and Output Compression. Those features and the parameters they relate to are summarized in Table 1 for each of the manufacturers. EDR Minimum and Maximum define the range of stimulation levels for each electrode. IMR (expressed in dB) is the range of input sound levels that is being mapped into the EDR at any given instant in time. Input Gain is the application of a gain factor to the broadband input signal. Output Gain is the application of a gain factor per electrode. Input Compression is the long-term compression of the broadband input signal due to automatic gain control (AGC) systems and the like. Output Compression is the instantaneous compression applied per channel (in the mapping function).
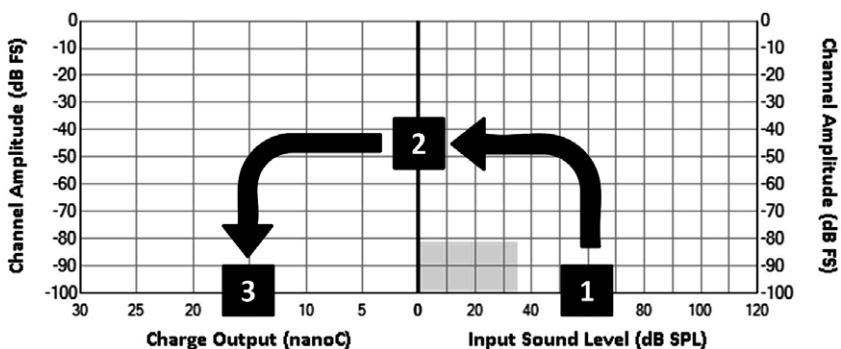


Fig. 2. The empty intensity-coding function plot showing the three dimensions relating to the three signal-processing stages: (1) the acoustical dimension on the right horizontal axis, representing the broadband input sound level (dB SPL); (2) the digital dimension on the central vertical axis, representing the narrowband channel amplitude (dB FS); (3) the electrical dimension on the left horizontal axis, representing the charge output (nC).

**TABLE 1. Processing features that are taken into consideration in the intensity coding function with their corresponding parameter names across brands**

|  | Cochlear | MED-EL | Advanced Bionics | Neurelec |
|---|---|---|---|---|
| EDR minimum | T (CL) | THR (QU) | T (CU) | Min (μs) |
| EDR maximum | C (CL) | MCL (QU) | M (CU) | Max (μs) |
| IMR | C-SPL - T-SPL | Adaptive Sound Window | IDR | [Fixed] (85 dB) |
| Input gain | Sensitivity (dB) | AGC Sensitivity (%) | Sensitivity (dB) | Analog Gain (dB) |
| Output gain | Gain (dB) or ADRO | [Fixed] (0 dB) | Gain (dB) | Gain (dB) |
| Input compression | ASC | AGC Compression Ratio | AGC | [Fixed] |
| Output compression | Loudness Growth (Q) | Maplaw Compression | [Fixed] | Volume |

This table lists the parameters that are primarily related to the features. In MED-EL's system both Input Gain and Input Compression are affected by the combination of AGC Sensitivity and Compression Ratio parameters.
ADRO, adaptive dynamic range optimization; AGC, automatic gain control; ASC, autosensitivity; EDR, electrical dynamic range; IDR, input dynamic range; IMR, instantaneous mapping range; MCL, most comfortable level; Q, loudness growth; THR, threshold.

Even for those MAP parameters that are included, not all technical details are implemented by the ICCI application. As such it should be considered and used as an assistive tool that provides an indication of the behavior of these systems rather than an exact simulation of their signal-processing algorithms. This is important because it may give rise to inconsistencies between the technical documentation provided by the CI companies and the output of the ICCI application. The reader should understand that for technical accuracy, the documentation from the CI companies always prevails. Nonetheless, the authors are convinced that the possible inconsistencies in the ICCI application are compatible with the principal aim of this article and the ICCI application, namely to grant a principal understanding of the intensity coding in CIs. Some of the authors have responsible positions at CI companies. Their contribution has been essential to the development of the graphical representations and the content of this article. But neither they nor the CI companies can be held legally responsible for any such inconsistencies that have been unavoidable in the interest of the comprehensibility of this article and the associated application. The use of ICCI and interpretation of the ICF plots is intended as an assistive tool for the competent clinical CI-programmer who remains fully and solely responsible for their use in the clinic.

## Sources of Information

The ICFs were synthesized from a number of existing documentation sources and verified through interviews with the manufacturers' engineers. Because all manufacturers provide general descriptions of their CI systems through their fitting software, basic concepts have been used from Custom Sound 3.2 Help contents and the Cochlear™ Clinical Guidance Document (Cochlear Ltd. 2012), the Maestro 4.0 Help contents (MED-EL G.m.b.H 2012a) and a FocusOnFineHearing™ Technology document (MED-EL G.m.b.H 2012b), the SoundWave 2.1 Help contents (Advanced Bionics LLC 2012), and the Digimap 3.4 Help contents (Neurelec SA 2012). These sources are designed to provide assistance in performing specific programming tasks, but often lack detail and cohesion with regard to the composite signal-processing chain as a whole. To be able to construct the ICFs additional information was required. That information has been partly obtained from published articles (Hochmair 2006; Koch 2007), presentations at conferences, and fragmented documentation that has been collected by the authors over the years. Individual interviews with company engineers were conducted to complete and validate the required sources.

## The ICF Plot

**Input Signals** • The ICF is plotted in response to any of three types of signals: a pure tone (adjustable in frequency between 100 and 8000 Hz), a speech signal or a white noise signal. It is thereby assumed that: (1) the frequency of the pure-tone signal equals the center frequency of the observed channel's filter band, and as such completely falls into that channel (there is 0 dB attenuation with regard to the broadband energy); (2) the speech signal does not fall into a single channel, it is distributed over multiple channels in such a way that the observed channel receives energy from the speech signal that is equal to the broadband energy attenuated by 12 dB; (3) the white noise signal is distributed over all channels such that the observed channel receives energy from the white noise signal that is equal to the broadband energy attenuated by 24 dB. The attenuation of 24 dB relates to the fact that the average CI channel has a bandwidth of 493 Hz that is 1/16 of the total bandwidth (8 kHz) of current-generation systems. For the speech signal, having at any given moment a bandwidth between those of a pure tone and white noise, the attenuation inflicted by the filter bank was chosen to be the mean of the attenuations of the pure tone (0 dB) and the white noise (24 dB) signals, hence 12 dB. It must be noted that the choice of these attenuations is a simplification and that in reality the attenuation is highly dependent on how the channel bandpass filter is organized in relation to the input signal (e.g., a white noise will be attenuated more in low-frequency channels than in high-frequency channels because channel bandwidth typically increases with frequency). It is also assumed that all input signals feature a stable long-term intensity by which AGC systems reach convergence (i.e., broadband intensity is maintained stable for a time longer than the system's attack time). This response to long-term intensities incorporates, among others, the static gain function of AGC systems (a new gain is determined from this function when the long-term intensity of the input signal changes). Nonetheless, the ICF plots in ICCI, as illustrated in Figure 3, allow showing the response to rapid fluctuations (open symbols) around this long-term intensity (filled symbols), of which it is assumed that they do not trigger the slow detectors of AGC systems. Appendix B displays these kinds of plots for the four CI brands at three different intensities (35, 65, and 95 dB SPL; see Supplemental Digital Content, http://links.lww.com/EANDH/A140).

**Microphone, Preemphasis, System Noise Floor** • In the ICF plots the front end preprocessing and filter bank steps are combined and depicted on the right chart (Fig. 2). The
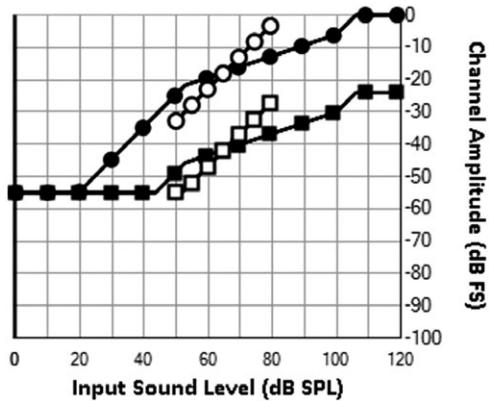
Fig. 3. Example plot displaying MED-EL's transformation of acoustical input to channel amplitude for both pure-tone (circles) and white noise (squares) signals. The response to long-term intensity is plotted in filled symbols. The response to fast deviations (±15 dB) from a long-term average intensity of 65 dB SPL is plotted in outline symbols.

microphone's frequency response is not considered (i.e., it is assumed to be flat). The system noise (primarily determined by the microphone noise floor) is assumed to be white within the range being processed (0.1 to 8 kHz) and equivalent to 35 dB SPL within an acoustical band of that same range. As with the white noise input signal, the energy per channel is assumed to be equal to the broadband energy attenuated by 24 dB, and would therefore be equivalent to approximately 11 dB SPL per channel on average, as illustrated in Figure 4. For all devices, the preemphasis filter is assumed to be an A-weighting filter. The ICCI application allows adjusting the observed channel's center frequency, such that the effect of preemphasis becomes apparent. In this article, however, all ICF plots depict channels centered around 1000 Hz, hence the attenuation of the preemphasis filter equals zero.

**Mapping** • The mapping step is depicted on the left chart of the ICF plots (Fig. 2). It maps the channel's amplitude into the "Electrode-Activation Level" expressed in the unit that is presented to the clinician in the manufacturer's fitting software. Strategy-specific features (e.g., current steering; Koch 2007) are not considered in the ICF plots. There is an option
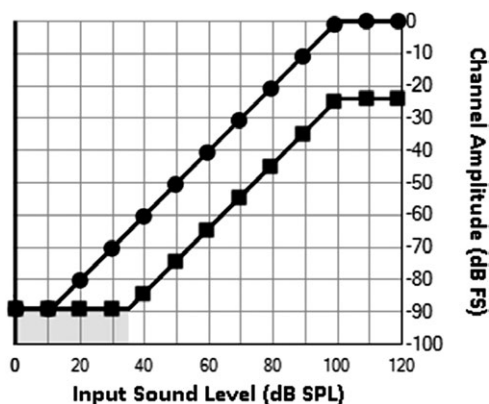
Fig. 4. Example plot showing Neurelec's system noise floor (equivalent to 35 dB SPL) as the light gray area at the bottom left. It causes an 11 dB SPL equivalent floor effect on the channel amplitude of pure tones (circles) and a 35 dB SPL equivalent floor effect on white (0.1 to 8 kHz) noise signals (squares).

to convert the manufacturer-specific level to its general charge equivalent, expressed in nanoCoulomb per pulse phase, as illustrated in Figure 5.

## RESULTS

We have established a three-stage graphical representation of intensity coding for the four brands of current-generation CI systems. This representation is available as static graphs as shown in Appendix A, but it is more useful in the interactive dynamic software application ICCI (given later in the article; see Supplemental Digital Content, http://links.lww.com/EANDH/A140). It illustrates the impact of MAP parameters on the coding of intensity and allows understanding and predicting the transition from free-field acoustical signals through digital band limited energies into electrical stimuli at the output of the CI channels/electrodes. This approach reveals a number of MAP features that are common to all four implant brands and a number of device-specific features and how they impact the coding of sound.

### Common Features

**Input Sensitivity/Gain** • All brands expose a parameter that can be used to adjust the microphone sensitivity. In Cochlear and AB devices this is called Sensitivity. In Neurelec devices this is called Analog Gain. These three operate as a fixed gain applied to the input signal. In MED-EL, microphone sensitivity is defined by a combination of the AGC Sensitivity and the AGC Compression Ratio. Increasing the AGC Sensitivity shifts the AGC knee-point toward low-level sounds, resulting in softer level sounds to be represented within the EDR. Increasing the Compression Ratio reduces the part of the EDR that is assigned to signals above the AGC knee-point, also resulting in softer level sounds to be mapped into the EDR.

In addition to static microphone sensitivity, some CI systems have a mechanism that automatically adjusts the microphone sensitivity based on environmental sound levels. In Cochlear devices this is called Autosensitivity (ASC), which uses the environmental noise level to determine an appropriate input gain. In MED-EL and AB devices this is called Automatic Gain Control (AGC), which are dual-loop AGC systems. Although they are implemented quite differently, both use the input signal level to determine an appropriate gain. They contain a slow detector with a relatively long attack time, which means that they work as a volume control. Neurelec processors do not have an AGC feature.

**Noise Floor, IDR, IMR, and Saturation** • All manufacturers provide default values for their fitting parameters that cause the system noise (as described earlier in the article) to be excluded from the EDR and therefore to not be perceived by the recipient. This is accomplished by effectively limiting the range of signal levels (IMR) that is mapped into the EDR at any given time. In Cochlear's device the position and size of the IMR can be set using the T-SPL and C-SPL parameters (Fig. 6). AB provides an input dynamic range (IDR) parameter that affects the size of the IMR only. Neurelec and MED-EL have a fixed IMR. The settings for input sensitivity/gain as described earlier in the article also impact the effective IMR in all brands. Moreover, the use of AGC systems introduces the need to distinguish two different approaches to the input range: (1) the range that is instantaneously considered (IMR) and (2) the entire range (IDR) that
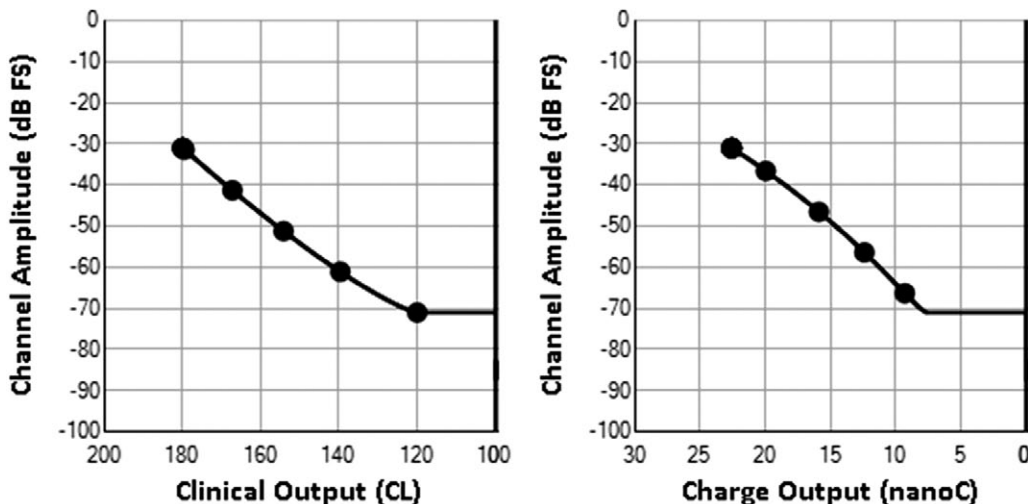
Fig. 5. Example plots showing Cochlear's mapping function expressed in both their clinical unit (left chart) and its equivalent charge per pulse phase (using a Pulse Width of 50 µsec) (right chart). FS indicates full scale.

is covered when shifting this instantaneous range in response to the automated adjustment of an input gain/sensitivity factor.

All four brands have an upper limit for input signal level around 100 dB SPL. Beyond that level the systems are saturated. Given the 11 dB SPL equivalent noise floor within a single channel, this means that CI systems could provide IDRs of up to 90 dB. This 90 dB, however, is to be mapped into an EDR that is an order of magnitude lower (typically below 10 dB), meaning that there is a trade-off between range and resolution. Cochlear maximizes the electrical resolution by limiting its IMR to a default of 40 dB. In MED-EL devices a fixed IMR of 55 dB is used. AB uses 60 dB by default. Neurelec maximizes range and maps 85 dB into its EDR.

**Cochlear** • Figure 7 shows ICF plot for Cochlear with the impact of its MAP parameters. Cochlear uses the term IIDR to refer to IMR. Threshold (T) and maximum comfort (C) levels determine the range and position of the EDR. The IMR/IIDR is set by the T-SPL and C-SPL parameters, but changes in microphone sensitivity (also when determined dynamically by the ASC) also impact its position (they alter the softest level sound that is mapped into the EDR). Changes in Sensitivity also alter the automatic gain control (AGC) knee-point (i.e.,

the input level corresponding to C-level stimulation, not to be confused with AB's or MED-EL's AGC systems, which serve as volume controls). As a consequence, C-SPL and T-SPL parameter values only reflect actual input levels if the Sensitivity parameter value is taken into account (i.e., a C-SPL value of 65 only maps 65 dB SPL to C-level if Sensitivity is set to 12 (default); if Sensitivity is set at 0 then a C-SPL value of 65 would map 77 dB SPL to C-level. In addition, the ASC might adjust the sensitivity based on environmental sound levels. In quiet, a program with ASC acts similarly to a program with no additional input processing, with a fixed sensitivity setting. But if the level of background noise is above the ASC Breakpoint (57 dB by default), sensitivity is reduced according to the level of the noise so that the peaks of speech exceed the long-term average noise spectrum by at least 15 dB (Cochlear Ltd. 2012). ASC can only reduce, not increase, the Sensitivity value set in the MAP (if a Sensitivity of 0 is set, ASC has no effect).

An AGC system with infinite compression keeps the broadband input signal within range. Signals above the C-SPL are not clipped but their peak output is effectively limited to C-SPL (65 dB) by the AGC (Attack Time = 5 msec; Release Time = 75
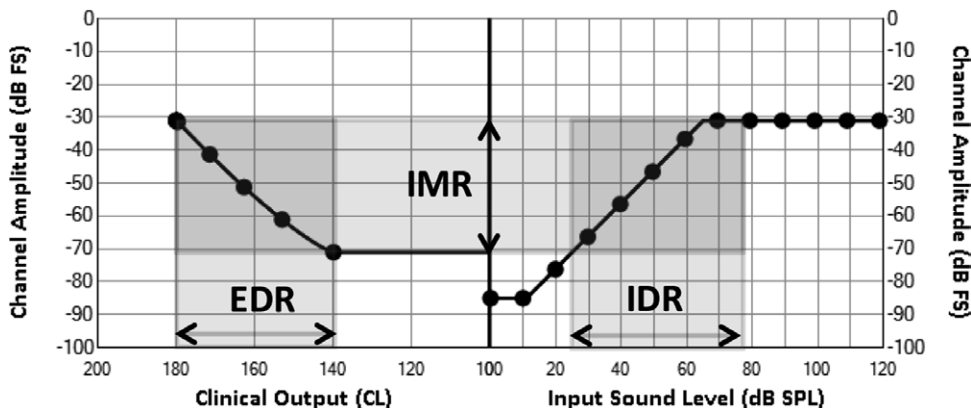


Fig. 6. Example plots showing Cochlear's mapping of IDR into an EDR between 140 and 180 CL when using default parameter values (T-SPL=25, C-SPL=65, Sensitivity = 12). The IMR of 40 dB (between T-SPL and C-SPL) is potentially extended to an IDR of 52 dB when ASC is enabled and Sensitivity is at 12. ASC indicates autosensitivity; EDR, electrical dynamic range; IDR, input dynamic range; IMR, instantaneous mapping range.
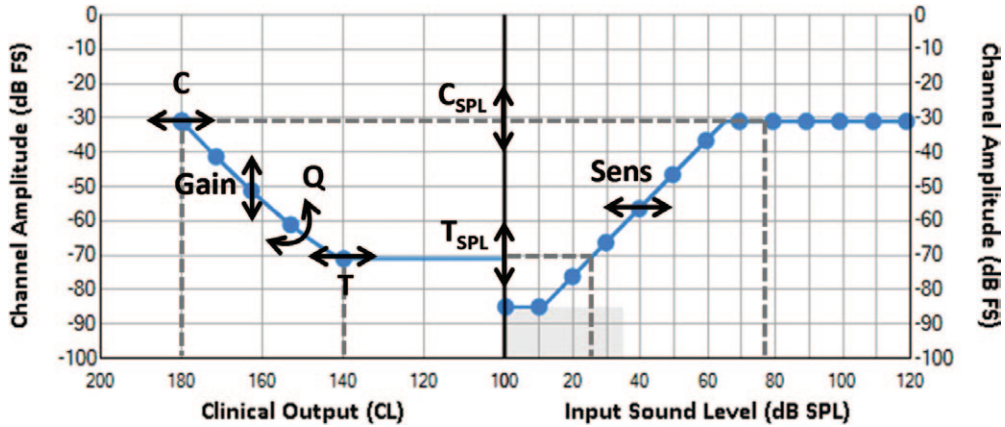
Fig. 7. Intensity-coding function plots for Cochlear devices in response to a pure tone, showing preprocessing impacted by Sens at the right and mapping impacted by T-SPL, C-SPL, T, C, Gain, and Loudness Growth (Q) at the left. Straight arrows depict translations; curved arrows indicate changes in the shape of the curve. C indicates maximum comfort; Sens, sensitivity; T, threshold.

msec). The AGC acts to prevent clipping by its fast attack and slower decay.

Channel-specific gains are applied to the channel input (i.e., filter bank output). They are either determined by the static clinical gains specified in the MAP or by Adaptive Dynamic Range Optimization. Clinical Gains range from −12 to +10 dB and have the effect of amplifying the channel's input with a fixed value, but they are ignored when Adaptive Dynamic Range Optimization is enabled.

Stimulation above C-level can never occur, even if clinical gains of +10 dB are set. When applying positive gain to a channel, saturation (yielding C-level stimulation) occurs for that channel at an input level of C-SPL minus Gain, for example, 55 dB SPL when C-SPL is 65 and channel gain is +10. When applying a negative gain, a channel will never reach C-level stimulation. Rather, saturation (maximal stimulation level) occurs below C-level as the input presented to the channel is already limited (to C-SPL) by the fast-acting AGC and subsequently attenuated by applying a negative channel gain.

Amplitudes below T-SPL do not produce any stimulation. Indeed signals below the T-SPL are effectively ignored. Setting T-SPL below 25 dB may result in excessive electrical (system) noise being perceived constantly.

Increasing C-SPL (even more so when combined with an increase in C levels) may cause a different psychophysical loudness perception. To compensate for this, the Custom Sound software automatically adjusts the Loudness Growth (Q) parameter. The Q parameter sets the percentage of dynamic range (EDR) that is allocated to the top 10 dB of the input being mapped (IMR). A lower Q value is more compressive and curves the function more. According to the manufacturer, Q is not meant to be seen as compression, and Cochlear does not advise making major changes to Q.

The pulse width is not considered in the current level (CL) unit used in the fitting software. As a consequence, doubling the pulse width while keeping the same T and C levels will cause the channel to deliver twice the amount of charge (per pulse phase) to the electrode. The accuracy of the pulse amplitude is limited to the integer values between T and C. As a consequence, in a channel with T = 140 and C = 170, every channel amplitude is mapped to 1 of the 31 distinct pulse amplitudes available between 140 and 170.

**MED-EL**

Figure 8 shows the ICF plot for MED-EL with the impact of its MAP parameters. MED-EL's AGC system is a dual-loop
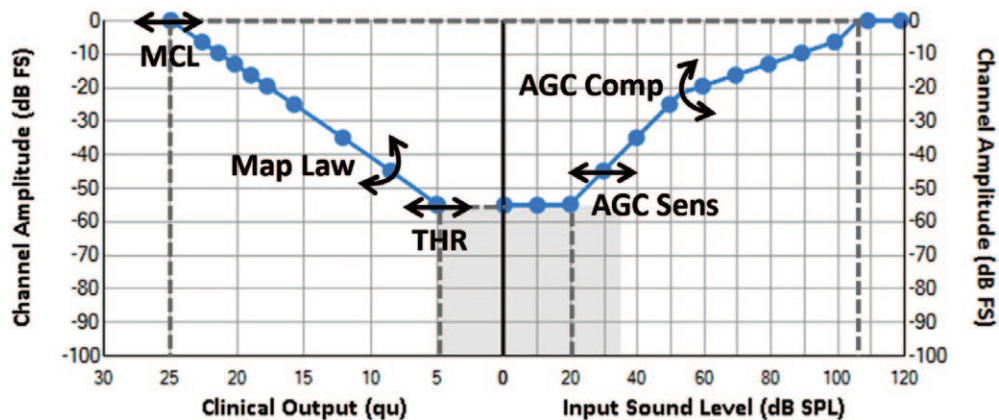


Fig. 8. Intensity-coding function plots for MED-EL devices in response to a pure tone, showing preprocessing impacted by AGC Sens and Comp on the right and mapping impacted by THR, MCL, Gain, and Map Law compression on the left. Straight arrows depict translations; curved arrows indicate changes in the shape of the curve. AGC indicates automatic gain control; Comp, compression ratio; MCL, most comfortable level; Sens, sensitivity; THR, threshold.

AGC (Stöbich 1999). The fast and slow detectors work in parallel on the same input signal. The resulting outputs of the two detectors are weighted to determine the corresponding gain. With default compression of 3:1 and sensitivity of 75%, the static gain of the dual-loop AGC has its knee-point at 52.7 dB SPL. The fast detector has 4-msec attack and 16-msec release time; the slow detector has 100-msec attack and 400-msec release time. The knee-point is shifted (+14 dB to −5 dB with respect to the default value) when changing the AGC Sensitivity parameter. The AGC system keeps the broadband input signal within range. The AGC fast detector acts to prevent clipping by its fast attack and slower decay. Input of 106 dB SPL causes stimulation at most comfortable level (MCL). The upper limit for signal level is 100 dB SPL with 6 dB headroom available to allow for rapid fluctuations (peaks), see Appendix B for ICFs showing rapid fluctuations (see Supplemental Digital Content, http://links.lww.com/EANDH/A140).

MED-EL uses the term Adaptive Sound Window for their 55 dB IMR. The position of this window is managed by the AGC system. By moving the IMR in response to the input signal's intensity MED-EL obtains an IDR of about 75 dB. Threshold (THR) and MCL levels determine the range and position of the EDR. Stimulation above MCL level can never occur. In the commonly used Innsbruck (IBK) volume mode, no stimulation below THR can occur for any volume setting (0 to 100%) and THR is the minimum stimulation level for any enabled channel, meaning that the system will deliver charge to all enabled channels that have a THR greater than 0, even if there is no input signal present.

The Maplaw Compression parameter defines the loudness growth function to map channel (envelope) amplitudes into the EDR. A higher compression value assigns a larger portion of the EDR to softer sounds. A lower compression assigns a larger portion of the EDR to louder sounds.

The amplitude of a stimulation pulse is given in current units (CU). One current unit is approximately 1 µA. The charge of one phase is defined as the product of stimulation current in CU and pulse phase duration, which is displayed in microseconds in the fitting software. One charge unit (QU) is approximately 1 nC. The MED-EL fitting software is charge-based so that Pulse Amplitude (CU) cannot be manually adjusted. Rather, Pulse Charge (QU) is set in the fitting software. Doubling the phase duration (e.g., by increasing minute Duration) while keeping

the same THR and MCL (QU) levels, will not cause the channel to deliver more charge (per pulse phase) to the electrode, instead the pulse amplitude (CU) is decreased automatically.

**Advanced Bionics**

Figure 9 shows the ICF plot for AB with the impact of its MAP parameters. Sensitivity is applied as a constant gain to the broadband input, before the signal is processed by the AGC. AB's AGC system also features two detectors (fast and slow). The resulting outputs of the two detectors are compared to determine the corresponding gain. The slow detector has a long attack time, meaning that the system works as a volume control that keeps the input level within range. It features a compression factor of 12:1, a 240-msec attack time and 1500-msec release time, and a broadband threshold of about 63 dB SPL. The fast detector has a threshold of 71 dB SPL and is used to prevent clipping. AB's upper-limit signal level is 97 dB SPL.

The position of the IMR window in AB devices is managed by the AGC system. The IDR parameter determines the size of the IMR window. The T level and most comfortable level (M) determine the range and position of the EDR. Stimulation above M level is possible. When AGC is disabled, a 5000 Hz pure-tone input signal of just over 60 dB SPL causes stimulation at M level. Increasing the input level further causes an increase in clinical units (CU) beyond M level. Similarly, stimulation also continues linearly below T Level. Setting a large IDR value or high Sensitivity in combination with correct (measured) T levels may result in electrical (system) noise being perceived constantly. This relates to the microphone noise as described earlier in the article, which is then mapped into the EDR.

AB does not provide a parameter to change the shape of the mapping function. It is, however, possible to limit a channel's output by setting a clipping level for the channel. This prevents stimulation from exceeding the chosen level. Other than that, a channel's (envelope) amplitude expressed in dB is always mapped linearly into the EDR (i.e., an increase of 1 dB causes a fixed increase in CU). The CU used in the fitting software incorporates both Pulse Amplitude and Duration and is therefore proportional to charge (Coulomb). Pulse Amplitude (µA) cannot be manually adjusted using the fitting software. Doubling the pulse width while keeping the same T and M levels, will not cause the channel to deliver more charge (per pulse phase) to
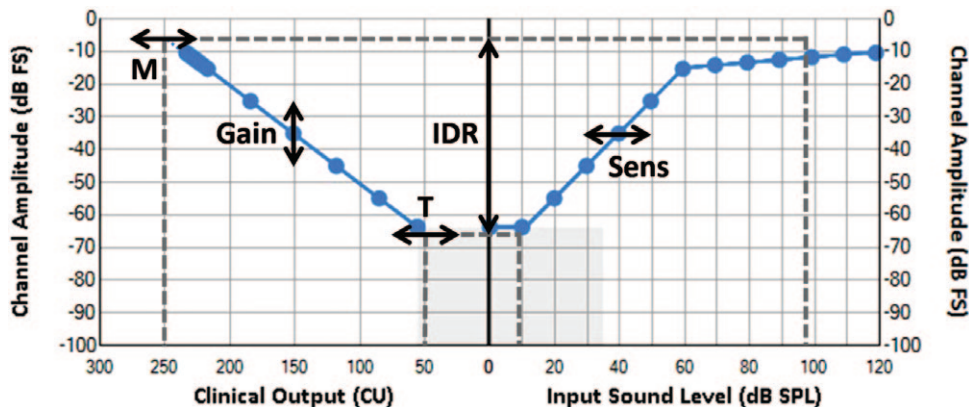


Fig. 9. Intensity-coding function plots for Advanced Bionics devices in response to a pure tone, showing preprocessing impacted by the Sens parameter on the right and mapping impacted by T, M, Gain, and IDR parameters on the left. IDR indicates input dynamic range; M, most comfortable level; T, threshold; Sens, sensitivity.
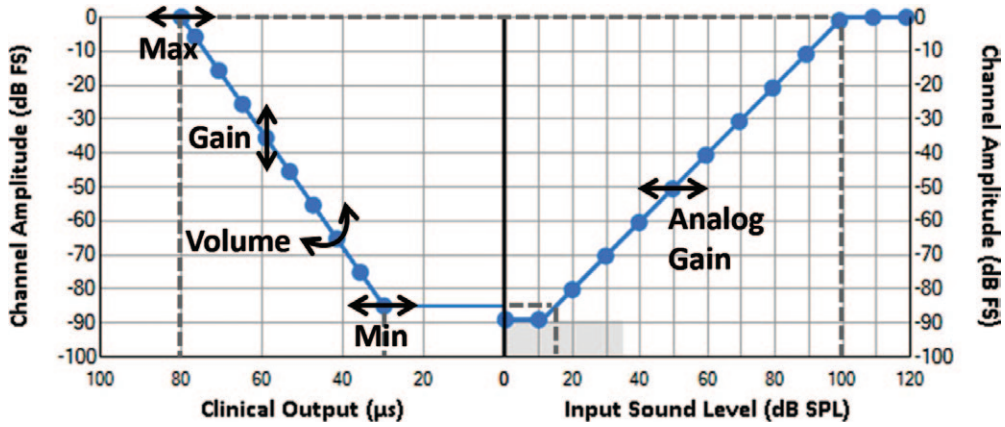
Fig. 10. Intensity-coding function plots for Neurelec devices in response to a pure tone, showing preprocessing impacted by the Analog Gain parameter at the right and mapping impacted by Min, Max, Gain, and Volume parameters at the left. Straight arrows depict translations, curved arrows indicate changes in the shape of the curve. FS indicates full scale.

the electrode, instead the pulse amplitude (µA) is decreased automatically.

### Neurelec

Figure 10 shows the ICF plot for Neurelec with the impact of its MAP parameters. The Neurelec system has no automatic gain or sensitivity control. The implant recipient has the option to adjust the sensitivity manually. With an Analog Gain of 0 dB, peaks above 100 dB SPL are clipped by the A/D converter. Neurelec uses a fixed IMR of 85 dB. Min and Max levels determine the range and position of the EDR. Signals of 100 dB SPL are mapped to Max Level, 15 dB SPL corresponds to Min Level.

Stimulation above Max Level can never occur, even if the Volume parameter is at maximum. Channel amplitudes below 15 dB SPL do not produce any stimulation. Indeed signals below this level are effectively ignored.

Neurelec allows setting gains (ranging from 0 to −10 dB) on each of the 63 bands of the fast Fourier transform (FFT) output. These gains are applied before the combination into channels and may therefore affect the channels that are selected for stimulation (Maxima Selection). The Volume parameter defines the loudness growth of the mapping function. A higher Volume value assigns a larger portion of the EDR to softer sounds. A lower Volume assigns a larger portion of the EDR to louder
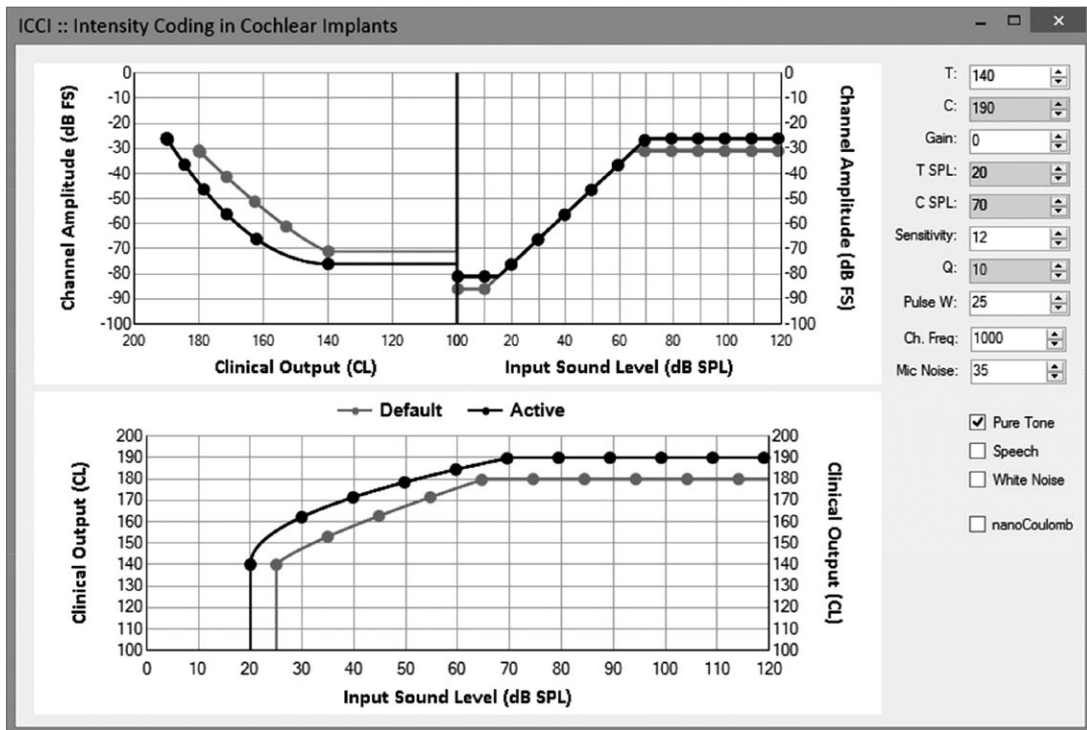


Fig. 11. A screenshot of the intensity coding in cochlear implants application showing the intensity-coding function for Cochlear's device. A default map where T equals 140 CL and C equals 180 CL is plotted in gray. A map where C is changed to 190 CL, T-SPL to 20, C-SPL to 70, and Q to 10 is plotted in black. The top charts show the separate preprocessing and mapping functions, The bottom chart is the result of merging the top charts into a single transformation of acoustical input level into electrical output level. C indicates maximum comfort; FS, full scale; T, threshold.

sounds. Neurelec keeps the pulse amplitude fixed and adjusts the pulse width to code for loudness. The clinical unit for Min and Max levels, displayed in the fitting software, therefore relates to a dimension of time/duration. The pulse amplitude is not considered in the clinical unit used in the fitting software. As a consequence, doubling the pulse amplitude (Amplitude parameter) while keeping the same Min and Max levels, will cause the channel to deliver twice the amount of charge (per pulse phase) to the electrode.

### Interactive Plots in the ICCI Application

An interactive application called ICCI that allows plotting the ICFs for the four brands is available on request from the first author B. Vaerenberg. A Web-based version is also available at http://www.otoconsult.com/fitting/icci. As shown in Figure 11, the application allows the user to adjust the fitting parameters and view the resulting changes on the plots. As explained previously, the top plots depict both transformations from acoustical level into digital level (preprocessing plus filter bank) and from digital level into electrical level (mapping). In addition, the result of merging these processes into a single transformation from acoustical input level into electrical output level is depicted by the application in its bottom graph. Static plots illustrating the effect of each parameter are included in Appendix A (see Supplemental Digital Content http://links.lww.com/EANDH/A140).

### DISCUSSION

The fitting of CIs is a technical procedure, which requires a thorough insight into these systems' sound processing. With the ever-increasing complexity of the underlying technology and given the fact that many CI centers have offered and programmed several CI brands to date, the professional fitter is facing a greater than ever challenge when trying to fully predict the impact of parameter changes on the implant's behavior. One way to cope with this increasing complexity is to simplify the act of fitting by limiting the number of parameters to adjust and by adopting approximations and rules of thumb to make global profile optimizations. There are indications, however, that addressing more of the many parameters available may lead to better outcome in specific, if not most cases (Vandali 2000; Friesen 2001; Zeng 2002). For this to be done in a knowledgeable manner, the level of understanding and predicting these devices' behavior is certainly less than what engineers need when designing them, but is likely to be more than what is commonly available.

The effect of parameter changes is explained in clinical guidance documents, but very often this information is fragmented, limited to a single parameter at a time and not integrated in the complete input–output behavior of the system. In addition, every manufacturer has its own way of presenting its system's behavior and uses proprietary names for parameters that basically do the same thing (e.g., Input Sensitivity/Gain, IDR/IIDR/Adaptive Sound Window). This increases the load on clinicians to understand the behavior of the CI systems they are programming on a daily basis.

When the behaviors of these different CI systems are synthesized into a uniform graphical representation, the specific features of a particular CI system become accessible to clinicians in a more transparent way, which in turn may assist them in the programming of these systems. Using the ICCI application, we have come to a number of remarkable observations. For example, it is clear that manufacturers handle the compression of 90 dB of input range into a couple of dBs of electrical range quite differently. If we consider the default settings for each of the devices, then we see that IMRs of 40, 55, 60, and 85 dB are chosen by Cochlear, MED-EL, AB, and Neurelec, respectively. It
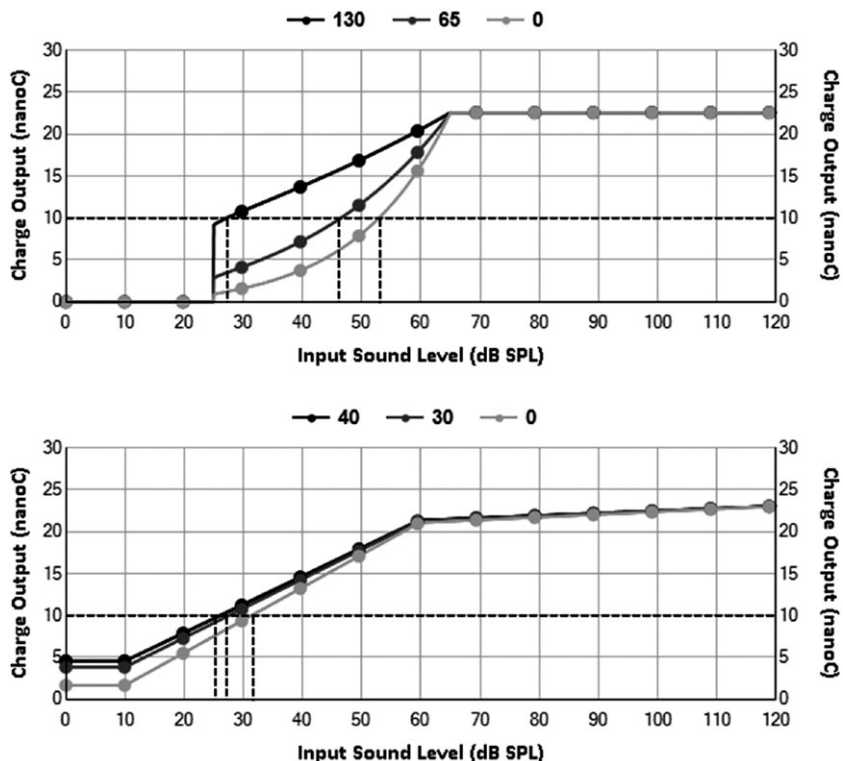


Fig. 12. The effect of setting T, assuming 10 nC is the recipient's detection threshold. In Cochlear (upper graph) a measured value of 130 CL using a 50-μsec Pulse Width results in an audiometric threshold of 27 dB. Setting T to half of that or to 0 CL increases the audiometric threshold to 46 and 53 dB, respectively. In Advanced Bionics (lower graph) the audiometric threshold is far more stable. A measured T of 40 CU results in a audiometric threshold of 25 dB. Setting T to 10% (30 CU) of M or 0 CU increases the audiometric threshold only by 2 and 7 dB, respectively. T indicates threshold.
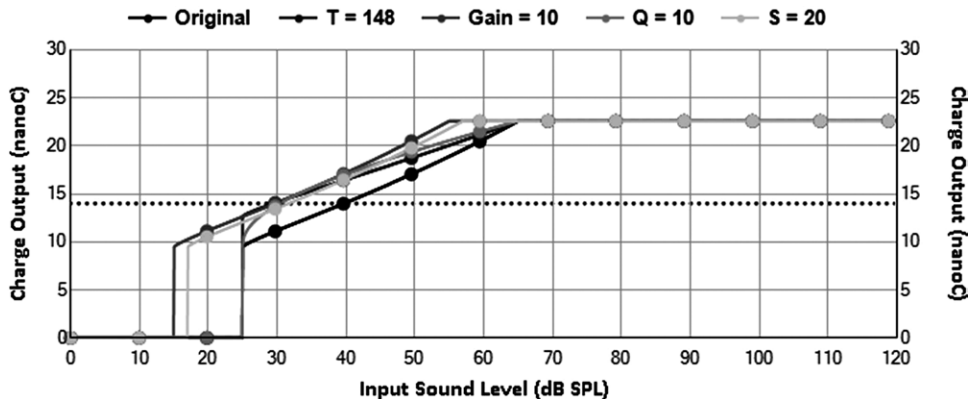
Fig. 13. Improving an audiometric threshold from 40 dB to 30 dB can be done in several ways. In Cochlear's system, changing T from 132 to 148 CL, Gain from 0 to 10 dB, Q from 20 to 10, or Sensitivity from 12 to 20 all have approximately the same effect on the detection threshold. All these manipulations, however, do have different effects on other properties of the mapping function. Q indicates loudness growth; T, threshold.

may be that Neurelec, in the absence of an AGC function, opted for such a large IDR to cover the diverse listening situations in daily life. The other brands have automated gain controls, allowing them to maximize the intensity resolution at any given environmental noise level. The drawback of such an approach may be that the loudness growth in the upper intensity range is limited to some extent, which introduces a phenomenon that is not known to be present in a normal auditory system.

Another interesting observation is that the default mapping functions in all brands are more or less linear when considering the channel input amplitude expressed in dB and the channel output level expressed in nC. While the other manufacturers use a clinical unit that linearly relates to charge output, Cochlear uses a clinical unit (CL) that relates to charge (nC) exponentially (an increase of 1 unit in CL has the effect of increasing the current amplitude by approximately 2%). However, the plots show that Cochlear uses a mapping function that compensates for this. It does not map channel amplitude dBs to CL entirely linearly. In a typical map, where Q is 20, the ICF is slightly curved. But when converted to charge (nC) the function becomes approximately linear again. Altogether, their mapping is not very different from that of other manufacturers, when default parameters are used. So, the differences in output compression techniques between manufacturers are more attributable to limiting IMR and using automatic gain controllers than they are to compressive mapping functions.

Although this information is not readily available, we assume that the microphone noises, and therefore the noise floors of all systems are rather similar. It should not be a surprise to learn that all brands use similar microphones on their CI processors. The fact that all systems saturate around 100dB SPL (when configured for maximum input range) also has to do with all brands facing the same technological limitations in analog to digital conversion and sampling range.

AB is the only brand in which stimulation continues below T Level by default. Cochlear and Neurelec implants cease stimulation when the input to a channel falls below the IMR. MED-EL keeps stimulating at THR. This implies that in AB, from a technical point of view, the IDR and T parameters are mutually redundant (e.g., one could achieve the very same effect of adjusting IDR, by adjusting T).

For setting T levels, the four brands use two distinct approaches: MED-EL and AB would not advise against setting THR/T at zero or at a fixed fraction of MCL/M; Cochlear and Neurelec recommend measuring T levels for each CI recipient. Indeed, the plots show that in Cochlear, setting T at zero would increase the audibility threshold dramatically, as illustrated in Figure 12. In MED-EL and AB, this effect is considerably smaller, due to the nature of their mapping function.

The plots can be used to give an indication of how to resolve issues related to fitting. For example, if audiometric thresholds are higher (worse) than target, the plots assist in identifying the MAP parameters and the direction and magnitude by which they can be adjusted to improve this outcome. A typical intervention might be to increase the EDR Minimum. Taking Cochlear's device as an example, the plots show that this is indeed effective. It is, however, not the only way to reach this goal. As illustrated in Figure 13, the audibility threshold may also be improved by either increasing Sensitivity or Gain, decreasing Q, or a combination of those adjustments. Decreasing T-SPL, however, does not improve the audibility very much.

The same manipulations may also be used to increase speech perception at low intensities (<50 dB SPL). To improve perception of loud speech, when, for example, a rollover effect (i.e., a decrease in speech intelligibility at higher presentation levels) is observed, one may adjust parameters related to AGC compression, or decrease EDR Maximum levels. If in the context of loudness scaling or other psychoacoustic measures one aims at improving the difference limen of intensity, decreasing the IMR to maximize the mapping resolution could be a first approach. In any case, all these adjustments are very likely to not only have the intended effect, but may also induce side effects on other aspects of the intensity coding. The plots are instrumental in uncovering these side effects, so that one may choose the adjustment that is most effective in resolving an issue without compromising other important requirements for an optimal fitting.

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