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Comparisons between detection threshold and loudness perception for individual cochlear implant channels

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Abstract

Objective—The objective of the present study, performed in cochlear implant listeners, was to examine how the level of current required to detect single-channel electrical pulse trains relates to loudness perception on the same channel. The working hypothesis was that channels with relatively high thresholds, when measured with a focused current pattern, interface poorly to the auditory nerve. For such channels a smaller dynamic range between perceptual threshold and the most comfortable loudness would result, in part, from a greater sensitivity to changes in electrical field spread compared to low-threshold channels. The narrower range of comfortable listening levels may have important implications for speech perception.

Design—Data were collected from eight, adult cochlear implant listeners implanted with the HiRes90k cochlear implant (Advanced Bionics Corp.). The partial tripolar (pTP) electrode configuration, consisting of one intracochlear active electrode, two flanking electrodes carrying a fraction (σ) of the return current, and an extracochlear ground, was used for stimulation. Single-channel detection thresholds and most comfortable listening levels were acquired using the most focused pTP configuration possible ($\sigma = 0.8$) to identify three channels for further testing – those with the highest, median, and lowest thresholds – for each subject. Threshold, equal-loudness contours (at 50% of the monopolar dynamic range), and loudness growth functions were measured for each of these three test channels using various partial tripolar fractions.

Results—For all test channels, thresholds increased as the electrode configuration became more focused. The rate of increase with the focusing parameter σ was greatest for the high-threshold channel compared to the median- and low-threshold channels. The 50% equal-loudness contours exhibited similar rates of increase in level across test channels and subjects. Additionally, test channels with the highest thresholds had the narrowest dynamic ranges (for $\sigma = 0.5$) and steepest growth of loudness functions for all electrode configurations.

Conclusions—Together with previous studies using focused stimulation, the results suggest that auditory responses to electrical stimuli at both threshold and suprathreshold current levels are not uniform across the electrode array of individual cochlear implant listeners. Specifically, the steeper growth of loudness and thus smaller dynamic ranges observed for high-threshold channels are consistent with a degraded electrode-neuron interface, which could stem from lower numbers

of functioning auditory neurons or a relatively large distance between the neurons and electrodes. These findings may have potential implications for how stimulation levels are set during the clinical mapping procedure, particularly for speech-processing strategies that use focused electrical fields.

Keywords

Cochlear implant; loudness; dynamic range; electrode configuration

Introduction

Cochlear implants are highly successful neural prostheses for individuals with severe-to-profound hearing loss. However, speech perception is highly variable across individuals, and performance can be markedly reduced in the presence of background noise. A number of studies suggest that some of this variability reflects the particular capabilities of cochlear implant listeners to make use of intensity cues in speech. For example, Fu and Shannon (2000) found that implant listeners' ability to discriminate small changes in intensity was correlated with speech recognition scores, particularly in realistic listening environments with background noise. Consistent with those findings, artificial compression of the electrical dynamic range has been shown to have a significant negative impact on speech recognition, especially for vowels and phonemes in the presence of background noise (Zeng & Galvin, 1999; Loizou et al., 2000). However, while a few studies have compared average dynamic ranges and speech performance among individuals or groups of implant listeners (e.g. Blamey et al., 1992; Bento et al., 2005; D'Elia et al., 2012), it remains unclear how the processing of intensity cues of individual electrode channels may influence speech recognition.

One important factor to consider in relating cochlear implant performance to single-channel measures of intensity processing is the degree of variability among channels. For instance, a considerable amount of cross-channel variability has been observed for temporal modulation sensitivity (e.g. Chatterjee and Yu, 2010; Garadat et al, 2012) and the slopes of loudness growth functions (e.g., Chatterjee and Shannon, 2000; Cohen, 2009a). At the low end of intensity processing, detection thresholds can also be variable, particularly when measured with a focused electrode configuration such as bipolar, tripolar, or phased-array (Pfingst and Xu., 2004; Bierer, 2007; Long et al., 2014; Chua et al, 2011). Interestingly, these latter studies have shown that a greater degree of channel-to-channel threshold variability within subjects predicts poorer performance in speech perception. Although the reason for this correlation is not well understood, there is some evidence that high-threshold channels have a diminished capacity to transmit spectral, temporal and loudness cues. In Bierer and Faulkner (2010), for example, psychophysical tuning curves of channels with relatively high thresholds (measured with the focused tripolar electrode configuration) were significantly broader than those obtained with low-threshold channels. Such a pattern of activation implies a reduction in the transmission of spectral/spatial information because neurons responding to a broadly-tuned channel can be more easily stimulated by neighboring channels, which may negatively affect speech perception. Additionally, a study by Chatterjee and Yu (2010), in which the ability of listeners to discriminate two electrodes was

compared to the sensitivity of temporal modulations in amplitude, suggests that the same conditions that lead to poor spatial discrimination have an adverse effect on the processing of temporal and intensity information. It follows that a better understanding of the perceptual characteristics of individual channels could further our understanding of speech perception by cochlear implant listeners.

One possible explanation for the observed variability of perceptual characteristics across a cochlear implant array – and among subjects – is that the status of the electrode-neuron interface is different from channel to channel. The electrode-neuron interface relates to the effectiveness by which an implant electrode activates the auditory nerve. Several factors can affect the interface, including the viability of the spiral ganglion cell bodies and their processes, the distance between the electrode and the inner wall of the cochlea where the ganglion cells reside, and the degree of bone and tissue growth surrounding the electrode array (see Bierer, 2010 and Pfingst et al, 2011 for reviews). A recent computational model of cochlear electrical stimulation suggests that focused electrode configurations are especially sensitive to factors that degrade the electrode-neuron interface (Goldwyn et al., 2010). Consistent with this prediction, the Bierer and Faulkner study (2010; see also Bierer, 2007) demonstrated that both thresholds and psychophysical tuning curves were more variable from channel to channel when measured with a focused partial tripolar configuration than with the monopolar configuration.

In the current study, channels classified by tripolar threshold into “low-”, “median-”, and “high-threshold” channels were examined in the context of intensity processing, specifically loudness growth and dynamic range. The results of the Bierer and Faulkner (2010) study suggest that when elevated current levels are required to reach threshold, the neurons contributing to the response are spread over a relatively wide extent of the cochlea. In that study, the difference in the sharpness of tuning between low- and high-threshold channels persisted for monopolar probe stimuli, which had similar stimulus levels and dynamic ranges across channels. Furthermore, the sharpness of tuning was relatively unaffected by probe level, similar to observations of Nelson and colleagues (2008; see also review by McKay, 2012). Together, these findings suggest that the differences between low- and high-threshold channels is an intrinsic property of their electrode-neuron interfaces, rather than the specific stimulus parameters used to obtain them (see McKay, 2012 and Fielden et al., 2013). If such perceptual differences occur with relatively low level sensation levels, as used in these previous studies, an important question is what happens at current levels well above threshold. The present study examines this question by comparing loudness growth and dynamic range for channels with low and high tripolar thresholds.

One motivation for the present study design comes from a recent electrophysiological study that used a similar focused threshold paradigm to classify channels (Bierer et al, 2011). In that work, amplitude growth functions of electrically-evoked auditory brainstem responses (EABRs) were found to be steeper for channels having a high tripolar threshold. The apparently faster neural recruitment with increasing current levels for high-threshold channels is consistent with a broader extent of neural activation across the cochlea. If loudness reflects a similar pattern of neural recruitment as the EABR growth functions, then

steeper loudness growth and smaller dynamic ranges would be expected for channels with higher tripolar thresholds.

Methods

Subjects

Eight postlingually deafened adults who are native speakers of American English participated in the study. All subjects wore an Advanced Bionics HiRes90K implant (Advanced Bionics Corp., Sylmar, CA) and had at least 6 months experience with the device prior to their participation. Details about the individual subjects are listed in Table I. Each participant provided written consent, and the experiments were conducted in accordance with guidelines set by the Institutional Review Board at the University of Washington.

Stimuli

Stimuli consisted of trains of biphasic, charge-balanced pulses (cathodic phase first on the active electrode) with 102 μ s per phase presented at a rate of 918 pulses/second and 200 ms duration. Channel numbers denote the position of the active electrode on the implant array, and for this experiment ranged from 2 (apical) to 15 (basal). All channels were stimulated in the partial tripolar (pTP) electrode configuration with various current fractions. As described previously (Jolly et al., 1995; Litvak et al., 2007; Bierer, 2010), the pTP configuration consists of an intracochlear active electrode, with a fraction of the return current divided equally between each of the two nearest flanking electrodes and the remaining current directed to an extracochlear ground electrode. When the current fraction, denoted by σ , is 0, all current is directed to the distant return electrode, which is equivalent to the monopolar (MP) configuration. When σ is 1, all return current from the active electrode is directed to the flanking electrodes, which corresponds to the focused tripolar (TP) configuration. (In this paper, MP and TP channels are often referred to as pTP channels with fractions of 0 and 1, respectively.) Values of σ between 0 and 1 give progressively more focused current and electrical field patterns.

Pulse trains were delivered to the implant using a personal computer connected to a clinical interface and dedicated Platinum Series Processor. The stimuli were controlled through the Bionic Ear Data Collection System (BEDCS version 1.15.158; Advanced Bionics Corp., Sylmar, CA) and a custom MATLAB graphical user interface (Mathworks, Natick, MA). The same computer was used to obtain psychophysical responses. Based on the clinically measured impedance values and voltage compliance limit, the maximum allowable current for each electrode was calculated at the beginning of every test session (Advanced Bionics, personal communication).

Subjects were asked to follow visual prompts on the computer monitor and respond appropriately using a computer mouse. The subjects were allowed to practice new tasks until they could perform them properly. They did not receive trial-by-trial feedback. Data analysis and plotting was mainly performed in units of decibels (re 1 μ A) to accommodate the large differences in current between configurations and the highly variable dynamic ranges across channels and subjects.

Single-Channel Thresholds, Most Comfortable Levels and Dynamic Ranges

For each subject, thresholds were measured using the largest common pTP fraction for which a threshold could be obtained on all fourteen channels. (We refer to channels tested in this fashion as tripolar, because the configuration is the most focused for a given subject; values of σ ranged from 0.8 to 1.) The channels with the lowest, median, and highest thresholds were identified for further testing. Additional thresholds were measured for this subset of channels for pTP fractions of $\sigma = 0$, $\sigma = 0.5$, $\sigma = 0.62$ or 0.68 , and 0.75 . All thresholds were measured with an adaptive two-down/one-up, three-interval, three-alternative forced-choice procedure, which converged on the 70.7 percent correct point on the psychometric function (Levitt, 1971). For each threshold, twelve reversals were obtained and the average was based on the last 8 reversals. For the first two reversals, 2-dB steps were used, and for the remaining 8, 0.5-dB steps were used. Two series of reversals were completed for each threshold. Time limitations resulted in missing data points for S31 for the middle sigma value.

Most comfortable level (MCL) was measured for each test channel and pTP configuration by presenting a soft supra-threshold level pulse train and asking the subject to manually adjust the level up or down until he or she reached the subjective rating of “loud but comfortable,” corresponding to 7 on the 1 to 10 clinical loudness rating scale. Dynamic range was calculated as the difference between MCL and threshold in decibels. For some channels, MCL could not be reached because the voltage compliance limit was reached prior to the subject reporting a “loud but comfortable level”. For further details regarding threshold and MCL measures see Bierer and Faulkner (2010).

Growth of Loudness

Loudness growth functions were measured in each of the above-identified channels. If the loudness rating for a particular channel was not at least “5” described as “Medium”, then the pTP fraction was not tested for that channel. Eight stimulus levels, equally spaced on a decibel scale and ranging from threshold to 10% above MCL (if 10% above MCL was within the compliance limit, otherwise the additional 10% was not used), were presented in successive trials in pseudorandom order. Each loudness judgment was based on 6 sequential presentations of the stimulus pulse train, and an average loudness estimate was calculated from eight such trials. Subjects rated loudness on a scale from 0 (“can’t hear it”) to 100 (“too loud”) using a graphical slider interface. The loudness estimations were analyzed using a logarithmic scale. Because a fixed-end loudness scale could distort loudness estimates near the end points, loudness ratings less than 5 and greater than 90 were not included in any of the analyses.

Equal-Loudness Contours

For each of the three test channels, stimulus levels were balanced across pTP configurations to 50% of the MP dynamic range in decibels using a two-interval, forced-choice, double-staircase procedure (Jesteadt, 1980). One of the two intervals was a reference stimulus, presented in the MP configuration fixed at the 50% level; the other was presented in one of the pTP configurations at a variable level. Subjects were asked which interval sounded louder. In two tracks, starting from one high and one low current, level was changed in 2 dB

steps for the first two reversals and 0.5 dB for the remaining four reversals. Balanced levels were calculated by averaging the last four reversals from both tracks. At least two runs were collected and averaged for each condition; if the two estimates differed by more than 1 dB, a third run was completed, and all three runs were averaged.

The data presented in this paper were obtained over multiple sessions. Typically, the 14 tripolar thresholds were obtained first to determine the low, median and high threshold channels. These threshold profiles took approximately 3 or 4 hours to obtain. In the following three or four sessions, the remaining tasks (measuring most comfortable levels, loudness balancing, loudness estimations and thresholds with monopolar and other pTP configurations) were conducted in variable order.

Results

Figure 1 plots detection thresholds for each subject as a function of cochlear implant channel from apex to base. Each panel displays data for one subject and the electrode configuration is indicated by the symbol: monopolar ($\sigma = 0$, circle), partial tripolar ($\sigma = 0.5$, square), and tripolar ($\sigma = 0.8$, triangle). Tripolar channels with the lowest, median and highest thresholds for each subject are indicated by the vertical dashed lines and fill of the symbol. These channels were subsequently used for the loudness experiments. Note that the cochlear position of the lowest and highest threshold channels varied among subjects. (The threshold data for S22, S23, and S26 are the same as in Bierer and Faulkner (2010). For S23, two low-threshold channels were chosen, one apical and one basal.)

Thresholds (dB re 1 μ A) for the three test channels identified with TP stimulation are plotted in Figure 2 as a function of partial-tripolar fraction (σ). Best-fit lines minimizing the squared error are shown by dashed lines for each channel. As demonstrated in Fig. 1, threshold generally increased with increasing pTP fraction, consistent with the results of Bierer and Faulkner (2010) and of Litvak and colleagues (2007). These trends were monotonic for the high-threshold channels, but for some subjects became saturated or even decreased at higher currents for the median- and low-threshold channels. Similar to the findings of Bierer and Faulkner (2010), for four subjects the threshold versus fraction slope was greatest for the channel with the highest threshold (S22, S23, S26, S29). In contrast, for the remaining subjects the slopes were similar among channels (S27, S28, S30, and S31). Despite this intersubject variability, slopes were significantly steeper for channels with the highest TP thresholds when all eight subjects were considered. A repeated measures analysis of variance (ANOVA) with a Greenhouse-Geisser correction (epsilon = 0.538, Mauchly's $W = 0.142$, $p = 0.003$) indicated a main effect of tripolar threshold ($F_{1,076, 7.533} = 6.7$, $p = 0.032$). A Tukey-Kramer multiple comparison test revealed that low and median slopes were significantly lower than high-threshold channel slopes ($p < 0.05$).

In Figure 3, stimulation levels perceived as equivalently loud to a monopolar stimulus presented at 50% of its dynamic range are plotted in the same manner as Fig. 2. Steeper functions mean that for more focused configurations greater current is needed for the subject to perceive the stimulus as equally loud to the monopolar reference. As with the threshold data, with increasing pTP fraction higher current was required to maintain the same

subjective level of loudness. Unlike the threshold data, however, the current versus fraction trends were monotonic for all channels and all subjects. Equal-loudness contours were steepest for the channels with the highest TP threshold for four of the subjects (S22, S23, S28, S29). The similar patterns for the threshold (Fig. 2) and loudness balanced data (Fig. 3) suggest there are two variants of high- versus low-threshold channel behavior: those with parallel and those with divergent dependence on σ . Across subjects there was a significant trend that the high-threshold channels have steeper slopes (repeated measures ANOVA, $F_{2,14} = 3.81$, $p = 0.048$; for this analysis the sphericity assumption was met, $p = 0.197$). A post hoc Scheffe's multiple comparison test revealed that the high-threshold channel contours were significantly steeper than the median and low-threshold channels ($p < 0.05$). No other significant differences were found between low- and median-threshold contours.

Dynamic range as a function of TP threshold is plotted in Figure 4. Within a panel, each subject is represented by a different symbol. The top row of panels displays the absolute values of threshold (abscissa) and dynamic range (ordinate), while the bottom row plots the values relative to each subject's cross-channel average. Note that the normalized data points corresponding to the median-threshold channel always occur in between the low- and high-threshold data points along the abscissa and are usually closest to zero. Also, the largest pTP fraction used for dynamic range measurements was 0.75, while the TP threshold that defined channel type was obtained with a fraction of at least 0.8 (i.e., TP threshold was not a determinant in the dynamic range calculation). Grey filled symbols in the right-most panels indicate that MCL (a loudness rating of 7) could not be reached due to compliance limits, and instead the highest allowable current was used; for these cases the subject's loudness rating was at least a 5.5. For S26, data for one additional focused configuration was obtained and included in the analyses. A statistically significant negative correlation between dynamic range and TP threshold was found for the pTP configurations but not for MP (Pearsons correlation coefficients are shown in each panel). To further investigate the relation between TP threshold and dynamic range, a multiple linear regression analysis across pTP fractions was performed. The results indicated an adjusted $R^2 = 0.372$ ($F_{3,21}=5.737$, $p=0.005$). The same analysis for the normalized data (lower panels) showed an adjusted $R^2 = 0.378$ ($F_{3,21}=4.55$, $p=0.005$).

Figure 5 plots examples of single-channel loudness estimations as a function of stimulus level (dB re 1 μ A) for one subject, S22, both for a linear (top row) and logarithmic (bottom row) loudness estimation scale (ordinates). All subsequent analyses are based on the log-loudness scale. Increasing pTP fraction is shown in panels from left to right. Best-fit lines (dashes) were fit to the log of the loudness growth functions to minimize the mean squared error. Slopes for the functions were obtained from the best-fit lines. The loudness growth functions of S22, for whom the thresholds and loudness contours are of the divergent type (see Figs. 2 and 3), exhibit a large degree of separation. While this separation is most apparent for the focused configuration (right panels), the high-threshold channel has steeper slopes than the low- and median-threshold channels for all four configurations.

The relation between configuration, TP threshold, and the slope of the loudness functions is illustrated for all subjects in Figure 6. Each panel represents data for a different electrode configuration. The top row of panels plot the log-loudness slope (log estimate/dB) as a

function of TP threshold, and the lower panels plot slope values relative to the cross-channel average for each subject and configuration as a function of relative TP threshold (dB). For both the absolute and the relative measures, a significant positive correlation was observed such that higher TP thresholds predicted steeper loudness growth functions for all tested configurations (Pearson's correlation analyses are shown in each panel). This trend is most obvious for the highest current fraction, but holds for the monopolar ($\sigma = 0$) data as well. For absolute threshold comparisons, multiple linear regression analysis across pTP fractions indicated an adjusted $R^2 = 0.358$ ($F_{4,20}=4.34$, $p=0.011$). For the relative threshold comparisons, multiple linear regression analysis indicated an adjusted $R^2 = 0.691$, suggesting that the dependent variable, relative log-loudness slope, was predicted by the relative tripolar threshold across σ values ($F_{4,20}=14.39$, $p < 0.001$). These data are consistent with the dynamic range measures (Fig. 4), but are not constrained to stimuli where MCL could be reached.

Figure 7 presents scatter plots comparing absolute (left) and relative TP thresholds (right) to the difference between loudness growth slopes for monopolar and the most focused pTP configuration tested ($\sigma = 0.62$). A Pearson's correlation test revealed a moderate negative correlation of high significance ($r = -0.607$, $p = 0.001$ for absolute levels and $r = -0.556$, $p = 0.005$ for relative values). This result demonstrates that TP thresholds are predictive of the effects of electrode configuration on loudness growth.

Discussion

The results of this study are generally consistent with the working hypothesis that tripolar threshold can be used to qualitatively assess the status of the electrode-neuron interface of individual channels. The present study found that dynamic ranges are smaller (Fig. 4) and loudness growth functions are steeper (Fig. 6) for channels having a relatively high TP threshold, whether evaluated on an absolute basis across subjects or relative to within a subject. Furthermore, loudness growth functions of high-threshold channels were steeper with the more focused partial tripolar configuration than with monopolar for 6 of 8 subjects (Fig. 7). These findings closely mirror the relation between thresholds and amplitude growth functions observed previously with EABRs (Bierer et al., 2011). Finally, equal-loudness contours at threshold and at 50% of the monopolar dynamic range, measured as a function of the partial tripolar focusing parameter, were parallel or steeper for high-threshold channels in a subset of subjects (Fig. 3). A narrower dynamic range or steeper growth of loudness does not itself indicate degraded transmission of intensity information, as the number of discriminable intensity steps over the listener's loudness range may be similar for low- and high-threshold channels. However, as discussed below, the covariance of loudness perception with tripolar threshold adds to previous evidence that both measures are influenced by the electrode-neuron interface and that the interface is highly variable across the cochlea. This finding has potential implications for optimizing clinical fitting procedures.

Single-Channel Thresholds, Electrode Configuration, and the Electrode-neuron Interface

To explain channel-to-channel variability in cochlear implant perceptual measures, we recently developed a model of electrical excitation of the cochlea (Goldwyn et al., 2010). The model consisted of two components: the electrical field produced by a current source within a three-dimensional cylindrical representation of the scala tympani and surrounding bone; and the subsequent activation of spiral ganglion neurons having variable sensitivity to extracellular voltage gradients. Localized changes in the electrode-neuron interface were simulated by varying the distance between the electrode array and neurons (i.e. the radial position within the cylinder) or by altering the number and distribution of responsive neurons. With partial tripolar stimulation, simulated thresholds generally increased as current focusing increased. Higher thresholds were also observed for channels with a relatively large electrode-neuron distance or that were centered near a region of low neural density, or “dead region”. Importantly, such threshold elevations were most pronounced with the tripolar configuration. A greater sensitivity to local irregularities in the electrode-neuron interface may explain the larger variability in threshold with focused stimulation observed in previous studies (Pfungst and Xu, 2004; Bierer, 2007; Long et al., 2014; Chua et al, 2011).

In accordance with the model simulations and previous studies in humans (Pfungst & Xu, 2004; Pfingst et al., 2004; Mens & Berenstein, 2005; Bierer, 2007; Bierer & Faulkner, 2010), the results of Fig. 2 show that thresholds increased with current focusing. Across subjects this increase was significantly more rapid for channels with a high tripolar threshold. However, on closer examination, two types of patterns were observed. In 4 of the 8 subjects, there was a considerable difference in slopes between the low- and high-threshold channels. In the remaining subjects, the threshold-fraction functions were parallel. Note that no subjects exhibited steeper slopes for the low-threshold channel. A possible explanation of the inter-subject variability in Fig. 2 is that, as the σ parameter decreases toward the monopolar configuration and the spatial extent of cochlear activation broadens, the resulting loudness perception becomes less dependent on neurons located in close proximity to the center electrode. For example, while a high tripolar threshold could be indicative of a neural dead region located close to the active electrode, the monopolar threshold for the same channel may depend on the response of neurons farther from the active electrode. This characteristic of monopolar stimulation was demonstrated by Goldwyn et al., (2010) and is a consequence of the variable sensitivity of neurons to electrical field gradients along the axonal axis, resulting in broader threshold activation patterns for monopolar stimulation compared to tripolar, even at the higher current levels required for the latter. Another explanation is that parallel patterns are a result of fairly uniform neural viability but variable electrode positions; that is, for the subjects in which the low-threshold and high-threshold channel contours increase similarly with focusing, the electrode for the high-threshold channels could be farther from the neural elements. Future modeling and cochlear imaging efforts may help elucidate the electrode-neuron interface mechanisms that contribute to these response trends.

Equal-Loudness Contours, Dynamic Ranges and Loudness Growth Functions

The difference between threshold contours for low- and high-threshold channels described above also held for 50% equal-loudness contours (Fig. 3). The effect was only marginally significant, but a lower significance would be expected, as the smaller dynamic ranges of high-threshold channels necessarily leads to less variable 50%DR equal-loudness contours across electrodes compared to the variation in thresholds. Additionally, compliance limits prevented loudness balancing at large pTP fractions for some subjects, such as S26, which may have affected the statistics.

Loudness growth functions were also significantly different for channels with low and high tripolar thresholds, whether analyzed with absolute measures or relative within each subject. For focused configurations, the high-threshold channel had a steeper growth of loudness and consequently a narrower dynamic range than the low-threshold channel. For the monopolar configuration, this relation did not hold for every subject; nevertheless, thresholds were significantly correlated with the slopes of monopolar loudness growth functions when all subjects were considered. Together, the loudness growth and equal-loudness contour data demonstrate that a classification of implant channels based on *threshold* stimulus levels can predict perception at *suprathreshold* levels (as summarized in Fig. 4, Fig. 6, and Fig. 7). The predictive power diminished to some degree when using broader configurations, but, as discussed in the previous subsection, this would be expected if tripolar thresholds were sensitive to the localized status of the electrode-neuron interface rather than to the condition of the cochlea or implant further away.

The computational modeling study of Goldwyn et al (2010) provides some insight into how localized changes in the electrode-neuron interface may affect perception at levels above threshold. As expected, the extent of simulated neural activation along the longitudinal axis of the cochlea generally became wider as current level increased. For a fixed total number of activated neurons and in the absence of a simulated dead region, the spatial distribution of neural excitation was typically narrowest for the tripolar configuration (see Fig. 7 in Goldwyn et al.), due to the relatively steep voltage roll-off with distance from the active electrode. Consequently, the recruitment of neurons for a decibel increase in current was limited for tripolar stimulation, resulting in a much shallower neural growth function compared to broader configurations like monopolar (see Fig. 8 in Goldwyn et al.). This difference in monopolar and tripolar neural growth functions is consistent with observations in animal models (Bierer and Middlebrooks, 2002; Bonham and Litvak, 2008). With a dead region centered on the active electrode, however, the neural growth functions were considerably steeper than those without a dead region. A larger electrode-neuron distance could incur a similar effect. These scenarios, in which neurons distant from the stimulation site are more readily recruited when the current level is high, may explain the steeper loudness growth functions observed in the present study for high-threshold tripolar channels. The model also supports the underlying assumption that loudness depends on summed neural activity. Additionally, both the data and model simulations exhibit smaller differences between low- and high-threshold slopes of growth functions for monopolar stimulation. (A direct comparison of measured and simulated loudness growth functions for one of the subjects, S26, appears in Fig. 6 of Bierer, 2010.)

In as much as evoked potential growth and loudness both reflect neural recruitment, the electrophysiological experiments of Bierer et al (2011) provide empirical support for a configuration-dependent link between threshold and loudness growth. In their study, EABR potentials were obtained for channels with the lowest and highest tripolar thresholds. For stimulation with the partial tripolar ($\sigma = 0.5$) configuration, the amplitude of wave V of the EABR waveform complex increased more steeply as a function of current for the high-threshold channel of most subjects tested. For monopolar stimulation, however, no significant trend was observed. This finding parallels the results of the present study, in which loudness growth functions for the high-threshold channels were steeper than those for the low-threshold channels, with a greater level of significance for the focused pTP configuration.

Taken together, the loudness, EABR, tuning curve, and computational modeling results offer an emerging picture of how two different types of cochlear implant channels might engage the auditory system over a range of current levels. (1) When a low tripolar threshold indicates close proximity to healthy neurons, increasing the current level activates progressively more neurons across a widening region of the cochlea. Because of the nature of electrical field gradients and the distribution of variably-sensitive neurons across the cochlea, this neural recruitment is relatively shallow for focused configurations like partial tripolar; the broader (but lower amplitude) electrical field of the monopolar configuration, on the other hand, provides access to a larger pool of sensitive neurons distributed over a greater portion of cochlea, so neural recruitment is steeper as level increases. (2) In contrast, a channel with a high tripolar threshold, indicating reduced access to viable neurons, requires more current to activate the minimum number of neurons necessary to achieve perceptual threshold. Consequently, even with nominally focused partial tripolar stimulation, the electrical fields are already broad enough that further increases in current lead to relatively rapid neural recruitment, not unlike the low-threshold scenario in (1) for monopolar stimulation. On the other hand, increasing current levels for a high-threshold channel with the monopolar configuration does not necessarily produce even steeper neural recruitment functions – compared to partial tripolar stimuli in (2) or monopolar stimuli in (1) – because the electrical field is broad enough to be minimally influenced by the adverse factors that led to the high tripolar threshold.

Other Loudness and Modeling Studies

Fig. 7 demonstrates that TP threshold accounts for some of the across-subject and channel variability in growth function slopes between focused and broad configurations. Specifically, for low-threshold channels loudness slopes were generally shallower for pTP stimulation than for MP, while for high-threshold channels the opposite was true. This may explain, in part, why a consistent effect of configuration has not been observed in previous studies, for which test channels are typically chosen in a more random fashion. For example, Chua et al (2011) showed that the slopes of loudness growth, plotted as a function of percent dynamic range in dB, were significantly shallower for the tripolar configuration compared to monopolar, similar to the low-threshold channels in the present study. On the other hand, a study from the same research group as Chua et al demonstrated larger dynamic ranges for monopolar than bipolar or tripolar (Zhu et al, 2012; Table 2), which contradicts the present

findings. Likewise, a study of intensity discrimination using monopolar and bipolar configurations showed inconsistent effects of electrode configuration (Drennan and Pfingst, 2005). In all of these previous studies, however, the test electrodes were chosen on the basis of position along the implant array (e.g. apical, middle, basal) and were the same for every subject. Thus, any potential influences of the electrode-neuron interface would not have been accounted for. Interestingly, Drennan and Pfingst (2005) speculated that the higher neural density and closer electrode-neuron distance for apical channels stimulated with focused stimulation led to improved intensity discrimination. Future studies relating intensity discrimination and the putative status of the electrode-neuron interface could further our understanding of loudness processing and its potential influence on speech perception in cochlear implant listeners.

Several investigators have applied computational models to explain loudness perception data in the context of neural activation patterns (Litvak et al., 2007; McKay et al., 2001; Cohen, 2009a). Cohen investigated the potential influence of electrode-neuron interface quality on loudness perception (2009b). As in Goldwyn et al (2010), the model simulated electrical field spread in the cochlea and the subsequent activation of auditory nerve fibers in response to electrical pulse trains. It was found that one essential model parameter, representing the degree of neural survival in the cochlea, had a strong influence on dynamic range and loudness growth curvature. Specifically, a lower number of viable neurons led to a larger dynamic range and shallower loudness growth, opposite of the trend reported by Goldwyn et al (2010). It should be pointed out, however, that neural loss in the Goldwyn study was simulated as a single localized dead region of varying size rather than as variable survival over the extent of the cochlea, as has been observed in histological samples (e.g. Johnsson et al., 1981). In contrast, only global neural survival was considered in the Cohen model. Nevertheless, the results of the Cohen study suggest that widespread neural loss in the cochlea may have a profound effect on loudness perception.

Additional support for an essential role of the electrode-neuron interface in influencing threshold and suprathreshold perception in cochlear implant listeners is evident in a histological study by Kawano and colleagues (Kawano et al., 1998). Neural survival, bone and tissue growth and electrode placement in 5 cochlear implant listeners was analyzed post-mortem and compared with previously collected psychophysical data using the bipolar configuration. It was observed that a higher degree of neuron survival around each electrode tended to occur for electrodes with larger dynamic range for four of the five subjects. It was also found that thresholds were higher when the distance between electrodes and Rosenthal's canal (which houses the cell bodies of the spiral ganglion) was greater. Elevated thresholds and decreased dynamic ranges were also associated with the presence of bone growth and intracochlear fibrous tissue. These findings are generally consistent with the results and central hypothesis of the present study: that channels with higher focused thresholds, smaller dynamic ranges, and steeper loudness growth are the product of a degraded electrode-neuron interface.

Clinical Implications

The results of this and other studies investigating perceptual variability from channel-to-channel in cochlear implants may have implications for clinical mapping procedures. Considering that implant listeners probably do not make full use of their programmed channels to extract critical cues in speech stimuli, prudent deactivation of channels most adversely affected by a poor electrode-neuron interface has the potential to improve speech understanding. Optimizing the array of active channels is particularly important as current focusing and current steering strategies become more commonplace (Koch et al., 2007; Srinivasan et al, 2013), as, for instance, dead regions could be avoided by stimulating at optimal locations between them.

Previous studies have attempted to identify implant channels for deactivation using behavioral and physical measures other than tripolar threshold (Collins et al. 1997; Garadat et al., 2012; Noble et al., 2013). For example, Garadat and colleagues used temporal modulation detection as a metric to identify poorly functioning channels (Garadat et al., 2012). Channels with low temporal resolution were deactivated from the implant listeners' maps, resulting in improvements on some speech perception tasks. Noble and colleagues (2013) used computed tomography and computational modeling to estimate the physical location of the electrodes relative to spiral ganglion neurons. In that study, channels were chosen for deactivation to reduce the predicted stimulation of overlapping neural populations based on patient-specific CT imaging analysis of the cochlea. Improvements in speech perception were seen in patients after one month of experience with the new settings. Based on the covariance of tripolar thresholds to a variety of perceptual and physiological measures, one could speculate that tripolar thresholds would be sensitive to the same adverse factors – large electrode-neuron distance and neural degeneration – as the channel-identification metrics used in these other studies. Tripolar thresholds are straightforward and simple measures, and thus could be readily implemented in a clinical setting for devices that support focused stimulation. Further studies are necessary to develop clear criteria of what constitutes a “poor” channel in terms of clinical mapping, and whether tripolar threshold alone is sufficient for this purpose. Other questions concern the number of channels a typical listener requires, and how to program the implant when all channels meet the “poor” criteria.

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Figure 1.

Single-channel behavioral thresholds measured using three electrode configurations for all subjects. Each panel plots the single-channel detection thresholds for a given subject (indicated in the top left corner). The abscissa represents cochlear implant channel from apical to basal and the ordinate represents detection threshold in decibels relative to 1 μ A. Electrode configuration is indicated by symbols and for the triangles is between 0.8 and 1. The vertical dashed lines in each panel indicate the lowest, median, and highest threshold

channels obtained with the TP configuration. Multiple sessions were required to obtain the data plotted in this figure.

Figure 2.

Single-channel behavioral thresholds across subjects and configurations for the lowest, median and highest tripolar threshold channels. Each panel shows the single-channel detection thresholds for a given subject as a function of partial tripolar fraction (σ) on the abscissa. The fill of the symbol indicates the stimulus channel type based on tripolar threshold; open = lowest, grey = median, black = highest (corresponding to the vertical dashed lines in Fig. 1). The dashed lines represent the linear fits minimizing the mean-squared error for each channel.

Figure 3.

Single-channel equal-loudness contours across subjects and configurations for the lowest, median and highest tripolar threshold channels. Each panel shows the single-channel equal-loudness contours for a given subject as a function of partial tripolar fraction (σ) on the abscissa. Conventions are as in Fig. 2.

Figure 4.

Dynamic range as a function of tripolar threshold. Top panels show the within-subject and within-configuration dynamic range versus TP threshold on absolute scales. Bottom panels show the same data plotted relative to the respective means across low-median-high channel type, for MP (left), pTP with $\sigma = 0.5$ (middle), and pTP with $\sigma > 0.6$ (right). Subject is indicated by symbol. Dashed lines depict linear regression fits. In the right panels, filled symbols indicate stimuli for which the most comfortable level could not be reached but were rated at least 5.5 on the loudness rating scale.

Figure 5.

Loudness growth functions for the three test electrodes across electrode configurations (left to right) for one cochlear implant subject, S22 on a linear (top) and logarithmic scale (bottom). The loudness estimations on a scale from 1 to 100 (ordinate) are plotted as a function of stimulus level in dB re 1 μ A (abscissa) for the channels with the lowest (blue), median (green) and highest (red) tripolar thresholds. Numbers in the legend correspond to channel number for the test electrodes. Each panel presents data for a different electrode configuration with monopolar on the left and partial tripolar $\sigma = 0.75$ on the right. The dashed lines are least-squared error linear fits to the log of the loudness estimate functions.

Figure 6.

Log-loudness slope as a function of tripolar threshold. Top panels show the within-subject and within-configuration log-loudness slope versus TP threshold on absolute scales. Bottom panels show the same data relative to the respective means across channel type, for different configurations: from left to right, MP, pTP with $\sigma = 0.5$, $\sigma = 0.62/0.68$, and $\sigma = 0.75$. Subject is indicated by symbol. Dashed lines depict linear regression fits.

Figure 7.

The relation between tripolar threshold and the difference between MP and focused pTP log-loudness slopes for absolute values (left) and relative values (right). The difference in log-loudness slopes for MP minus pTP slopes (ordinate) as a function of relative tripolar threshold in dB (abscissa). Subjects are indicated by symbol. Dashed lines represent the linear regressions.

Table 1

Subject Information

Subject Number	Sex	Age (years)	Duration of Severe Hearing Loss (years)	Duration of CI Use (years)	Etiology of Hearing Loss
S22	F	70	15	2.5	Hereditary
S23	M	65	22	2.5	Noise Exposure
S26	F	31	4	3	Atypical Meniere's Disease
S27	M	80	20	1.5	Unknown
S28	F	71	40	6 months	Hereditary
S29	M	80	30	3	Noise Exposure
S30	F	46	23	5	Hereditary
S31	F	46	41	9 months	Ototoxicity