Understanding Frequency Encoding and Perception in Adult Users of Cochlear Implants

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Abstract

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An increasingly common approach to treating severe to profound hearing loss in children and adults is cochlear implantation, with over 219,000 users worldwide. While most cochlear implant (CI) users achieve high levels of speech understanding in quiet, many continue to struggle with speech in the presence of background noise and music appreciation. One working hypothesis is that these deficits are related to impaired spectral resolution with CIs. Both device-related and patient-related factors are thought to contribute to impaired spectral resolution. This series of experiments were conducted to identify ways to improve spectral resolution in individuals using two approaches. First, a device-centered approach was used with a focused stimulus configuration to determine if it was possible to identify poorly-functioning CI channels with the hopes of

improving how CIs are mapped clinically, using a behavioral (experiment 1) and an objective measure (experiment 2). Given that spectral resolution is likely not entirely related to device factors, and not all aspects of the device are modifiable or controllable, we also set out to determine if we could improve the implant user's ability to make use of the information they receive with their implant through auditory training (experiment 3).

The results of the first two experiments demonstrated that problem channels can be identified using behavioral (single-channel thresholds and tuning curves) and objective measures (EABR). Using a single-subject design, we also showed that it is possible to alter perception of frequency through auditory training. All trained subjects improved on tests of speech in noise, but improvements did not generalize to tests of speech in quiet, music perception, or quality of life. The pattern of learning suggests that the improvements cannot be entirely explained by improved spectral resolution, and motivates future directions aimed at understanding what mechanism of action in the training paradigm contributed to improved speech reception in noise.

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Most of all, I thank my family, my parents Kathy and Tom Faulkner, and my brother, Terry, for their continued support and encouragement. Finally, I could not have made this journey without my fiancé, John, who has stood beside me through the highs and lows, helps me find perspective and balance, and never fails to make me laugh.

DEDICATION

This dissertation is dedicated to the memory of my father, Thomas A. Faulkner, who encouraged me to follow my passion and taught me, by example, about the importance of serving others.

INTRODUCTION

An increasingly common approach to treating severe to profound hearing loss in children and adults is cochlear implantation. According to the FDA, over 219,000 individuals worldwide have been fit with cochlear implants, including 42,600 adults and 28,400 children in the United States (NIH, 2011). The use of cochlear implants (CI) has been shown to significantly improve the quality of life for those who receive them. However, despite the improved ability to hear in quiet, most cochlear implant listeners still struggle with speech in the presence of background noise as well as music appreciation. One working hypothesis is that these deficits are related to impaired spectral resolution with cochlear implants. Spectral resolution is defined as the ability to resolve the individual frequency components within a complex sound and is important for speech perception in background noise (e.g., Badri et al., 2011; Leek and Summers, 1996). Impaired spectral resolution has been documented in CI listeners (Bierer, 2010; Bierer and Faulkner, 2010; Nelson and Jin, 2004; Fu and Nogaki, 2004; Nelson et al., 2003; Qin and Oxenham, 2003; Donaldson and Nelson, 2000; Henry et al., 2000; Fu et al., 1998; Nelson et al., 1995). Both device-related and patient-related factors are thought to contribute to impaired spectral resolution. Device-related factors refer to the encoding of incoming signals at the periphery, while patient-related factors refer to how well people make use of the information provided by their device.

To better understand what is meant by device- and patient- related factors and how they contribute to the perception of frequency, it is first necessary to: 1) review how sound is encoded by the device, and 2) describe the types of patient-related variables thought to contribute to impaired spectral resolution.

Description of the CI device

A CI is a surgically implanted device for those with varying degrees of deafness who no longer receive adequate benefit from hearing aids. CIs are very different from hearing aids. Rather than amplifying acoustic signals, a CI converts acoustic signals into electrical signals which are delivered directly to the auditory nerve.

As illustrated in Figure 1, the CI consists of internal and external components. The external speech processor (1) contains the microphone that picks up sounds in the environment and converts them into digital signals. The signals are then transmitted across the skin using radio frequency signals to the internally implanted receiver-stimulator (2). The digital signal is converted into electrical pulses, which are sent to the electrode array that is surgically inserted into the cochlea (3). The electrode array has electrode contacts along the length of the array to stimulate different regions of the cochlea with electric current pulses, which in turn, stimulate groups of auditory neurons (4). The number and spacing of the electrode contacts varies by manufacturer and range from 8 to 24.

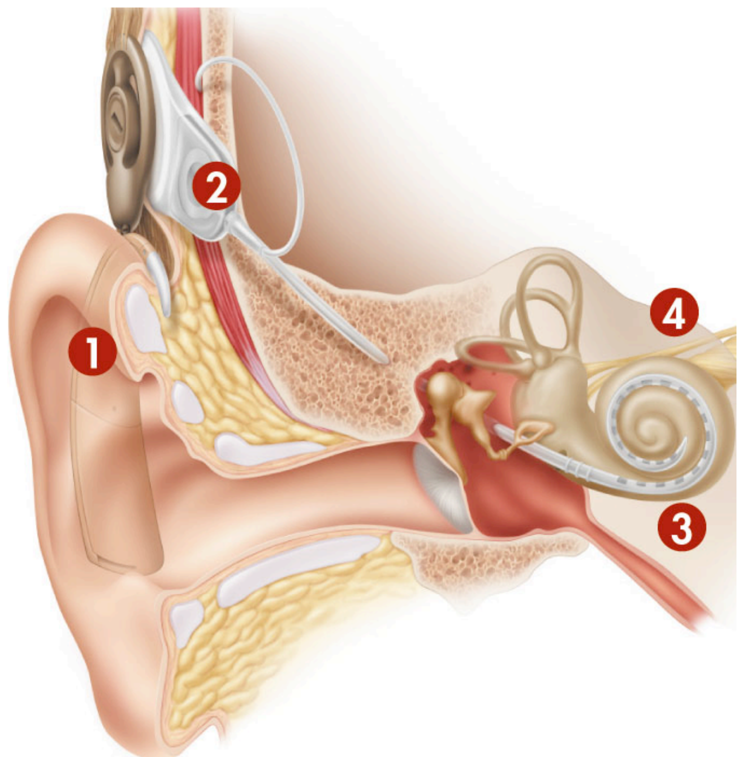

Figure 1. Cochlear implant illustration. Cochlear Corporation.

Frequency encoding

The auditory system is tonotopically ordered, meaning that sounds with different frequencies stimulate different regions along the basilar membrane in the cochlea, with high frequencies stimulating near the base of the cochlea and low frequencies stimulating at the apex. Evidence of distinct tonotopy is present along the auditory pathway, from single auditory nerve fibers (e.g., Kiang and Moxon, 1972) to the midbrain (Merzenich, 1974) and finally at the cortex (see Ehret, 1997 for a review). The CI was designed to take advantage of this arrangement by stimulating in a tonotopic or cochleotopic manner. Incoming signals are analyzed and separated into a limited number of frequency bands or "channels," where low frequency sounds activate apical electrodes, and high frequency sounds activate basal electrodes. The temporal envelope, or the slow modulations in amplitude over time, is extracted at the output of each filter and determines the level of current applied to each electrode over time, while the temporal fine structure, or the rapid oscillations within a frequency band, is discarded. Therefore, only the amplitude spectrum of the acoustic sound is encoded, thus the primary mechanism for frequency encoding with a modern cochlear implant is by place coding (i.e. cochleotopic stimulation). In quiet listening situations, envelope cues are sufficient for speech perception, but for tasks involving speech understanding in noise or music perception, the loss of fine structure information is detrimental to performance (e.g., Xu and Pfingst, 2008). Additionally, the process of encoding frequency with a CI is not as straightforward in clinical application because there are both device- and patient-related factors that can further limit how frequency information is encoded and utilized by individual listeners.

Device-Related variables

There are several device-related factors that can further alter how the spectral cues from the acoustic signal are transmitted and represented in the auditory system. These factors include the limited number of spectral channels, lack of channel independence, channel interaction, disruptions in the frequency-to-place coding, and the electrode-neuron interface. As an example, even though CIs have 8 to 22 channels along the electrode array to stimulate regions along the cochlea tonotopically, many CI listeners may only be able to functionally make use of only seven to ten channels (Friesen et al., 2001). Also, modern CIs employ the monopolar electrode configuration which has been shown to stimulate a broad electrical field in the cochlea (Jolly et al., 1996; Kral et al., 1998), with a large degree of overlap across channels. While the use of more focused stimulus configurations (e.g., tripolar or phased-array) improves spectral resolution by reducing the current spread to adjacent regions of auditory neurons (e.g., Bierer and Faulkner, 2010), preliminary attempts to apply these configurations into clinical maps have shown promise, with listener's performing as well or slightly better with focused stimulation (Berenstein et al., 2008). Finally, in many cases, the normal place-code is disrupted because there is a mismatch between the frequency of an incoming signal and the place of stimulation (e.g., Svirsky et al., 2004; Fu et al., 2002; Rosen et al., 1999).

Finally, the electrode-neuron interface refers to the effective distance between an electrode and its closest responsive auditory neurons and can be influenced by both device- and patient-related factors, including: surgical placement and insertion depth of the electrode array, fibrous and bony tissue growth around the electrode array, and variable patterns of spiral ganglion nerve survival. Many of the device-related manipulations, such as increasing the number of channels and the use of focused stimulation configurations, are limited by the electrode-neuron interface. All of these device-related factors are examples of how much of the spectral detail from the original acoustic signal may not be preserved at the output of the device. However, the poor spectral resolution documented in CI listeners can not be entirely explained by the device-related factors, given the encoding of the signal, there are likely patient-related factors that might limit how the auditory system can respond to the information provided by their device.

Patient-Related variables

Even when people are fit with similar CI devices, the ability to make use of incoming signals can drastically differ across individuals. Performance variability has been attributed to many potential sources, including: etiology of deafness, age at implantation, amplification history, and duration of deafness, just to name a few. But to date, no single predictor to speech perception scores has been entirely explained by either patient- or device-related factors (Firszt et al., 2004; Skinner et al., 1994). While most CI listeners understand speech in quiet, perception consistently breaks down for many individuals when speech is presented in noise (Zeng et al., 2005; Firszt et al., 2004; Stickney et al., 2004; Turner et al., 2004; Zeng 2004; Nelson et al., 2003; Fu et al., 1998). Additionally, CI listeners show poor performance on music perception, for example, requiring larger frequency differences than normal hearing listeners to hear pitch changes as well as poor melody and timbre recognition (for a review, see Drennan and Rubinstein, 2008).

More specifically, patient-related factors contributing to spectral resolution include both peripheral (e.g., nerve survival) and central (e.g., central changes as a result of years of auditory deprivation). As previously discussed, the electrode-neuron interface can be affected by patient-specific factors, including nerve survival, which is likely related to the duration of severe hearing loss prior to implantation. Whatever the cause, poor electrode-neuron interface has been related to poor spatial resolution, or the ability to distinguish stimulation on different channels. Beyond the peripheral encoding of the signal, the spectral strength of the signal, or how well the frequency content of a given signal is maintained through the auditory system, can be influenced by neural reorganization in the auditory pathway as a result of biological aging, years of auditory deprivation prior to implantation, or the reintroduction of stimulation with the CI (biological aging: Tremblay and Burkard, 2007; deprivation: Willott, 1996; stimulation: Fallon, Irvine and Shepherd, 2008; Leake et al., 2000). As previously reviewed, the electrically transmitted signal is different in many ways from its original acoustic signal, and these patient-factors might alter how well a given listener can overcome the mismatch.

Further, other general cognitive factors such as attention, memory, and learning might play a significant role in translating these neural codes into meaningful events (Hartley and Moore, 2002; Patterson et al., 1982). For example, when compared with listening to speech in quiet, listening to speech in noise activates more complex neural networks, including linguistic, attentional, memory, and motor areas of the brain (Salvi et al., 2002). CI listeners might also use different listening strategies, or apply different perceptual weights to acoustic cues than their normal hearing counterparts, and their listening strategies might change depending on the listening conditions (i.e. global spectral contrasts versus fine spectral details). For these reasons, there is interest in determining if it is possible to modify some of these patient-related factors through training.

Summary

In summary, spectral resolution is hypothesized to be a contributor to impaired perception, for both speech in noise and music, because of all the different types of variables, both device- and patient-related, that impoverish the signal that the CI listener receives. The goal of the research presented in this dissertation was to identify ways to improve spectral resolution in individuals, so that their speech perception and music appreciate improve and hopefully their quality of life is also enhanced. To do this, a series of experiments were conducted, using two different approaches to studying device- and patient-related variables related to spectral resolution. (1) First, a device approach was used to determine if it was possible to characterize channels with poor electrode-neuron interfaces with the hopes of improving how individual CI channels are programmed clinically. Experiment 1 addressed this issue using a behavioral approach. Experiment 2 involved objective measures - the electrically-evoked auditory brainstem response. (2) Second, given that spectral resolution is likely not entirely related to device factors, and not all aspects of the device are modifiable or controllable, we set out to determine if it was possible to train people to make better make use of the information they receive through their CI device (Experiment 3). The primary research aims for each experiment are as follows:

Experiment 1:

Do channels with elevated thresholds measured with a focused stimulation configuration show poor spatial selectivity? We hypothesized that a focused stimulation configuration would be more sensitive to local factors (e.g., local spiral ganglion survival) and that elevated thresholds likely reflect a poor electrode-neuron interface. To test this hypothesis, we measured single-channel thresholds with a focused stimulus configuration to identify channels with low, median, and high thresholds for each subject. The widths and slopes of forward-masked psychophysical tuning curves (PTCs) characterized the extent of spatial selectivity for each of these channels.

Experiment 2:

Can the electrically-evoked brainstem response (EABR), measured with a focused stimulation configuration, be used to identify channels with a poor electrode-neuron interface? We hypothesized that the EABR might be sensitive to channels with a poor electrode-neuron interface, and that channels with higher EABR thresholds would have steeper amplitude growth functions. To test this hypothesis, we measured EABR growth functions on channels suspected of having good or poor electrode-neuron interface, identified behaviorally in experiment 1. The slopes of the EABR amplitude growth functions were compared across channels and configurations and related to the behavioral thresholds and PTCs obtained in experiment 1.

Experiment 3:

1) Does focused auditory training using spectral ripple stimuli result in improved spectral ripple discrimination in adult CI listeners? We hypothesized that spectral resolution will improve with training, as indicated by improved spectral ripple threshold, and that changes in spectral ripple threshold will reflect changes in the way CI listeners use the frequency information provided to them through their device. Because of the heterogeneity across CI users we also expect that changes in performance will follow different time lines for different CI users, and that patterns of generalization and retention will also vary across individuals. To do this, we designed a lab-based training program using spectral ripple stimuli and measured CI listeners' performance before, during, and following auditory training, using a single-subject design.

2) Do improvements on spectral ripple discrimination transfer to untrained tasks of speech-in-quiet, speech-in-noise, music perception, and quality of life? We hypothesized that because speech perception is dependent, in part, on spectral resolution, improvements in spectral resolution would transfer to untrained tasks that were directly related to spectral resolution (speech and music), and to quality of life.

Identifying Cochlear Implant Channels with Poor Electrode-Neuron Interface: Partial Tripolar, Single-Channel Thresholds and Psychophysical Tuning Curves

Julie Arenberg Bierer and Kathleen F. Faulkner

Objective: The goal of this study was to evaluate the ability of a threshold measure, made with a restricted electrode configuration, to identify channels exhibiting relatively poor spatial selectivity. With a restricted electrode configuration, channel-to-channel variability in threshold may reflect variations in the interface between the electrodes and auditory neurons (i.e., nerve survival, electrode placement, and tissue impedance). These variations in the electrode-neuron interface should also be reflected in psychophysical tuning curve (PTC) measurements. Specifically, it is hypothesized that high single-channel thresholds obtained with the spatially focused partial tripolar (pTP) electrode configuration are predictive of wide or tip-shifted PTCs.

Design: Data were collected from five cochlear implant listeners implanted with the HiRes90k cochlear implant (Advanced Bionics Corp., Sylmar, CA). Single-channel thresholds and most comfortable listening levels were obtained for stimuli that varied in presumed electrical field size by using the pTP configuration for which a fraction of current (σ) from a center-active electrode returns through two neighboring electrodes and the remainder through a distant indifferent electrode. Forward-masked PTCs were obtained for channels with the highest, lowest, and median tripolar ($\sigma = 1$ or 0.9) thresholds. The probe channel and level were fixed and presented with either the monopolar ($\sigma = 0$) or a more focused pTP ($\sigma \geq 0.55$) configuration. The masker channel and level were varied, whereas the configuration was fixed to $\sigma = 0.5$. A standard, three-interval, two-alternative forced choice procedure was used for thresholds and masked levels.

Results: Single-channel threshold and variability in threshold across channels systematically increased as the compensating current, σ, increased and the presumed electrical field became more focused. Across subjects, channels with the highest single-channel thresholds, when measured with a narrow, pTP stimulus, had significantly broader PTCs than the lowest threshold channels. In two subjects, the tips of the tuning curves were shifted away from the probe channel. Tuning curves were also wider for the monopolar probes than with pTP probes for both the highest and lowest threshold channels.

Conclusions: These results suggest that single-channel thresholds measured with a restricted stimulus can be used to identify cochlear implant channels with poor spatial selectivity. Channels having wide or tip-shifted tuning characteristics would likely not deliver the appropriate spectral information to the intended auditory neurons, leading to suboptimal perception. As a clinical tool, quick identification of impaired channels could lead to patient-specific mapping strategies and result in improved speech and music perception.

(Ear & Hearing 2010;31;247–258)

INTRODUCTION

The design of multichannel cochlear implants takes advantage of the tonotopic organization of the cochlea, mapping the spectral information in an acoustic signal to an array of stimulating electrodes. In response to a stimulus, the spatial distribution of auditory nerve fibers activated by each channel contributes to the perception of that stimulus. Previous modeling, physiology, and psychophysical studies have suggested a number of factors that can affect the tonotopic extent of auditory nerve activation (physiological – van den Honert and Stypulkowski 1984; Bierer & Middlebrooks 2002; psychophysical – Pfingst & Xu 2004; modeling – Hanekom 2005). Perhaps, the best understood of these factors is electrode configuration (Jolly et al. 1996; Kral et al. 1998), which determines how current flows between electrodes for a given implant channel. Poor surgical placement of the implant (Cohen et al. 2003; Wardrop et al. 2005a,b), electrode insertion depth (Gani et al. 2007; Kos et al. 2007; Skinner et al. 2007; Finley et al. 2008), and fibrous and bony tissue growth around the electrode array (Spelman et al. 1982; Hanekom 2005) may also have profound effects on current flow in the cochlea. Any of these conditions, or a combination of them, may increase the effective distance between an electrode and its closest responsive auditory neurons and thus degrade the selectivity of that electrode. Another condition that may increase the electrode-to-neuron distance is the variable patterns of spiral ganglion degeneration (Hinojosa & Marion 1983; Nadol et al. 2001). If such impaired electrodes could be readily identified, the clinical mapping procedure could be tailored to minimize their adverse effect on listening performance. This study shows that implant channels exhibiting relatively wide psychophysical tuning patterns—implying an impairment of spatial selectivity—may be detected by measuring thresholds using a spatially focused electrode configuration.

Modern cochlear implants primarily use the monopolar (MP) electrode configuration, which consists of an active intracochlear electrode and a remote, extracochlear return. On the basis of electrostatic principles, this configuration produces a relatively broad electrical field in the cochlea (Jolly et al. 1996; Kral et al. 1998). More restricted electrical fields require an arrangement of two or more intracochlear electrodes to act as current sinks and sources. Examples include bipolar (BP), which is used clinically, quadrupolar or tripolar (TP) (Jolly et al. 1996; Kral et al. 1998; Litvak et al. 2007), and various phased-array configurations (van den Honert & Kelsall 2007). Phased-array configurations sharpen electrical fields by counterweighting several return electrodes relative to the active electrode. Physiological studies revealing the extent of activation in the auditory system to cochlear implant stimulation are generally consistent with electrostatic predictions. For instance, the TP configuration has been shown to be more tonotopically restricted than BP, which in turn is more restricted than MP (Kral et al. 1998; Bierer & Middlebrooks 2002; Snyder et al.

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2004; Snyder et al. 2008; Bonham & Litvak 2008). Several lines of psychophysical evidence, including simultaneous channel interaction (Boex et al. 2003; de Balthasar et al. 2003; Bierer 2007) and cross-channel masking (Chatterjee et al. 2006; Nelson et al. 2008), provide additional support for the improved spatial selectivity of focused configurations.

A recent study by Bierer (2007) suggests that the TP configuration is sensitive to factors within the cochlea that can degrade transmission of information to the auditory nerve. Thresholds obtained with the TP mode exhibited substantial variability from channel to channel. Somewhat less channel-to-channel variability was observed, on average, with the BP configuration, and the lowest variability in all subjects was observed with the MP configuration. One interpretation of these findings is that those channels with increased TP thresholds had a poor electrode-to-neuron interface, whether due to the loss of nearby spiral ganglion neurons, the placement of electrodes far from the osseous spiral lamina, or other factors; threshold perception for those channels could be reached only when the current was high enough to stimulate distant, but viable, portions of the spiral ganglion. In contrast, the smaller channel-to-channel variability observed with the MP configuration (and to a lesser degree than the BP configuration) was presumably due to its broader electrical field, which could stimulate distant neurons with only a small increase in current. The Bierer study further showed that listeners with poorer word recognition were those with highly variable thresholds with TP. However, no attempt was made to implicate any particular high-threshold TP channel as contributing to a functional impairment. This study was designed to explore whether such a relationship exists by comparing TP threshold measures and psychophysical tuning curves (PTCs).

In acoustic hearing, PTCs are used to evaluate the frequency selectivity of the auditory system. Moore and Alcantara (2001) measured PTCs in hearing-impaired subjects to assess the location and extent of cochlear "dead regions," defined as an area of inner hair cell loss and/or spiral ganglion neuron damage. The PTCs were obtained using narrowband sounds of varying center frequency to mask a pure-tone probe with a fixed frequency and level. In normal-hearing listeners, the tip of each PTC closely approximated the probe frequency. In contrast, for hearing-impaired listeners, the PTC tip at some probe frequencies was shifted basally or apically, consistent with the activation of auditory neurons at the edge of a dead region. Moore concluded that PTCs are a sensitive diagnostic tool for identifying irregularities in hearing across frequencies, even when pure-tone audiograms do not reveal a localized deficit.

Studies of cochlear implant listeners suggest that PTCs are also sensitive to localized cochlear factors affecting electrical hearing. Nelson et al. (2008) observed tip-shifted and broader tuning curves in some subjects. Similar to the acoustic forward masking studies of Moore, they interpreted these PTC shapes as possibly reflecting localized spiral ganglion loss. However, only one channel was evaluated per subject, and therefore, changes in tuning properties across the array could not be assessed. In other studies, in which PTC or forward masking patterns were measured on multiple channels for each subject, significant channel-to-channel variability of tuning width and shape was observed, especially with a focused configuration (Chatterjee et al. 2006; Hughes & Stille 2008). However, a

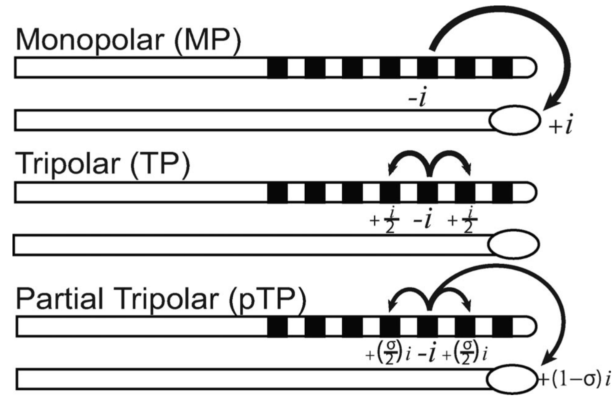

Fig. 1. Schematic of electrode configurations. The current path from active to return electrode(s) is represented by the arrows. The magnitude of current at the active electrode, *-i*, indicates that biphasic pulses were delivered cathodically first. For monopolar (top), the return current is delivered to an extracochlear return electrode located in the case of the internal device; for TP (middle), the return current is equally divided and delivered to the two flanking intracochlear electrodes. For pTP (bottom), a fraction (σ) of the return current is directed to the extracochlear return electrode.

direct relationship between tuning properties and single-channel thresholds was not explored. As discussed above, thresholds measured with the TP configuration also exhibit significant variability across the implant array of an individual listener. If high TP thresholds and tip-shifted or broad tuning reflect the same underlying cochlear conditions, some correlation between these metrics might be expected. For example, the additional current required for a high-threshold TP channel to achieve perception (compared with a channel with a lower threshold) implies that neurons relatively distant to that channel's nominal place along the cochlea would be stimulated because of longitudinal current spread. If the channel is the probe channel in a forward masking task, at least some remote channels would make effective maskers, leading to broad or tip-shifted tuning curves. In addition, the restricted current spread of the TP configuration should make it an effective probe stimulus for measuring PTCs, which, in theory, are more sensitive to local factors if the extent of auditory nerve activation is limited (Moore & Alcantara 2001; Kluk & Moore 2006; Nelson et al. 2008).

Despite the potential advantage of the TP configuration to identify channels with a poor electrode-neuron interface, its practical application is limited because of its relatively high current requirements, a trade-off that has been well documented (Spelman et al. 1995; Kral et al. 1998; Mens & Berenstein 2005; Bierer 2007). A relatively novel electrode configuration, partial tripolar (pTP), has been proposed as one way of balancing current spread and current level requirements (Mens & Berenstein 2005; Jolly et al. 1996; Litvak et al. 2007). pTP is a hybrid of the MP and TP configurations, whereby a fraction of the return current is delivered to the flanking electrodes, whereas the remainder flows to the distant extracochlear ground. A fraction of zero is equivalent to MP (all current is directed to the extracochlear electrode), whereas a fraction of one is TP (no current is directed to the extracochlear electrode; Fig. 1). In this study, single-channel pTP thresholds were measured using the largest current fraction that allowed

TABLE 1. Subject information

Subject number	Sex	Age (yr)	Duration of severe hearing loss (yr)	Duration of cochlear implant use	Etiology of hearing loss	HINT sentences in quiet	CNC words in quiet	CNC phonemes in quiet
S9	F	64	24	3 yr	Hereditary	99%	48%	75%
S22	F	67	12	9 mo	Hereditary	92%	68%	76%
S23	M	62	19	1.5 yr	Noise exposure	94%	68%	85%
S24	F	74	2	1 yr	Unknown	100%	54%	73%
S26	F	30	2	1.8 yr	Atypical Menieres disease	99%	56%	78%

CNC, Consonant Nucleus Consonant test from the Minimum Speech Test Battery for Adult Cochlear Implant Users, House Ear Institute and Cochlear Corporation, 1996.

threshold measurements across all channels for each subject. To test the hypothesis that a focused configuration can predict which channels have impaired spatial selectivity, PTCs were obtained for probe channels corresponding to the lowest, median, and highest pTP threshold channels for each subject. The effect of configuration on PTC characteristics was tested by applying both the pTP and MP configurations for the probe channel stimulus. Together, these methods offer a promising approach to identify cochlear implant electrodes that may be functionally impaired by spiral ganglion loss, poor electrode placement, or other factors.

SUBJECTS AND METHODS

Subjects

Five adult cochlear implant listeners, four women and one man, participated in this study. The subjects, ranging in age from 30 to 74 years, were all native speakers of American English who had become deaf postlingually. All participants wore a HiRes90k implant with a HiFocus 1J electrode array (Advanced Bionics Corp., Sylmar, CA) having a center-to-center electrode distance of 1.1 mm, and all had at least 9 months of experience with their implants. Details about individual subjects can be found in Table 1. The speech perception scores shown in Table 1 are from the most recent standard clinical visit (within 3 mo of the beginning of the experiment) for each subject. The subjects made between eight and 13 visits to the laboratory (2- to 4-hr long visits) to complete the data collection. Each participant provided written consent, and the experiments were conducted in accordance with guidelines set by the Human Subjects Division of the University of Washington.

Stimuli

Biphasic, charge-balanced (cathodic phase first on the active electrode) pulses were presented with 102 μsec per phase and at a rate of 980.4 pulses/sec. The active electrode and the return electrodes that complete the current loop are called a "channel." Channel numbers denote the position of the active electrode on the implant array and, for this experiment, ranged from 2 (apical) to 15 (basal).

The electrode configurations used include MP, TP, and pTP at various current fractions (Fig. 1). The MP configuration consists of a single active intracochlear electrode and an extracochlear return electrode. The TP electrode configuration consists of an intracochlear active electrode, with the return current divided equally between each of the two nearest flanking electrodes. The pTP configuration is similar, but only a fraction of the return current, represented by "σ," is delivered to the flanking electrodes; the remainder flows to the distant extracochlear ground. Therefore, a σ of 0 is equivalent to MP (all current is directed to the extracochlear electrode), and a σ of 1 is equivalent to TP (no current is directed to the extracochlear electrode). In this article, MP and TP channels are often referred to as pTP channels with fractions of 0 and 1, respectively.

The stimuli were delivered to the implant using a clinical interface controlled by the Bionic Ear Data Collection System, version 1.15.158 (Advanced Bionics). All stimulation levels used were within the current levels supported by the implant. On the basis of the clinically measured impedance values, the maximum compliance limits were calculated at the beginning of each test session (Advanced Bionics, personal communication). The compliance limit was defined as the maximum voltage supported by the device (8 V) divided by the impedance.

A personal computer, which was used to run the Bionic Ear Data Collection System software, communicated with the clinical interface through a dedicated Platinum Series Processor. The same computer was used for online data collection through subject response using a computer mouse. Subjects were given practice sessions when new tasks were introduced until they became familiar with the task. They did not receive trial-by-trial feedback.

Data analysis and plotting was mainly performed in units of decibels (relative to 1 mA) throughout this study because the proportional changes in current between conditions are emphasized. The dynamic range was also highly variable from channel to channel and from subject to subject, making a logarithmic scale more suitable for comparisons.

Single-Channel Thresholds

For each subject, thresholds on every channel were obtained for a 204-msec pulse train using three configurations: MP, pTP at a fixed current fraction of $\sigma = 0.5$, and the largest common pTP fraction for which a threshold could be obtained on all 14 channels (either $\sigma = 1$ or 0.9). For clarity, this latter configuration will be referred to as "TP," even when the fractional current is not strictly 1. (For one subject (S24), this fraction was 1; for all others, the fraction was 0.9). The term "pTP," unless specified, will be reserved for channels having a current fraction intermediate to the most restricted ($\sigma \leq 0.9$) and the most broad ($\sigma = 0$, MP).

The lowest, median, and highest threshold channels obtained with the TP configuration for each subject were identified for further testing. Thresholds were measured for this subset of channels for several pTP fractions. All thresholds were measured with an adaptive two-down one-up, three-interval, three-alternative forced-choice procedure that converged on the 70.7% correct point on the psychometric function (Levitt 1971). Each run started at a level estimated to be suprathreshold, and subjects responded using a mouse to indicate the interval that contained the signal. Twelve reversals (i.e., changes in the direction of the signal level) were measured for each trial, and the levels for the last eight were averaged and taken as threshold. For the first two reversals, the signal was adjusted in 2-dB steps; for the other 10 reversals, it was adjusted in 0.5-dB steps.

For each condition, at least two runs were collected and averaged. If the threshold of two runs differed by >1 dB, a third run was collected and all three runs were averaged. Generally, the third run's threshold was between those of the first two runs. The standard deviation for the last eight reversals from each run was measured; if the value exceeded 1 dB, the subject was given a break and threshold was subsequently remeasured. In addition, because large numbers of trials were also an indication of subject fatigue, the total number of trials per run was limited to 75. To reduce subject fatigue, conditions (e.g., thresholds, most comfortable level [MCL]) were alternated within each testing session as much as possible.

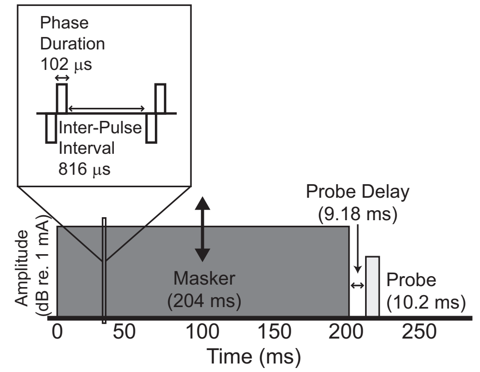

Fig. 2. Schematic of forward masking paradigm showing masker/probe amplitude (decibels relative to 1 mA) as a function of time (milliseconds). The masker was a 204-msec pulse train that varied in stimulus level but was fixed with a pTP fraction of $\sigma = 0.5$. The masker pulse train preceded the 10.2-msec probe pulse train by 9.18 msec. The probe configuration was either $\sigma = 0$ or the largest sigma possible for each subject. The probe was fixed in level to 3 dB above threshold, unless otherwise specified. The masker varied in level to reach masked threshold, the amount of masking necessary to just mask the probe. The inset shows the biphasic pulses that made up both the masker and probe pulse trains. Pulses were 102 μsec/phase presented at a rate of 918 pulses per second (interpulse interval of 0.8 msec). The inset shows two biphasic pulses from the pulse train.

Most Comfortable Level

Accurate thresholds and MCLs were important for setting stimulus levels on the masker channels during the PTC procedure, which is described in the following subsection. The MCL was determined by presenting a pulse train just above threshold (204 msec pulse train, $\sigma = 0.5$) and asking the subject to adjust the level by clicking one of two options labeled "up" and "down." The subject was asked to set the level to the subjective rating of "loud but comfortable," corresponding to 7 on the 1 to 10 clinical loudness rating scale (Advanced Bionics). The level was changed in 1-dB steps until the subject clicked the "down" button; thereafter, it was changed in 0.5-dB steps. At least two runs were collected and averaged for each MCL condition. If the two measured MCLs differed by >1 dB, a third run was completed and all three runs were averaged. For two electrodes in S23, MCL could not be reached because the current level for maximal loudness perception exceeded the compliance limits of the device. In those cases, the maximum compliance limit was used as the MCL, and gray symbols are indicated in the figures.

Psychophysical Tuning Curves

PTCs were measured using a forward masking paradigm in which the masker was a 204-msec pulse train, with the configuration fixed at a pTP fraction of $\sigma = 0.5$. The masker threshold and MCL levels were then used to set the range of possible stimulus levels for each channel. Thresholds obtained in the first experiment with a pTP fraction of $\sigma = 0.5$ determined the lower limit, whereas the measured MCL determined the upper limit.

The probe electrodes for each subject were those with the highest, lowest, and median single-channel TP thresholds (thresholds based on the largest pTP fraction possible, either 0.9 or 1). The probe, a 10.2-msec pulse train fixed at 3 dB above threshold, was presented 9.18 msec after the masker (Fig. 2). Two configurations were used for the probe: MP ($\sigma =$ 0) and the largest pTP fraction that would accommodate a current level 3 dB above threshold without exceeding the compliance limit. (For most subjects, the pTP fraction for the more restricted probe configuration was smaller than the one used to measure single-channel TP thresholds [i.e., <0.9].) The two types of probe stimuli will be referred in this article as "MP" and "pTP." The probe presentation level was calculated as 3 dB above the probe-only threshold using the same pTP fraction, where threshold was measured as described previously except that a 10.2-msec pulse train was used.

The level of the masker was adjusted to determine how much masking was required to just mask the probe. The masked thresholds were obtained using a procedure similar to the one described above for the probe, an adaptive, two-up one-down, three-alternative forced-choice tracking procedure. The masker was presented in all three intervals, whereas the probe was presented randomly, but with equal probability, in only one of the three intervals. The subject was asked to indicate which interval sounded "different." For each probe channel, the sequence of testing began with the masker presented on the same channel as the probe. The masking channel was then changed to other channels successively more distant from the probe channel. If masker threshold reached masker MCL in any condition, an indication that masking could not be achieved, this channel was determined to be the edge of the tuning curve. The apical and basal edges of each tuning curve were obtained in most conditions except when the probe channel was near the end of the electrode array. For some subjects, for completeness, if time allowed, masked thresholds were obtained for channels beyond the edge of the tuning curve (i.e., every other masker electrode) until the basal and apical ends were reached. The masker level was constrained at or below the MCL. Masker-alone threshold and MCL measurements were interleaved with PTC masking measurements to reduce the fatigue of the participants.

In four subjects, additional PTCs were obtained with the probe level fixed at 1.5 or 2 dB above the probe-alone threshold, and the probe configuration was limited to $\sigma = 0$. All other stimuli and procedures were the same as for the PTCs measured with a 3-dB probe level.

Quantifying PTCs

PTC data are shown in raw form as a function of masker level in decibels relative to 1 mA and normalized to the percentage of the masker-alone dynamic range. A unique aspect of this study is the consistent use of a restricted masker configuration of $\sigma = 0.5$, which allows for comparisons of changes in tuning properties that can only be a result of the varying probe configuration. The fixed masker configuration also allows for the analysis of data normalized to percentage of masker-alone dynamic range, which reduces the effect of channel-to-channel variability in masker-alone levels. For instance, in some cases for which the MCL was relatively high and the PTC was relatively shallow (S22 highest threshold channel), the shape of the tuning curve is hard to discern. The normalized data were used to accurately identify the tip and data points to be analyzed from the PTCs.

To compute the slopes of the apical and basal sides of the PTCs, first the tip of the tuning curve (i.e., the lowest masker level required to mask the probe) was identified using the normalized masker levels. Once the tip was identified, the level for which the curve crossed 80% of masker dynamic range was used for the endpoint in either direction. Then, using the raw data points from the tip through the endpoint, a least-square error line was obtained, and the slope was calculated from that line in decibels per millimeter. In the few cases in which the tuning curve was shallow such that the data did not fall below 80%, the minimum was used as the tip, and the curve was fit to the point at which the data reached masker-alone MCL. In addition, the depths of the PTCs were taken as the difference between the minimum and maximum masker levels measured in decibels relative to 1 mA. Finally, the widths of the PTCs reported in millimeters were measured at 50% above the tip of the normalized PTCs.

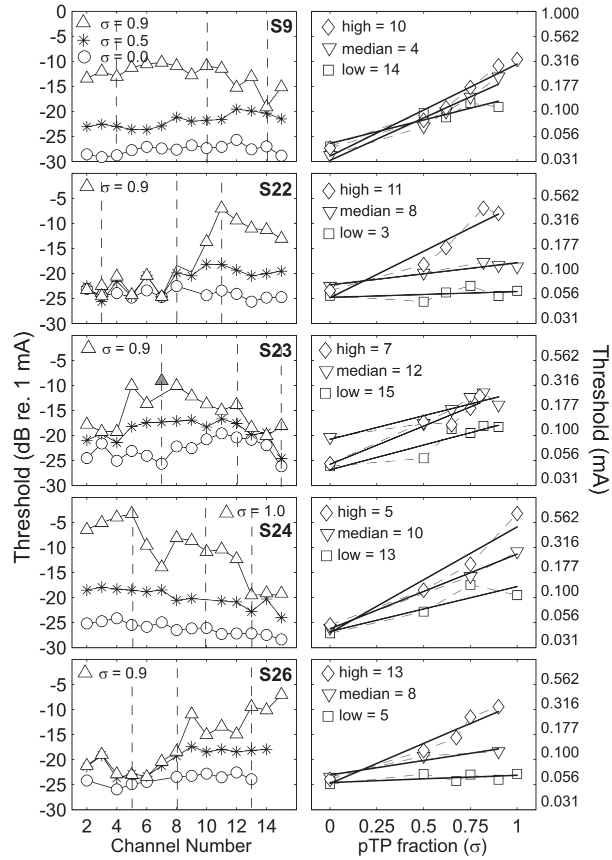

Fig. 3. Single-channel thresholds across subjects and configurations. In the left column, each panel shows the single-channel detection thresholds for a given subject (indicated in the top left corner). The abscissa represents cochlear implant channel from apical to basal, and the ordinate represents detection threshold in decibels relative to 1 mA. Electrode configuration is indicated by symbols and, for the triangles, is different for each subject. The vertical dashed lines indicate the lowest, median, and highest threshold channels obtained with the largest pTP fraction for each subject. In the right column, each panel shows the single-channel detection thresholds for a given subject as a function of partial tripolar fraction (σ) on the abscissa. Symbols indicate the stimulus channel based on the threshold for the largest pTP fraction possible (corresponding to the vertical dashed lines in the left panels). The solid lines are the least-squared error calculations for each channel.

RESULTS

Single-Channel Thresholds

Figure 3 shows individual channel threshold data for each subject (rows). The data are plotted in two ways. The left column shows the threshold across all available channels for three different pTP fractions, whereas the right column shows the thresholds for the lowest, median, and highest threshold channels as a function of pTP fraction. In the left column, the ordinate shows thresholds in units of decibels relative to 1 mA, and the abscissa shows cochlear implant channel number from apical to basal. The symbols represent three pTP fractions denoted by σ: (1) the largest fraction possible for that subject (as indicated in the figure legend, either $\sigma = 0.9$ or $\sigma = 1.0$); (2) a fraction of $\sigma = 0.5$; and (3) a fraction of $\sigma = 0$, which corresponds to MP. Note that the gray triangle for S23 indicates that the compliance limit was reached; therefore, this data point was below the true threshold for that channel. The level at the compliance limit, however, was higher than all other thresholds for this subject; therefore, it was taken as the highest threshold channel. The vertical dashed lines identify the channels selected for further testing; the channels with the highest, median, and lowest thresholds were obtained using TP. The thresholds systematically decreased as a function of pTP fraction; those obtained with TP were highest and those obtained with MP were the lowest (t test, $\sigma \geq 0.9$ versus $\sigma = 0.5$, $p < 0.05$; $\sigma \geq 0.9$ versus $\sigma = 0$, $p < 0.05$; $\sigma = 0.5$ versus $\sigma = 0$, $p < 0.05$). Channel-to-channel variability was quantified within a subject as either the standard deviation of the unsigned difference between thresholds of adjacent channels (Bierer 2007) or the mean of the unsigned difference in thresholds of adjacent channels (Pfingst & Xu 2004). Both measures were included because the two measures emphasize different aspects of the data. The variability measure using standard deviation takes into account the expected channel-to-channel differences emphasizing local variability, whereas the mean emphasizes the absolute magnitude of the differences. A

TABLE 2. Channel-to-channel variability (decibels) based on the difference in standard deviation of thresholds (top) and the mean of thresholds (bottom)

Subject	$\sigma = 0$	$\sigma = 0.5$	$\sigma = 0.9$ or $\sigma = 1.0$
S9 (SD)	1.0286	0.9495	2.6670
S9 (mean)	0.8191	0.7119	2.0124
S22 (SD)	1.3099	2.9038	3.7743
S22 (mean)	1.1291	2.2681	3.2159
S23 (SD)	2.3073	2.0740	3.6723
S23 (mean)	1.8604	1.4813	2.6450
S24 (SD)	0.7814	1.5513	3.4750
S24 (mean)	0.6260	1.0837	2.4756
S26 (SD)	1.0656	1.8880	3.3835
S26 (mean)	0.8902	1.3836	2.8406

systematic decrease in variability as a function of pTP fraction was observed, with the greatest variability for $\sigma = 0.9$ or 1.0 and the least for $\sigma = 0$ (Table 2) (channel-to-channel variability based on both standard deviation and mean; paired t test, $\sigma \leq 0.9$ versus $\sigma = 0.5$, $p < 0.005$; $\sigma \geq 0.9$ versus $\sigma = 0$, $p < 0.005$; $\sigma = 0.5$ versus $\sigma = 0$, $p > 0.05$).

In the right column of Figure 3, the data are plotted with threshold on the ordinate and pTP fraction on the abscissa. The different symbols represent the lowest, median, and highest threshold channels measured with TP. For each channel plotted, threshold generally increased systematically with pTP fraction, consistent with the expected narrowing of electrical field size. The slopes of the least-square error fits (solid lines) were computed for each of the three channels and are listed in Table 3. The channels with the lowest thresholds (measured with TP) have shallow slopes with increasing σ. In contrast, the high-threshold channels have steep slopes. Note that for subjects S22 and S26, the thresholds measured for the lowest channels remained relatively constant despite the focusing of the electrode configuration.

Forward-Masked PTCs

Forward-masked PTCs were measured for each subject and for the experimental electrodes identified by single-channel thresholds: those with the highest and lowest threshold and sometimes the median (denoted by the vertical dashed lines in Fig. 3). The masker stimulus was always presented with a pTP fraction of $\sigma = 0.5$, whereas the probe stimulus was either $\sigma = 0$ or $\sigma \geq 0.55$ as denoted in the figure legend. Figure 4 illustrates PTCs for all subjects for the lowest (left column), median (middle), and the highest (right column) pTP threshold channels. The data represent the masker level required to just mask the fixed-level probe stimulus (ordinate), across individual masker channels (abscissa), and the different probe configurations are indicated by the symbols. The solid gray lines indicate the range of possible stimulation levels for the masker stimulus ($\sigma = 0.5$), representing the masker-alone threshold (lower bound) and masker-alone MCL (upper bound). The bold lines represent the best-fit line of the data used to calculate the slope, yielding a metric to quantify the sharpness of the tuning curves. Note that for S23, the gray symbols represent the two channels for which the masker-alone MCL was above the maximum current level possible for that subject.

TABLE 3. Slope of the threshold versus fraction functions (decibels per fraction) across subjects

Subject number	Lowest threshold channel	Median threshold channel	Highest threshold channel
S9	9.39	16.89	18.39
S22	1.16	4.40	18.56
S23	9.86	9.40	15.27
S24	8.94	14.90	21.15
S25	1.45	5.75	15.96

The difference between threshold and MCL defines the dynamic range. In Figure 5, the data are plotted as a function of masker-alone percent dynamic range (threshold [0%] to MCL [100%]) to further characterize the shape of the PTCs and to facilitate comparisons across subjects and channels. This normalization of the data is appropriate when the masker stimulus is the same for all of the sets of data within a given subject. As in Figure 3, the symbols indicate the probe configurations used. The width of the PTC was measured at half maximum, as indicated by the horizontal lines. Note that for the lowest threshold channel of S23 (left), only the apical side of the tuning curve could be measured because the probe channel was the most basal channel. In this case, the width was measured from the probe channel to the edge of the curve at half-maximum and then doubled. For S24, high-threshold channel, $\sigma = 0$ probe (right panel), the curve did not achieve a level of half-maximum on the apical side of the curve. In that case, the width is measured to the end of the electrode array and likely underestimates the width. For all other cases, the curves achieved a level of half-maximum on both sides of the curve. Note that width calculations for the median channels were not included for S24 and S26 because the curves were incomplete or not obtained. Also note that for S9 highest threshold channel and S24 low and high-threshold channels, a secondary peak is observed. One possible explanation for the secondary tips is cross-turn stimulation. Because the secondary tips are always basal to the probe electrode, these tips could not be a result of stimulating fibers of passage from more apical parts of the cochlea.

Figure 6 summarizes the slope calculations of the tuning curves in Figure 4. The left and right columns plot the slopes for the apical and basal sides of the tuning curves, respectively. The negative apical slopes are inverted. The top and middle rows plot the slopes for the $\sigma = 0$ and $\sigma \geq 0.55$ configurations, respectively. Each subject is represented by a symbol, and the fill of the symbol indicates the lowest (open), median (gray), and highest (filled) pTP threshold channel (connected by dashed lines for a given subject). The slope of the tuning curves becomes progressively shallower from the lowest to highest threshold channel for both probe configurations. This trend is more pronounced for the apical than the basal side of the tuning curve. (Spearman rank correlation coefficient for the apical slope: MP probe, $r = -0.66$, $p = 0.013$; pTP probe, $r = -0.775$, $p = 0.002$. Basal slope: MP probe, $r = -0.22$, $p = 0.5$; pTP probe, $r = -0.469$, $p = 0.124$). The least-square error fits are shown by solid lines. (For the basal slopes, the line fits were performed after removing the outlying data points for subject S24. However, the correlation statistics include data from all subjects). A comparison of slopes between the two configurations is plotted in the lower row of panels, with the

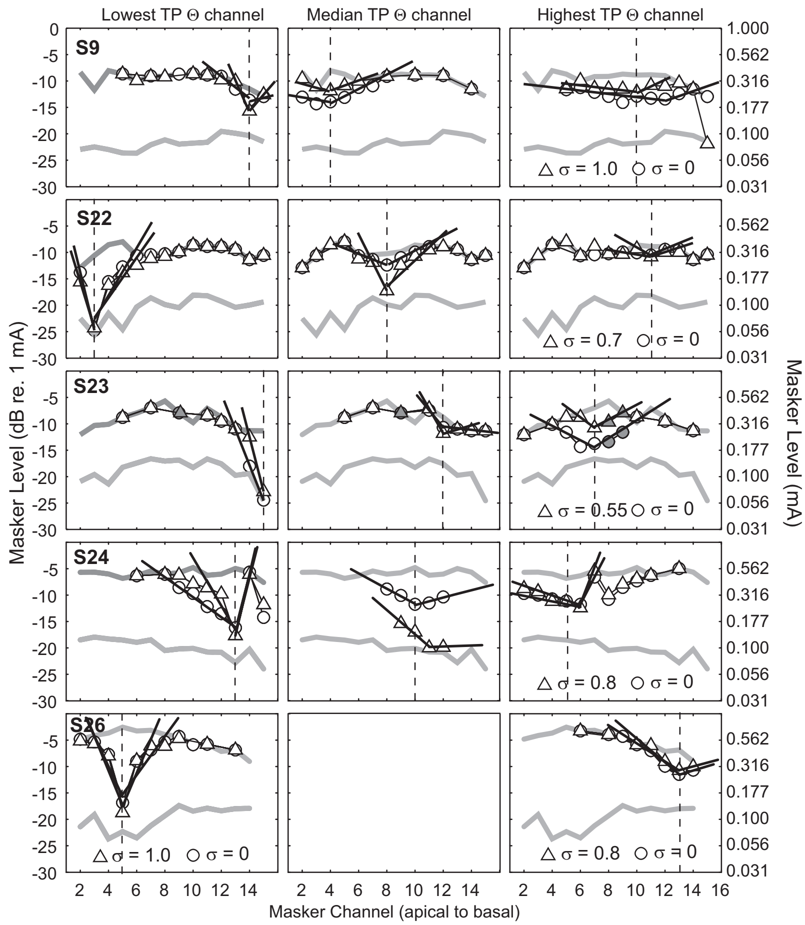

Fig. 4. Forward-masked psychophysical tuning curves (PTCs) for all subjects plotted using the raw masker levels that were measured as the masker level that just masked detection of the probe (ordinate in decibels [left] and mA [right]). Subject is indicated in the top of the left panels. The left, middle, and right panels are PTCs for the lowest, median and highest threshold channels, respectively (as indicated in Figure 3). The shaded gray lines represent masker-alone threshold and most comfortable level. Symbols indicate the probe configuration, and vertical dashed lines indicate the probe channel. The bold lines indicate the slope of the best fit line from the estimated tip of the PTC to the point at which the data crosses 80% of the masker-alone dynamic range. The lines were extended for ease of viewing.

$\sigma = 0$ data on the abscissa and $\sigma \geq 0.55$ on the ordinate. Slopes were steeper for the $\sigma \geq 0.55$ pTP condition for both the apical (Wilcoxon signed-rank test, $p < 0.05$) and basal ($p < 0.01$) sides of the PTCs.

Figure 7 shows the PTC widths (left column) and depths (right column) measured for $\sigma = 0$ (top row) and for $\sigma \geq 0.55$ (middle row). The depths of the PTCs were taken as the difference between the minimum and maximum masker levels measured in decibels relative to 1 mA. Conventions are as in Figure 6. These results show the trend that widths measured for the highest threshold channel were wider and depths were smaller than the lowest threshold channel (Spearman rank correlation coefficient for PTC width: MP probe, $r = 0.47$, $p = 0.105$; pTP probe, $r = 0.476$ $p = 0.10$. PTC depth: MP probe, $r = -0.495$, $p = 0.072$; pTP probe, $r = -0.455$, $p = 0.102$). In addition, the widths were narrower for PTCs measured with pTP compared with $\sigma = 0$ or MP (bottom row of Fig. 7; Wilcoxon signed-rank test, width, $p < 0.005$). However, no trend was observed in the depth measure of PTCs with changes in partial tripolar fraction (depth, $p > 0.1$).

Although the correlation analysis revealed statistical significance for only a subset of the PTC metrics as a function of TP threshold (see top two rows of Figs. 6 and 7), for individual subjects, the PTCs were always wider and had shallower apical slopes for the highest threshold channels (Wilcoxon signed-rank test, apical slope: MP probe, $p = 0.031$; pTP probe, $p = 0.031$. Width: MP probe, $p = 0.031$; pTP probe, $p = 0.09$). The basal slope was shallower for three of four subjects, and the PTC depth was shallower for four of five subjects for the highest threshold channels (Wilcoxon signed-rank test, basal slope: MP probe, $p = 0.125$; pTP probe, $p = 0.062$. Depth: MP probe, $p = 0.031$; pTP probe, $p = 0.09$). If the PTC metrics and thresholds are normalized relative to the mean for each subject, reducing the across subject variability, the trends described above are more pronounced. However, the data are not normalized because the relationships with the absolute TP thresholds are more clinically relevant.

PTCs with Low-Level Probe Stimuli

A second series of PTCs were obtained using a lower-level probe stimulus to clarify the possible influence of probe level on tuning properties. For example, in Figures 4 and 5, some of the PTCs were shallow, which made slope, width, and depth calculations difficult. This effect was likely related to the relatively high probe level of 3 dB above threshold; a high probe level requires higher masker levels. In this second experiment, we measured PTCs with a fixed probe level of 1.5 dB above threshold (or 2 dB for S9) for the lowest and highest threshold channels in four subjects. For this series, the probe channel was not varied but was fixed to $\sigma = 0$. Figure 8 shows each individual subject's PTC measured with the probe fixed at

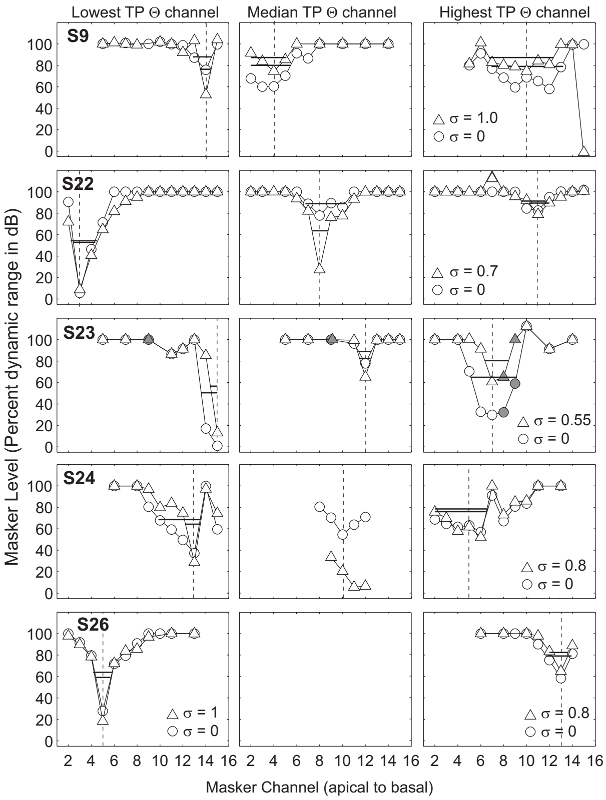

Fig. 5. Forward-masked psychophysical tuning curves (PTCs) for all subjects plotted using the masker levels relative to the percentage of masker-alone dynamic range in decibels (masker-alone MCL – masker-alone threshold) (ordinate). Conventions are as in Figure 4. Each row shows PTCs for a given subject indicated in the top of the left panels. The left, middle and right panels represent low-, median- and high-threshold probe channels, respectively. The width was measured at half-maximum indicated by the horizontal line.

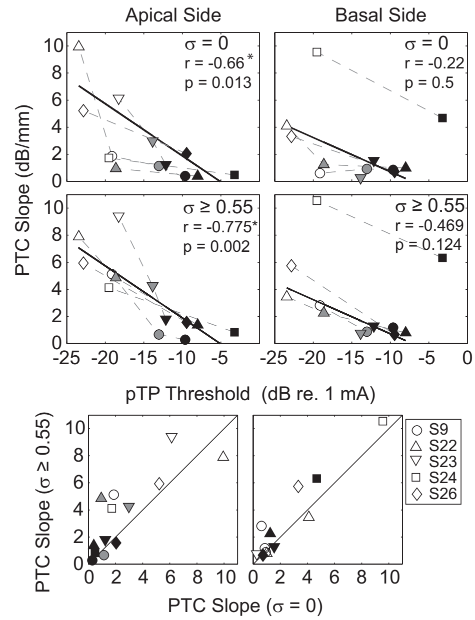

Fig. 6. Summary of PTC slope calculations. Symbols represent subject number, and fill represents threshold (open, lowest; gray, median; and black, highest). Data from a given subject are connected by a dashed line. The negative apical slopes are inverted. Top row shows the slope of the $\sigma = 0$ PTCs in decibels per millimeter (ordinate) as a function of pTP threshold for the apical (left) and basal (right) sides of the curves. The least-square error best fit line is shown in bold. Middle row shows the slope of the $\sigma \geq 0.55$ PTCs. Conventions are as in the top row. Lower row shows PTC slopes for $\sigma = 0$ (abscissa) versus $\sigma \geq 0.55$ probe (ordinate) for the apical (left) and basal (right) sides of the curves.

1.5 dB above threshold (indicated with filled circles) along with the previous PTC data collected with the probe fixed at 3 dB above threshold for comparison (unfilled circles). The results indicate that with a lower-level probe (1.5 or 2 dB above threshold), the tip of the tuning curve occurs lower in the dynamic range, meaning that less masker was required to just mask the probe. As for the previous set of PTCs, the slopes of the apical and basal sides were calculated and are shown with bold lines (see Fig. 10 for slope results).

Figure 9 represents the PTCs for lower-level probe stimuli with the data normalized to percentage of masker-alone dynamic range in decibels. In all cases but one (S23 low-threshold channel), the tip of the tuning curve is lower in the dynamic range for the 1.5-dB than the 3-dB probe level. As was done previously, the width of the PTCs was calculated at half-maximum (Fig. 10).

Figure 10 shows the summary data from the second experiment. Top panels show the apical (left) and basal (right) slopes of the PTCs for the lowest and highest threshold channels for a 1.5-dB probe stimulus. The least-square error fits are shown by the solid lines. The slopes were steeper for the lowest compared with the highest threshold channel, more so for the apical side than the basal side of the PTC (Spearman rank correlation coefficient for apical slope: $r = -0.857$, $p = 0.007$; basal slope: $r = -0.714$, $p = 0.071$; width: $r = 0.643$, $p = 0.086$; depth: $r = -0.5$, $p = 0.207$). The lower panels show the width (left) and depth (right) of the PTCs for those channels. The widths were larger for the highest threshold channel and the depths were smaller.

DISCUSSION

To summarize, the results of this study indicate that cochlear implant channels with high thresholds, when measured with a restricted electrode configuration, have relatively wide or tip-shifted PTCs (S9 and S24). This relationship between single-channel thresholds and tuning curve metrics suggests that the high thresholds observed for some channels are the result of a poor interface between those channels and nearby spiral ganglion neurons. These preliminary findings suggest that the measurement of single-channel thresholds using a restricted electrode configuration, such as TP or pTP, could be a simple yet effective clinical tool to identify impaired electrodes.

Single-Channel Thresholds

This study showed a systematic increase in threshold level and threshold variability with more focused stimulation (Fig. 3), extending the previous findings of studies comparing MP and BP (Pfingst & Xu 2004; Pfingst et al. 2004) and/or TP (Mens & Berenstein 2005; Bierer 2007). The authors of those

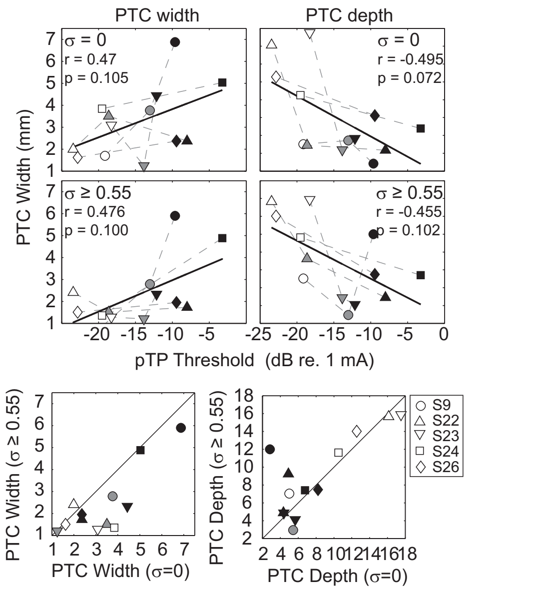

Fig. 7. Summary of PTC width and depth calculations. Symbols represent subject number and fill represents threshold (open, lowest; gray, median; black, highest). Data from a given subject are connected by a dashed line. The left column shows PTC width in millimeters and the right is PTC depth in dB. Top row shows widths and depths of the $\sigma = 0$ PTCs as a function of pTP threshold. The least-square error best-fit line is shown in bold. Middle row shows the widths and depths of the $\sigma \geq 0.55$ PTCs. Conventions are as in the top row. Lower row shows PTC widths and depths for $\sigma = 0$ (abscissa) versus $\sigma \geq 0.55$ probe (ordinate).

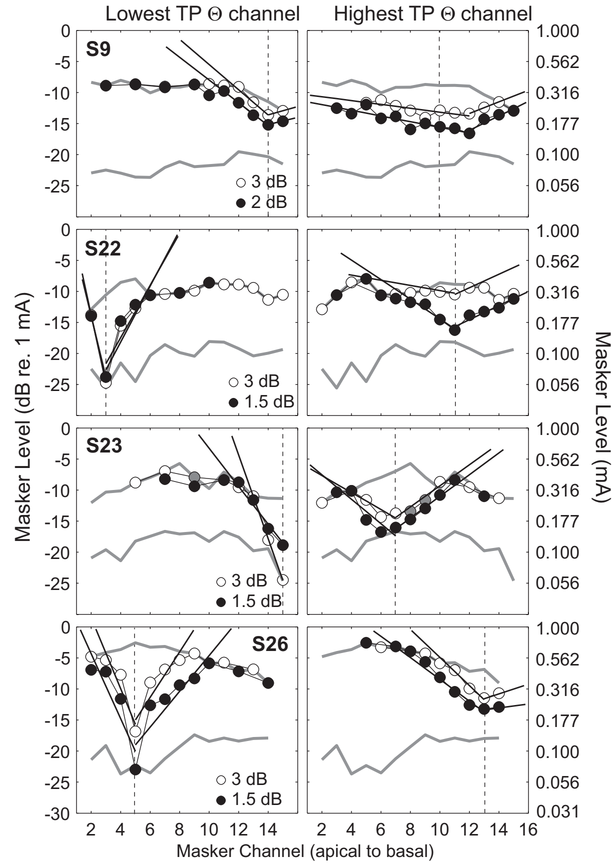

Fig. 8. Lower probe level PTCs using raw masker level in decibels relative to 1 mA. Conventions are as in Figure 4. Each row shows PTCs for a given subject indicated in the top of the left panels. The left and right panels show low- and high-threshold probe channels, respectively. The fill of the symbol represents the probe level, either filled (1.5 or 2 dB above threshold) or open (3 dB above threshold).

studies hypothesized that the electrodes with high thresholds were likely interfacing poorly with nearby spiral ganglion neurons. A close examination of the data from this study supports this theory. Specifically, it would be expected that electrodes having a good electrode-to-neuron interface would be relatively unaffected by pTP current fraction because viable neurons are sufficiently close that the electrical field does not need to be broad to achieve a threshold level of activation. Indeed, for subjects S22 and S26, threshold did not change appreciably as a function of pTP fraction for the channel having the lowest TP threshold (Fig. 3, right column). In contrast, electrodes that are not in close proximity to viable neurons would require a lot of additional current to achieve threshold (i.e., to activate more distant neurons) when using the TP configuration because the electrical field broadens slowly with current; only a small amount of additional current would be necessary for the MP configuration to activate the same distant neurons. In this study, at least half of the electrodes for each subject exhibited a substantial difference between TP and MP thresholds (Fig. 3, left column). In two subjects (S9 and S24), nearly all electrodes had this property. Thus, "good" electrode-neuron interfaces, based on the above description, were not prevalent in our population of cochlear implant subjects.

Psychophysical Tuning Curves

If a high TP threshold is indeed due to a poor electrode-neuron interface, as hypothesized above, a PTC applied to such a channel should show some deficit in spectral or spatial selectivity. Because a high-threshold probe channel would be activating spiral ganglion neurons that are displaced longitudinally along the cochlea, the best masking channels would also be displaced, leading to tip-shifted or wide tuning curves (i.e., poor selectivity). The two major findings of this study are consistent with this proposed mechanism: (1) PTCs were wider, with shallower slopes and depths, for the highest TP threshold channels; and (2) PTCs were sharper with a TP probe configuration.

Pairwise comparisons of implant channels tested in the same subject showed that the channel with the highest TP threshold consistently exhibited poorer spatial selectivity. With the more restricted pTP probe, apical and basal slopes were generally shallower, and tuning curve widths were greater for the high-threshold channel. Interestingly, these PTC properties also exhibited differences when an MP probe was used. Furthermore, when the subject data were pooled and the median-threshold channels were included, apical slope was significantly and inversely correlated with TP threshold, whereas the other tuning curve metrics showed similar trends but were not statistically significant. Together, these results show the strong relationship between high TP thresholds and low spatial selectivity, implying that such channels are interfacing poorly with nearby auditory neurons. In contrast, thresh-

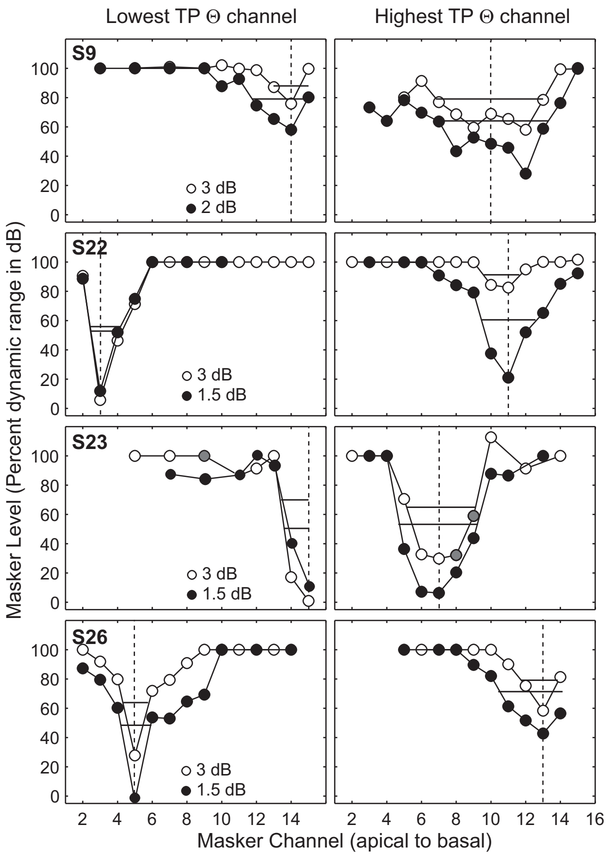

Fig. 9. Lower probe level PTCs using percentage of masker-alone dynamic range in decibels. Conventions are as in Figure 5. Each row shows PTCs for a given subject, indicated in the top of the left panels. The left and right panels represent low- and high-threshold probe channels, respectively. The fill of the symbol represents the probe level, either filled (1.5 or 2 dB above threshold) or open (3 dB above threshold).

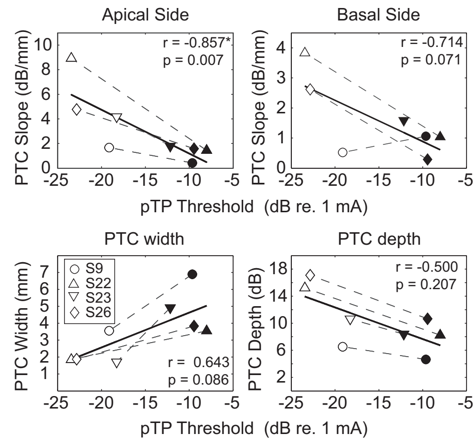

Fig. 10. Summary of lower probe level PTC slopes, widths and depths calculations. Data from a given subject are connected by a dashed line. Top row shows the apical (left) and basal (right) PTC slopes in decibels per millimeter as a function of pTP threshold (abscissa). Note that the negative apical slopes are inverted. The lower row shows the PTC width (left) and the PTC depth (right) as a function of pTP threshold (abscissa). The bold lines indicate the least-square error best fit to the data.

olds measured with the MP configuration would not be predictive of spatial selectivity because of the uniformity of MP thresholds within and across subjects.

Previous studies examining forward-masked tuning properties have reported variability across subjects and, when multiple channels were tested in individual subjects, variability among channels. A recent study by Hughes and Stille (2008) compared PTCs of channels based on their apical to basal position in the cochlea. They did not find tuning properties to be affected by channel location. Our study supports this finding. Specifically, in the five subjects who participated in this study, one subject's highest threshold channel was located basally, one was apical, and three were in the middle part of the array. Thus, the source of channel-to-channel variability in tuning properties was likely not related to cochlear position. Rather, much of the variability in this and previous studies may be explained by the nature of the electrode-to-neuron interface as assessed by TP threshold.

Another goal of this study was to show whether restricted electrode configurations produce narrower PTCs than the MP configuration, given the inconsistent findings reported by previous investigators with forward masking patterns (Shannon 1990; Boex et al. 2003; Cohen et al. 2003; Chatterjee et al. 2006; Kwon & van den Honert 2006) and PTCs (Nelson et al. 2008). A major source of variability in the previous studies may have been how the channels were selected: either one channel was chosen in the middle of the array or several channels were chosen to be distributed along the cochlea (e.g., apical, middle, or basal). As suggested by the current results, other factors of variability among channels may have obscured a modest effect of electrode configuration on the masking patterns. In this study, the channels were selected based on single-channel threshold patterns measured with TP. Significantly sharper tuning was observed for the pTP probe configuration with no observable trend with the longitudinal cochlear position. On the other hand, low-threshold TP channels showed a stronger effect of configuration than the high-threshold channels. Another difference is that in this study a potentially more restricted configuration, pTP, was used for the probe channel. In all of the previous studies mentioned above, the masker and probe configurations were varied in tandem, making comparisons of configuration effect more tentative. Because the configuration changes in this study were applied only to the probe stimulus (masker was fixed to $\sigma = 0.5$), the changes in tuning (i.e., wider tuning with MP than pTP) were primarily a reflection of probe configuration.

Acoustical Versus Electrical PTCs

Acoustic studies have used PTCs extensively to identify putative cochlear dead regions (Moore & Alcantara 2001). These studies have shown relatively wide tuning for areas of presumed cochlear dead regions as well as shifts in the location of tuning curve tips away from the probe frequency. The wider tuning curves likely reflected the loss of the active mechanism normally provided by outer hair cells. A shift in tip location was interpreted as off-place activation of the nearest area of healthy inner hair cells and/or spiral ganglion neurons (Moore & Alcantara 2001). In this study, we observed similar effects, with relatively wide electrical tuning curves obtained from channels having a putatively poor electrode-neuron interface as identified by high TP thresholds. In the case of cochlear implant listeners, we presume that the increase in width of tuning does not reflect a loss of outer hair cells. Rather, it is a result of the greater current spread necessary to activate the nearest viable neurons that may be distant from the probe electrode. Not only were PTCs wider for the highest threshold channels, but shifts in tip location were observed for two of the five subjects (S9 and S24; Figs. 4 and 5). Tip shifts did not occur for any subjects for their low-threshold channels.

In acoustic hearing, an increase in probe level results in wider PTCs because at low levels the nonlinear outer hair cells dominate the cochlear response (Huss & Moore 2003), whereas at higher stimulus levels, passive mechanical processes dominate (Nelson & Fortune 1991; Nelson & Donaldson 2001). In contrast, we did not observe a consistent effect of probe level for individual channels (Figs. 8 and 9) in support of the findings of Nelson et al. (2008). Therefore, the wider tuning properties for high-threshold TP channels cannot be explained by the relatively high probe levels required. Additional evidence against a probe level effect is apparent from a close inspection of the MP probe data. Specifically, for S9 and S26, approximately the same MP probe level was used for the low- and high-threshold TP channels; however, the PTCs in each subject were clearly wider for the high-threshold channel.

Clinical Implications

The results of this study, along with previous studies showing substantial channel-to-channel variability in single-channel thresholds, suggest that implant channels are not equivalent in sensitivity or spectral resolution. These findings may be relevant to the existing clinical techniques used to map cochlear implant channels. For example, if an audiologist could identify channels with a poor electrode-neuron interface based on TP thresholds, he or she might consider removing those channels from the patient's speech processing strategies, even if the sound-processing strategy uses the MP configuration. On close examination of the data, plotting PTC metrics as a function of TP threshold (Figs. 6, 7, and 10), it seems that a TP threshold greater than approximately −15 dB relative to 1 mA corresponds with broad PTCs. That finding could potentially be applied clinically as a screening tool to identify electrodes with poor spatial selectivity.

Furthermore, if a restricted configuration is preferred, the audiologist could optimize each channel for the most restricted pTP fraction that allows for reasonable thresholds and complete growth of loudness. Although recent studies have shown only modest improvements in spectral resolution and speech-in-noise tasks when using pTP stimulation strategies (Mens & Berenstein 2005; Berenstein et al. 2008), additional improvement might have been obtained if the pTP parameters had been optimized for each channel.

In addition to these findings, the lower current level requirements with pTP compared with true TP or BP configurations reduces one of the major limitations of using focused electrode configurations in sound processing strategies. The use of the pTP configuration also reduces the previous limitation of reduced growth of loudness by allowing stimulus levels to reach MCLs within the compliance limits of the device.

Future Directions

Measuring PTCs in a clinical setting is not practical because of time constraints in clinical practice and possible patient fatigue. However, it might be feasible to measure TP or pTP thresholds to make mapping decisions in a clinical setting. Future experiments will also explore the use of evoked potentials in the prediction of impaired cochlear implant channels.

Studies that examine spread of excitation patterns for individual channels (PTC shape or masking patterns) often do not relate those measures to speech perception (Chatterjee et al. 2006; Kwon & van den Honert 2006). Those authors suggest that a relationship is unlikely because the one or two channels tested in each subject are not representative of the implant as a whole. However, a relationship between speech perception and the tuning properties of individual channels might be found if the most and least spatially selective channel of each listener could be identified. This study suggests that single-channel TP thresholds would be an effective tool for identifying such channels. Unfortunately, speech perception was not explored in this study because all the subjects had excellent speech scores that were similar to one another (Table 1). Future studies will seek to include subjects with a range of speech perception abilities.

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Identifying Cochlear Implant Channels With Poor Electrode-Neuron Interfaces: Electrically Evoked Auditory Brain Stem Responses Measured With the Partial Tripolar Configuration

Julie Arenberg Bierer, Kathleen F. Faulkner, and Kelly L. Tremblay

Objectives: The goal of this study was to compare cochlear implant behavioral measures and electrically evoked auditory brain stem responses (EABRs) obtained with a spatially focused electrode configuration. It has been shown previously that channels with high thresholds, when measured with the tripolar configuration, exhibit relatively broad psychophysical tuning curves. The elevated threshold and degraded spatial/spectral selectivity of such channels are consistent with a poor electrode-neuron interface, defined as suboptimal electrode placement or reduced nerve survival. However, the psychophysical methods required to obtain these data are time intensive and may not be practical during a clinical mapping session, especially for young children. Here, we have extended the previous investigation to determine whether a physiological approach could provide a similar assessment of channel functionality. We hypothesized that, in accordance with the perceptual measures, higher EABR thresholds would correlate with steeper EABR amplitude growth functions, reflecting a degraded electrode-neuron interface.

Design: Data were collected from six cochlear implant listeners implanted with the HiRes 90k cochlear implant (Advanced Bionics). Single-channel thresholds and most comfortable listening levels were obtained for stimuli that varied in presumed electrical field size by using the partial tripolar configuration, for which a fraction of current (σ) from a center active electrode returns through two neighboring electrodes and the remainder through a distant indifferent electrode. EABRs were obtained in each subject for the two channels having the highest and lowest tripolar ($\sigma = 1$ or 0.9) behavioral threshold. Evoked potentials were measured with both the monopolar ($\sigma = 0$) and a more focused partial tripolar ($\sigma \geq 0.50$) configuration.

Results: Consistent with previous studies, EABR thresholds were highly and positively correlated with behavioral thresholds obtained with both the monopolar and partial tripolar configurations. The Wave V amplitude growth functions with increasing stimulus level showed the predicted effect of shallower growth for the partial tripolar than for the monopolar configuration, but this was observed only for the low-threshold channels. In contrast, high-threshold channels showed the opposite effect; steeper growth functions were seen for the partial tripolar configuration.

Conclusions: These results suggest that behavioral thresholds or EABRs measured with a restricted stimulus can be used to identify potentially impaired cochlear implant channels. Channels having high thresholds and steep growth functions would likely not activate the appropriate spatially restricted region of the cochlea, leading to suboptimal perception. As a clinical tool, quick identification of impaired channels could lead to patient-specific mapping strategies and result in improved speech and music perception.

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INTRODUCTION

Cochlear implants have provided improved hearing for individuals with severe or profound hearing loss, yet there exists substantial variability in listening abilities (e.g., Koch et al. 2004). Although some studies have shown that implant performance can be partially accounted for by etiology and duration of deafness (Gantz et al. 1993; Gfeller et al. 2008), both of which are thought to reflect spiral ganglion neuron loss, a direct correlation between global spiral ganglion neuron survival and performance on speech tests has not been established (Nadol & Eddington 2006). Nevertheless, patterns of local neuron degeneration may reduce the effectiveness of auditory nerve stimulation by individual implant electrodes, which can indirectly affect performance (Khan et al. 2005; Fayad et al. 2009). Other local factors, such as the electrode position within the cochlea and bone and tissue growth, may also have a negative impact on what we refer to as the electrode-neuron interface. Here, we examine whether electrically evoked auditory brain stem responses (EABRs) can be used to identify cochlear implant channels with a poor electrode-neuron interface.

Previous results from our laboratory and others have indicated that high variability in thresholds measured across the electrode array of individual subjects is a predictor of poor performance on speech tests (Pfingst et al. 2004; Bierer 2007; Long et al. 2010). The channel-to-channel variability was higher for focused electrode configurations, such as bipolar, tripolar (TP), and phased array, than for the broader monopolar (MP) configuration. These results are consistent with focused configurations having a greater sensitivity to the electrode-neuron interface, which we define as an implant channel's ability to activate the auditory nerve. Factors such as the radial distance between the implant electrodes and the osseous spiral lamina, the numbers and distribution of viable spiral ganglion neurons, the stimulability of those neurons (including the role of peripheral processes), and bone and tissue growth within the scala tympani may all impact the electrode-neuron interface. In a recent cochlear implant model of focused stimulation (Goldwyn et al. 2010), channels having a poor electrode-neuron interface had higher thresholds and produced broader activation patterns of auditory neurons along the cochlea. Such an association between high thresholds and broader spatial activation has been corroborated behaviorally using psychophysical tuning curve (PTC) measures (Bierer & Faulkner 2010). Broad spatial activation could result in the distortion of spectral cues because a signal delivered by a degraded channel may activate neurons beyond the targeted population of neurons. Therefore, identification of channels with a poor electrode-

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neuron interface might provide some insight into the variability in speech perception abilities across implant listeners.

EABRs may be a sensitive tool for identifying channels with a poor electrode-neuron interface. Previous studies in animal models have shown a relationship between the degree of spiral ganglion loss and evoked potential measures (Smith & Simmons 1983; Hall 1990; Shepherd et al. 1993; Miller et al. 1994, 2008). In those studies, the aspect of EABRs having the strongest correlation to neural loss was the slope of the amplitude growth function, assumed to be proportional to the number of responding nerve fibers. It is unclear what impact localized spiral ganglion cell loss might have had on the EABRs in these studies, because only global cell counts were made. However, in one of the studies, a significant correlation was measured only when the implant electrodes were placed relatively close to the modiolus (Shepherd et al. 1993). These findings suggest that EABRs may be sensitive to spiral ganglion loss and electrode-to-neuron distance, two important aspects of the electrode-neuron interface. Our modeling results (Goldwyn et al. 2010) suggest that both of these factors (a large radial distance and a reduced local neuron count) have comparable influences on the spatial pattern of activated auditory neurons over a range of current levels, i.e., both result in patterns that are relatively broad at threshold and widen relatively faster with increases in current. Because the amplitude growth of the EABR with current reflects, in part, the recruitment of peripheral auditory neurons, the slope of this function should be greater for channels affected by a poor electrode-neuron interface.

In this study, EABRs were measured on channels suspected of having a good or poor electrode-neuron interface based on behavioral criteria established in our previous study (Bierer & Faulkner 2010). Specifically, behavioral thresholds were measured across channels using the focused TP configuration, and the channels having the lowest and highest thresholds—hypothesized to have a good and poor electrode-neuron interface, respectively—were chosen for EABR testing. The perceptual and EABR measurements were made using both the MP configuration and the partial TP (pTP) configuration, a hybrid of the MP and TP configurations in which a fraction of current from the active electrode returns through the adjacent pair of intracochlear electrodes and the remainder through an extracochlear electrode (Mens & Berenstein 2005; Litvak et al. 2007). A fraction of zero is equivalent to MP, while a fraction of one is TP. The slopes of the EABR amplitude growth functions were compared across channels and configurations. In addition, growth functions were related to the behavioral thresholds and, for a subset of subjects, PTCs obtained in a previous study (Bierer & Faulkner 2010). We hypothesized that channels with higher EABR thresholds would have steeper growth functions, consistent with our computer model (Goldwyn et al. 2010) and preliminary results demonstrating steeper loudness growth for channels with high behavioral thresholds (Nye & Bierer 2010). We also hypothesized that EABR amplitude functions would grow more gradually for a focused pTP configuration than for the MP configuration, based on previous comparisons of MP and bipolar stimulation (Brown et al. 1996).

MATERIALS AND METHODS

The electrode configurations used in this study include MP, TP, and pTP. The MP configuration consists of an active intracochlear electrode and an extracochlear return electrode. The TP electrode configuration consists of an intracochlear active electrode, with the return current divided equally between each of the two nearest flanking electrodes (Jolly et al. 1996). The pTP configuration is also formed from three adjacent electrodes, but only a fraction of the return current, denoted by "σ," is delivered to the flanking electrodes, with the remainder flowing to the distant extracochlear ground. Therefore, a σ of 0 is equivalent to MP (all current is directed to the extracochlear electrode), and a σ of 1 is equivalent to TP (no current is directed to the extracochlear electrode). Data analysis and plotting were performed in units of decibels relative to 1 mA. Compared to a linear scale, the decibel scale can better accommodate the large differences in current level requirements among configurations and across channels and subjects.

All stimulation levels used in these experiments were within the compliance limits supported by the implant. Based on the impedance values, the maximum compliance of active and return electrodes was calculated at the beginning of each test session based on an algorithm provided by Advanced Bionics Corp. (2010, written personal communication). The compliance limit was defined as the maximum voltage supported by the device (8 V) divided by the impedance.

Subjects

Six adult cochlear implant listeners, five females and one male, participated in the study. All of the participants had been implanted with the HiFocus 1J electrode array, with center-to-center electrode distance of 1.1 mm, and the HiRes90k receiver-stimulator (Advanced Bionics Corp., Sylmar, CA). The six subjects ranged in age from 30 to 79 yrs, were native speakers of American English who had become deaf postlingually, and had at least 9 mos of experience with their implants. Pertinent subject demographics are shown in Table 1, including current age, gender, duration of severe hearing loss, known etiology,

TABLE 1. Subject demographics

Subject Number	Sex	Age (yrs)	Duration of Severe Hearing Loss (yrs)	Duration of CI Use	Etiology of Hearing Loss	CNC Words in Quiet (65 dB)	CNC Phonemes in Quiet (65 dB)
S9	F	64	24	3 yrs	Hereditary	48%	75%
S22	F	67	12	9 mos	Hereditary	68%	76%
S24	F	74	2	1 yr	Unknown	54%	73%
S26	F	30	2	20 mos	Atypical Ménière's disease	56%	78%
S29	M	79	32	2 yrs	Noise exposure	60%	78%
S30	F	45	29	16 yrs (2 yrs)*	Hereditary	50%	75%

** S30 underwent reimplantation after a failure of the initial device and has 2 yrs experience with the reimplanted device.*

and speech perception scores. The speech scores, based on the CNC words test at 65 dB SPL-A, were obtained in our laboratory within 3 mos of the beginning of the experiment. The behavioral and EABR testing required three to six visits to the laboratory, with each session lasting from 3 to 6 hrs. Each participant provided written consent, and experiments were conducted in accordance with guidelines established by the Human Subjects Division of the University of Washington.

Behavioral Measures

Two sets of behavioral thresholds were obtained in this study. First, thresholds for all electrodes were measured with pulse trains in the TP configuration. The channels with the highest and lowest TP thresholds served as the experimental electrodes for the remainder of the study. An additional set of thresholds at a lower pulse rate were obtained using the MP configuration and a pTP configuration with an intermediate current fraction. This second set of thresholds allowed for direct comparisons between behavioral and EABR thresholds.

TP Thresholds to Pulse Trains • For each subject, thresholds were obtained for a 204-msec pulse train at a rate of 918 pulses per second for each electrode using the TP configuration (for details see Bierer & Faulkner, 2010; a pTP current fraction of $\sigma = 0.9$ was necessary for three subjects to remain within the voltage compliance, but for simplicity we consider these data as "TP thresholds"). Pulse train thresholds for other configurations (MP, pTP with $\sigma = 0.5$) were also analyzed when those data were available from our previous study. The lowest and highest threshold channels obtained with the TP configuration for each subject were identified for further testing. All thresholds were measured with an adaptive two-down one-up, three-interval, three-alternative forced-choice procedure, which converges on the 70.7% correct point on the psychometric function (Levitt 1971). Each run started at a suprathreshold level, and subjects responded using a mouse to indicate the interval that contained the signal. Twelve reversals (i.e., changes in the direction of the signal level) were measured for each trial, and the levels for the last eight were averaged and taken as threshold. For the first two reversals, the signal was adjusted in 2 dB steps; for the other 10 reversals, it was adjusted in 0.5 dB steps. If the threshold of two runs differed by more than 1 dB, a third run was collected, and data from all three runs were averaged. The SD for the last eight reversals from each run was measured; if the value exceeded 1 dB, the subject was encouraged to take a break, and those data were not included and threshold was subsequently remeasured. The total number of trials per run was limited to 75.

All available channels were tested in this manner. Channels were deemed unavailable if they were not programmed in the patient's clinical map because they were outside the cochlea or elicited a nonauditory or intolerable percept. Channels that were deactivated were also not included as return electrodes for the TP or pTP configurations.

MP and pTP Thresholds to Low-Rate Pulses • Thresholds on the test channels were obtained using the same stimuli and methods used during EABR testing. The stimuli were ten 102 μsecs/phase, biphasic, charge-balanced (cathodic phase first on the active electrode) current pulses presented at a rate of 11.1 pulses per second. Both the MP and pTP configurations were used. For each subject, the pTP fraction selected was one, allowing for a reasonable growth of loudness from threshold to most comfortable level (MCL) or, if MCL could not be reached, to a level corresponding to at least a report of "3" on the loudness scale, as described in the subsequent section.

Most Comfortable Level • The MCL for each test channel was determined by presenting a suprathreshold version of the EABR stimulus (ten 102 μsecs/phase, biphasic, charge-balanced current pulses presented at 11.1 pulses per second; pTP fraction was dependent on the subject) and asking the subject to adjust the level by clicking one of two options labeled "up" and "down." The subject was asked to set the level to the subjective rating of "loud but comfortable," corresponding to 7 on the 1 to 10 clinical loudness rating scale (0, off; 1, just noticeable; 2, very soft; 3, soft; 4, comfortable but too soft; 5, comfortable but soft; 6, most comfortable; 7, loud but comfortable; 8, loud; 9, upper loudness limit; 10, too loud; Advanced Bionics, 2003). The level was changed in 1 dB steps until the subject clicked the "down" button; thereafter it was changed in 0.5 dB steps. At least two runs were collected and averaged for each MCL condition. If the two measured MCLs differed by >1 dB, a third run was completed, and all three runs were averaged.

Psychophysical Tuning Curves • PTCs were obtained in four of the six subjects for whom the lowest and highest threshold channels were not at the end of the electrode array (S9, S22, S24, and S26). Methods used to obtain and quantify PTC results are described in detail in Bierer and Faulkner (2010). PTCs were briefly measured using a forward masking paradigm in which the masker was a 204-msec pulse train with the configuration fixed at a pTP fraction of $\sigma = 0.5$. Thresholds and MCLs were measured to the masker stimulus (204-msec pulse train, $\sigma = 0.5$) to set the upper and lower limits for stimulation for each channel. The probe, a 10.2-msec pulse train fixed at 3 dB above the probe-alone threshold, was presented 9.18 msecs after the masker. The level of the masker was adjusted to determine how much masking was required to just mask the probe. Masked thresholds were obtained using the same adaptive forced-choice tracking procedure as described earlier. The masker was presented in all three intervals, while the probe was presented randomly, but with equal probability, in only one of the three intervals. The subject was asked to indicate which interval sounded "different."

Quantifying PTCs • A unique aspect of the Bierer and Faulkner study was the use of a focused masker configuration of $\sigma = 0.5$, which allowed for comparisons of changes in tuning properties that are primarily a result of the varying probe configuration. To compute the slopes of the apical side of the PTCs, the tip of the tuning curve (i.e., the lowest masker level required to mask the probe) was first identified using normalized masker levels. Once the tip was identified, the level at which the curve crossed 80% of masker dynamic range was the endpoint in the apical direction. Then, using the raw data points from the tip through the endpoint, a least-square error line was obtained and its slope was calculated in decibels per millimeter. In the few cases where the tuning curve was shallow, such that the data did not fall below 80%, the minimum was used as the tip, and the line was fit to the point where the data reached masker-alone MCL.

Electrically Evoked Auditory Brain Stem Responses

The stimuli were 102 μsec/phase, biphasic, charge-balanced (cathodic phase first on the active electrode) current pulses. Subjects were seated in a sound-attenuated booth and asked to rest quietly or sleep with their eyes closed during each 3- to

4-hr test session. The stimuli were delivered to the implant using a clinical interface controlled by the Bionic Ear Data Collection System, version 1.15.158 (BEDCS, Advanced Bionics Corp., Sylmar, CA) through a dedicated Platinum Series Processor. This PC-based system controlled the timing of the stimulus presentations and delivered an external trigger to the evoked potential recording system (Neuroscan™). The amplifier was transiently blocked for 1 msec poststimulus onset to prevent saturation from stimulus-related artifacts. Ag-AgCl electrodes were placed on the vertex (Cz, active), the mastoid of the ear contralateral to the implanted ear (M1/M2, reference), and the forehead (ground). The impedances of all recording electrodes were <2 kOhm. Recordings were band-pass filtered with a low-frequency cutoff of 100 Hz and a high-frequency cutoff of 3000 Hz.

EABRs were measured in response to both MP and pTP stimulation, at a rate of 11.1 pulses per second, averaged over 1500 sweeps. Because the aim of the experiment was to evaluate the amplitude growth function, stimulus levels ranged from the behaviorally measured MCL to below threshold, in at least five steps. Initial stimulation levels were at or near MCL and were decreased in large steps, until no response was detected, and additional levels were collected as time allowed. Two runs were collected for each stimulus level, with a third if movement was detected during online data collection. Preliminary data collection indicated that MP responses were more reliable, so MP was generally tested first to ensure that each subject had recordable responses. As a control, a no-stimulus run was also collected.

Postprocessing • All artifacts exceeding ±15 mV were rejected from averaging. The recording window spanned 17 msec, 2-msec prestimulus time and 15-msec poststimulus time. After artifact rejection (±15 mV), the remaining sweeps were averaged, and 5-point smoothing was applied. The recordings were batch-processed and converted to an ASCII format for further evaluation in Matlab (MathWorks, Natick, MA).

Wave V peaks and troughs were identified visually based on the latency, repeatability, and decrease in amplitude with decreasing stimulus level. Each waveform was compared with that generated with a no-stimulus control run. Amplitudes were calculated based on the difference (in μV) between the positive peak and the following trough for Wave V. To estimate EABR threshold, the amplitude growth function of Wave V was interpolated to 100 points. The threshold was defined as the lowest current level for which the Wave V amplitude was at least 0.1 μV. These interpolated data were also used to estimate the slope of the growth function between threshold and the highest level tested for each condition.

RESULTS

Figure 1 displays the initial set of behavioral thresholds measured at the higher pulse rate (918 pulses per second) for all subjects. Thresholds were obtained with TP configuration for all available channels (σ = 0.9 or 1, as indicated) and pTP (σ = 0.5) and MP (σ = 0) configurations for some of the channels, as indicated by the symbols. Each panel represents data for one subject (denoted by the subject number in the top right or top left of the panel) as a function of cochlear implant channel number from apical to basal. The vertical dashed lines indicate the channels with the lowest and highest TP thresholds that were chosen for electrophysiological testing. As in previous studies, thresholds generally increased as the configuration became more focused, i.e., as the current fraction σ increased (Bierer & Faulkner 2010). The channel-to-channel variability of threshold also increased with current fraction.

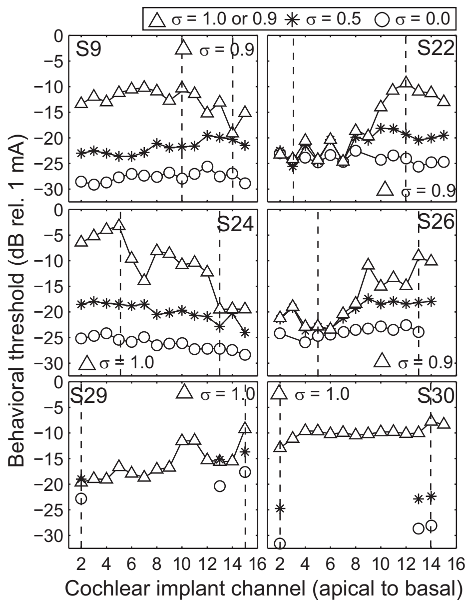

Fig. 1. Single-channel behavioral thresholds across subjects and configurations. Each panel plots the single-channel detection thresholds for a given subject (indicated in the top corner). The abscissa represents cochlear implant channel from apical to basal, and the ordinate represents detection threshold in decibels relative to 1 mA. Electrode configuration is indicated by symbols and for the triangles it is either 0.9 or 1. The vertical dashed lines indicate the lowest and highest threshold channels obtained with the largest pTP fraction for each subject (0.9 for S9, S22, and S26 and 1.0 for S24, S29, and S30).

Complete EABR amplitude growth functions were measured for each subject for the two test channels in both the MP (σ = 0) and pTP conditions (σ ranging from 0.5 to 0.8). For each channel, the largest pTP fraction was used that provided some growth of loudness before the compliance voltage limit was reached. Example EABR waveforms from one subject for the channels with the lowest (A and B) and highest (C and D) TP thresholds are shown in Figure 2. For this example, MCL was reached for all stimulus conditions. The Wave V peak-to-trough amplitude was measured for each stimulus condition and is indicated by the filled triangles for the highest tested levels. Clear responses were obtained for the highest levels tested, and the response amplitude decreased as lower stimulus levels were used (top to bottom). In this case, the response amplitude at MCL is smaller for the high-threshold channel for both electrode configurations tested (Figs. 2C and D).

The peak-to-trough amplitudes of Wave V as a function of stimulus level are plotted in Figure 3 for all subjects (top to bottom) for the lowest (left) and highest (right) threshold channels. The MP configuration is represented by black, with either open (low channel) or filled (high channel) symbols. The pTP configuration is represented by gray, with either striped (low channel) or filled (high channel) symbols. The pTP fraction used is indicated in the top of each row, and the numbers next to the symbols in the legend correspond to the

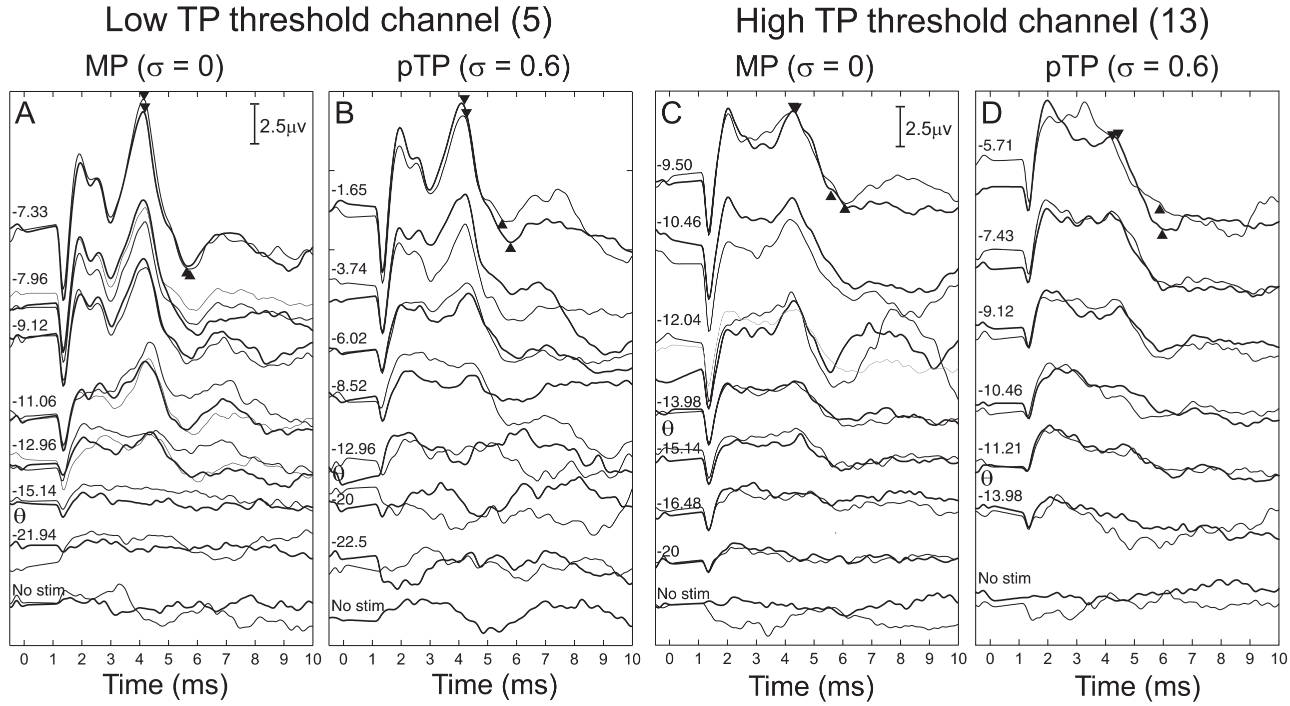

Fig. 2. Electrically evoked auditory brain stem response (EABR) waveforms for the lowest (A, left two panels) and highest (B, right two panels) TP threshold channels. The scale bar in microvolts, and the abscissa is poststimulus time in milliseconds. Within each panel, the EABR waveform is plotted with increasing stimulus levels from 0 current (bottom) to the most comfortable level (top) indicated at the left of each waveform. There are two waveforms plotted for each stimulus representing the replication of each waveform in black and gray lines. Within (A) and (B), the left panels show responses to the MP configuration while the right panels in (C) and (D) show responses to pTP stimuli with a fraction of $\sigma = 0.6$. Data are from S26.

subjects' 1 to 10 loudness ratings of the highest level tested, as described in the Materials and Methods section. The slope of each amplitude growth function was calculated from the least-square error line (shown in bold) fitting the data between the estimated threshold (stimulus level indicated by arrowhead below the abscissa) and the highest tested level.

The growth function slopes of low-threshold channels increased more gradually for the pTP than the MP configuration, which is consistent with a previous study (Brown et al. 1996). In contrast, the opposite effect can be seen for the high-threshold channels, such that the growth functions were often steeper for the pTP configuration. The difference in growth function slope between MP and pTP configurations is plotted for the low (diagonal stripes) and high (stippled) threshold channels for each subject in Figure 4A. For this analysis, a positive number indicates the expected result that slopes with the MP configuration are steeper than those with the pTP configuration. For the channels with the lowest TP threshold, the difference in slopes was positive for all six subjects, while for the channels with the highest TP threshold, the difference in slopes was negative for five of the six subjects (Wilcoxon signed rank test comparing channels with low and high thresholds, $p < 0.05$). Figure 4B displays the difference in growth function slope for the high- and low-threshold channels for the MP (open) and pTP configurations (gray filled) for each subject. A positive number indicates that the slope was steeper for the high-threshold channel, which was the finding with the pTP configuration for all six subjects. The distribution of results was variable for the MP configuration. A pairwise comparison of channels with low and high TP behavioral thresholds indicates a difference in amplitude growth functions for those channels (Wilcoxon signed rank test comparing slopes for high and low channels with MP and pTP configurations, $p < 0.05$).

EABR thresholds were compared with behavioral thresholds measured with the same low-rate pulse trains. Figure 5 plots behavioral thresholds (abscissa) as a function of EABR thresholds (ordinate) for the MP (left) and pTP (right) configurations. As in previous figures, the subject, stimulus configuration, and behavioral threshold status are represented by the symbol, color, and fill, respectively. The EABR thresholds for the high- and low-threshold channels for each subject were not correlated, so both data points were included in the regression analysis (Spearman's rank correlation: MP, $r = 0.26$, $p = 0.65$; pTP, $r = 0.14$, $p = 0.8$). Across subjects, a statistically significant correlation was measured between these two threshold estimates for both electrode configurations (Spearman's rank correlation: MP, $r = 0.650$, $p = 0.02$; pTP, $r = 0.881$, $p < 0.001$).

Four of the six subjects had previously participated in the study by Bierer and Faulkner (2010), which measured PTCs for the lowest and highest threshold channels. (PTCs were not measured in the remaining two subjects because both the lowest and highest threshold channels were at the end of the electrode array [S29 and S30].) That study demonstrated that the behavioral TP thresholds (i.e., the actual threshold current levels, not their categorical high/low status) were correlated with the sharpness of tuning, as quantified by the apical slope

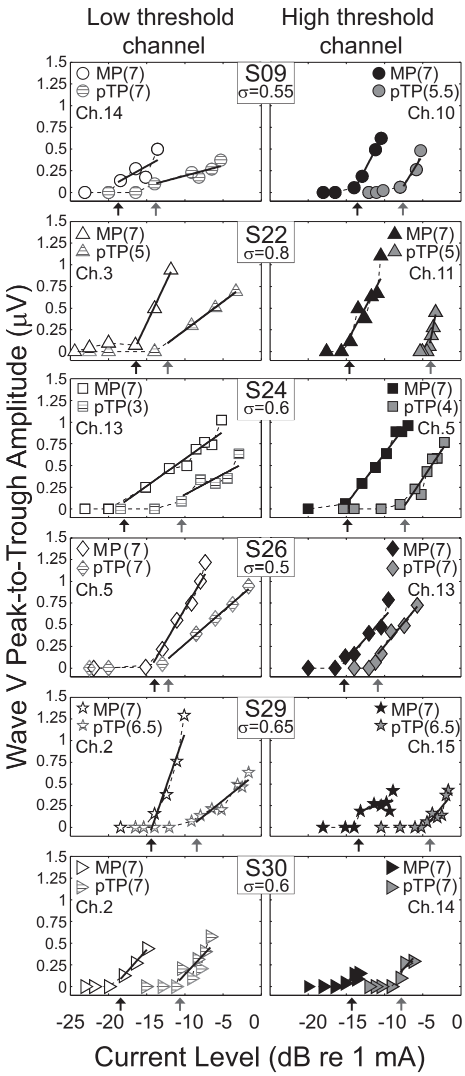

Fig. 3. Wave V amplitude growth functions for each subject and configuration. Amplitude was measured for Wave V of the EABR from the peak to the following trough in microvolts (ordinate) and is plotted as a function of stimulus level in decibels (abscissa). The left and right columns are for the channels with low and high TP thresholds, respectively. Electrode configuration is indicated by black for MP and gray for pTP stimulus configurations. Data from the high- and low-threshold channels are indicated by the fill of the symbol (open or striped for the low-threshold channels and filled for the high-threshold channels). The pTP fraction used for each subject is indicated in the top of each row. In the legend for each panel, the number in parentheses indicates the subjective rating of the loudness of the highest level tested for each stimulus configuration and channel. EABR threshold was taken as the level for which the amplitude growth function reached 0.1 μV, and the current level at threshold is indicated by arrows below the abscissa. A least-square error best-fit line is shown in bold from threshold to the highest tested level.

of the curves. Figure 6 indicates that the same relation holds with EABR thresholds. In this figure, the EABR thresholds were normalized by subtracting the average of the two test channels. The apical slope of the PTC (ordinate) is plotted as a function of relative EABR threshold (abscissa) for the MP (left) and pTP configurations (right). The PTC slopes for the

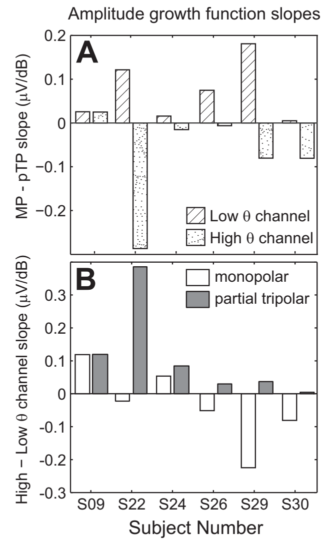

Fig. 4. A, The difference between amplitude growth function slopes (μV/dB) for monopolar and partial tripolar stimuli (abscissa) are shown for each subject for the low (diagonal stripes) and high (stippled) threshold channels. B, The difference between amplitude growth function slopes (μV/dB) for high- and low-threshold channels (abscissa) are shown for each subject for the MP (open) and pTP (gray filled) configurations.

high- and low-threshold channels for each subject were not correlated, so both data points were included in the regression analysis (Spearman's rank correlation: MP, $r = 0.60$, $p = 0.42$; pTP, $r = 0.0$, $p = 1$). As with behavioral thresholds, higher EABR thresholds measured with the more focused configuration were predictive of broader tuning (Spearman's rank coefficient $r = -0.762$, $p = 0.028$). The same trend exists for the MP data, but the two measures were not correlated to statistical significance (Spearman's rank coefficient $r = -0.619$, $p = 0.12$).

DISCUSSION

The results of the present study suggest that, with spatially focused stimulation, EABR measures are sensitive to the same underlying factors that result in high- and low-behavioral thresholds. As expected, behavioral and EABR thresholds were

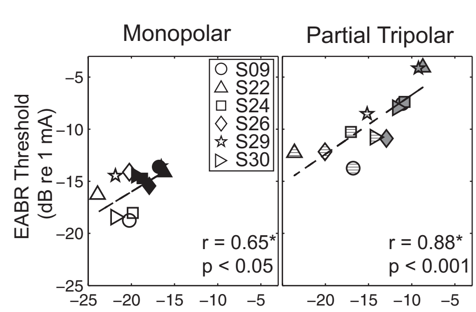

Fig. 5. Relation of behavioral threshold (abscissa) and EABR threshold (ordinate) for MP (left) and pTP (right) stimuli. As in previous figures, the symbol and fill of the data indicate the subject and the high- or low-threshold status of the channel, respectively. The solid line indicates a least-square error best-fit line to the data. The statistics are based on the nonparametric Spearman's rank correlation coefficient.

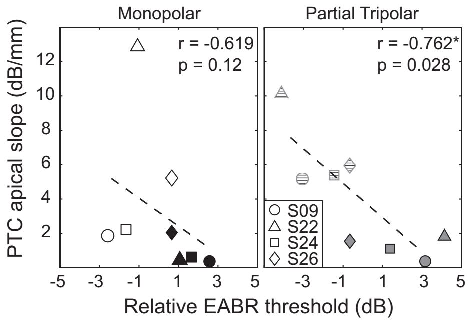

Fig. 6. Relation of the relative EABR threshold and sharpness of psychophysical tuning curves. EABR thresholds are relative to the average for each subject and are plotted on the abscissa. The apical slope of the psychophysical tuning curve in dB/mm is plotted on the ordinate. Symbol and fill indicate subject and electrode configuration. The solid line indicates a least-square error best-fit line to the data. The statistics are based on the nonparametric Spearman's rank correlation coefficient.

strongly correlated for both the MP and pTP configurations. However, only with the pTP configuration did the steepness of the EABR growth function depend on the behavioral classification of low- and high-threshold (TP) channels. In addition, in subjects for whom PTCs were obtained, these same high-threshold channels exhibited relatively broad tuning. As discussed later, both the steeper EABR amplitude growth and broader psychophysical tuning imply a more extensive tonotopic activation of the cochlea. These results suggest that the measurement of behavioral or evoked potential thresholds using a focused electrode configuration could be an effective clinical tool to identify channels affected by a poor electrode-neuron interface.

Relation to Previous Studies

Consistent with previous findings, the estimates of EABR threshold were predictive of behavioral threshold when the same stimulus parameters were used, especially for the pTP data (Brown et al. 1996). Generally, behavioral thresholds were slightly lower than EABR thresholds of individual channels, a finding that has also been shown previously (Brown et al. 2000). Although studies have demonstrated qualitative differences in EABR morphology based on apical to basal place of stimulation (Gallégo et al. 1996; Firszt et al. 2002; Thai-Van et al. 2002), no such differences were noted in the present set of experiments, because we did not systematically choose the test electrodes based on cochlear location. We did, however, observe morphological differences between channels with low- and high-TP thresholds.

Previous comparisons of amplitude growth functions using the bipolar and MP configurations have reported shallower slopes for the more focused stimulus (Schindler et al. 1977; Black & Clark 1980; Marsh et al. 1981; Abbas & Brown 1991; Brown et al. 1996). This was not a general finding in the present study. Although we did observe shallower growth functions for the focused pTP configuration with the low-threshold test channels (six of six subjects), the high-threshold TP test channels were actually steeper (five of six subjects). This suggests a fundamental difference between channels having a high and low TP threshold, which supports our hypothesis that TP threshold reflects a channel's local interface with nearby neurons.

Two primary factors of the electrode-neuron interface are the position of the electrode in the cochlea and spiral ganglion neuron survival. Both factors can influence the TP and pTP thresholds that we measured behaviorally and physiologically. However, a number of animal studies indicate that these factors may have different effects on evoked potential growth. For instance, when electrode-neuron distance was systematically increased by moving the stimulating electrode toward the lateral wall of the scala tympani, EABR thresholds increased and growth functions became steeper (Shepherd et al. 1993). This finding is consistent with recent computer models showing that a large electrode-neuron distance, which results in broader tonotopic activation, leads to a faster recruitment of neurons with increasing current (Litvak et al. 2007; Goldwyn et al. 2010). On the other hand, animal models of spiral ganglion survival have shown that reduced neuron survival results in elevated EABR thresholds but *shallow* amplitude growth functions (Hall 1990; Shepherd et al. 1993). That is, the two types of electrode-neuron interface factors, both leading to high thresholds, nevertheless had opposite effects on amplitude growth. However, we believe that in these animal experiments, the pattern of spiral ganglion loss was global, because deafness was induced by an ototoxic drug administered uniformly throughout the basal turn of the cochlea (Hall 1990; Shepherd et al. 1993). If this is true, then the broader tonotopic activation resulting from the higher current requirement was offset by the overall loss of neurons, causing slower neural recruitment and, therefore, a shallower growth function. In contrast, it is likely that the human subjects in the present study had a combination of global and localized neuron loss, as suggested by the high and variable TP thresholds across channels.

The implication that high TP thresholds reflect suboptimal stimulation of a localized region of the cochlea is supported by the PTC data measured in a subset of the subjects. The tuning curves had broader tuning for the test channels with high TP thresholds (Bierer & Faulkner 2010). A tip shift was also observed in some cases, which occurs when the degree of masking is greatest on a channel that is different than the probe channel. Analogous to acoustic studies of cochlear dead regions (Moore & Alcantara 2001), the tip shifts provide further evidence that the high-threshold test channels were near regions of low-functioning auditory neurons. Importantly, the differences between the tuning curve properties for low- and high-threshold channels were more evident when a focused pTP configuration was used for the probe stimulus rather than the MP configuration (Bierer & Faulkner 2010). Likewise, with the present EABR data, differences in threshold and amplitude growth between the low- and high-threshold test channels were also more pronounced with pTP stimulation.

Further support for a local neuron factor contributing to the differential effect of configuration on EABR amplitude growth comes from a recent computer modeling study. In that model, imposing a discrete region of spiral ganglion neuron loss adjacent to an electrode resulted in an elevated threshold and a steeper growth of neural recruitment with current (Goldwyn et al. 2010; see also Bierer 2010). These effects were most evident for focused configurations. In particular, neural recruitment for the MP configuration was relatively invariant to electrode placement or the spatial extent of local neuron loss. Thus, the inconsistent effects of the MP configuration on

EABR growth observed in the present study may be a reflection of this insensitivity to the local electrode-neuron interface.

The present study, as well as our previous modeling and psychophysical efforts, has primarily explored the spatial/spectral aspects of the electrode-neuron interface. However, temporal factors may also be important. For example, animal studies have shown that the surviving spiral ganglion neurons after a period of deafness may exhibit degraded temporal fidelity and longer refractory periods (Shepherd & Javel 1997; Shepherd et al. 2004; Vollmer et al. 2005). Also, there is recent evidence of a correlation between degraded spatial and temporal resolution in cochlear implant listeners (Chatterjee & Yu 2010). Therefore, to better understand the present findings in the context of underlying variability in the electrode-neuron interface, future studies should explore the contribution of temporal factors.

Clinical Implications and Future Directions

The results of the present study, along with those of previous behavioral studies, suggest that the individual channels of a listener's cochlear implant array do not activate nearby spiral ganglion neurons equally well (Bierer & Faulkner 2010). If a clear criterion could be developed as to what constitutes an ineffective channel—for example, one that considers channel selectivity, loudness growth, speech perception—the identification of such channels could lead to an improved clinical mapping strategy, whereby each channel is programmed based on the presumed nature of the electrode-neuron interface.

Part of the motivation for using EABR measures in this study was to determine whether such a tool would be sensitive enough to identify channels with a poor electrode-neuron interface. Although our study suggests that EABRs could be used for this purpose, it has also revealed practical limitations with this procedure. First, the ability to measure complete growth functions for multiple electrodes is more time consuming than a typical clinic visit. EABR testing for each electrode and each configuration required approximately 2 to 3 hrs. To test all of the electrodes in this manner would be prohibitive in the clinic, especially for young children. For this reason, future experiments will explore the use of the electrically evoked compound action potential (ECAP). ECAP measures are less time consuming than EABR measures, because they require much fewer averages and the stimuli can be presented at a faster rate. Also, a recent study by Hughes and Stille (2008) demonstrates comparable measures with ECAP to PTCs. A major downside of the ECAP is that the response can be influenced not only by the way the stimulating electrode interfaces with local neurons but also by the interface near the intracochlear recording electrode. The EABR, by virtue of its far-field scalp recording, reflects more of a whole-nerve response and may more closely reflect the interface of the stimulating electrode. In this respect, the EABR was a better physiological measure for preliminary evaluation of the electrode-neuron interface. Future studies will compare the EABR to ECAP responses obtained from different recording electrodes.

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Spectral Ripple Thresholds Can Be Improved With Training In CI Users: But What Are We Training?

INTRODUCTION

A. Spectral Resolution

Cochlear implants (CI) are a successful intervention for individuals with varying degrees of deafness who no longer benefit from hearing aids. While most CI listeners achieve high levels of speech understanding in quiet, their ability to perceive speech in noise and enjoy music is limited (Firszt et al., 2004; Turner et al., 2004; Stickney et al., 2004; Nelson et al., 2003; Zeng, 2004; Zeng et al., 2005; Fu et al., 1998; Drennan and Rubinstein, 2008). One frequently hypothesized contributor to impaired speech and music perception is poor spectral resolution (Bierer and Faulkner, 2010; Nelson and Jin, 2004; Fu and Nogaki, 2004; Nelson et al., 2003; Qin and Oxenham, 2003; Fu et al., 1998). Spectral resolution, or frequency selectivity, is defined as the ability to resolve the individual frequency components within a complex sound and is particularly important for speech perception in background noise (e.g., Leek and Summers, 1996; Badri et al., 2011). Impaired spectral resolution has been documented in CI listeners (Qin and Oxenham, 2003; Donaldson and Nelson, 2000; Fu et al., 1998; Nelson et al., 1995; Henry et al., 2000). There are many potential sources of this impairment. For example, there are factors related to the device, including the finite number of stimulating channels, mismatched frequency-place alignment, and spectral smearing as a result of channel interaction. Other variables relate to the patient wearing the device and include individual patterns of spiral ganglion nerve survival, etiology and duration of deafness, and general cognitive factors (e.g., attention, memory, motivation). All of these factors alter the way the acoustic information is electrically transmitted, and how the information is processed into neural events that contribute to perception.

While there are many ways perception can be influenced by device- and patient-related factors, one consistent research finding is that CI listeners are not always able to make use of the spectral detail provided by their device. The extent to which each CI user can make use of the frequency content available to them varies substantially. While improving the encoding of the signal at the level of the device is one approach (e.g., study I & II), another is to assist individuals in making better use of the information available to them. One example is *auditory training.*

B. Auditory Training

The use of auditory training exercises with CI listeners is not a new concept. Long ago, Watson (1991) emphasized the importance of teaching CI listeners to make use of the signals provided by their devices. A number of different approaches to auditory training in CI listeners have since been reported; including, training to improve channel discrimination (Fu and Galvin, 2008), music perception (e.g., Fu and Galvin, 2008; Galvin et al., 2007; Gfeller et al., 2002) speech perception in quiet (e.g., Busby et al., 1991; Dawson and Clark, 1997; Fu et al., 2005; Wu et al., 2007; Stacey et al., 2010), and speech in noise (Fu and Galvin, 2008; Oba et al., 2011) [see Fu et al., 2007, for a review]. Several training studies include normal hearing subjects listening to CI simulations of speech (e.g., Li and Fu, 1997; Fu et al., 2005a; Svirsky et al., 2004; Stacey and Summerfield, 2007; 2008; Loebach and Pisoni, 2008; Davis et al., 2005) and the identification of environmental sounds (Shafiro, 2008; Loebach and Pisoni, 2008). In most cases, positive perceptual changes have resulted, demonstrating that it is possible to improve perception in CI listeners using auditory training. As a result, auditory training programs marketed for CI listeners are now available to patients. Most of these training programs focus primarily on speech signals that are “designed to facilitate the remapping of sensations provided by the cochlear implant onto participant's existing linguistic knowledge" (Stacey et al., 2010). However, because of the large numbers of stimuli in these programs, and the highly complex and linguistically loaded nature of the stimuli, it is difficult to determine which specific aspects of training were effective. Was it a change in the sensory encoding of a signal, a shift in perceptual weighting or listening strategy, or merely a shift in bias?

Current gaps in knowledge related to auditory training have been raised by Boothroyd (2010) and Tremblay (2006, 2009, 2011). They point out that we lack information on which aspects of training are responsible for benefit, and which aspects of perception are changed. This gap in knowledge creates a critical barrier to the rehabilitation of people with hearing loss because individuals do not always respond as intended to the training program in which they participate. Even among normal hearing listeners, the effects of training are highly heterogeneous (e.g., Tremblay, Kraus, McGee, 1998; Tremblay, Shahin et al., 2009; Wright et al., 1997; Amitay et al., 2005). Without knowing which aspects of training are responsible for observed benefits that come with training, it is difficult to determine what aspects of the training paradigm were ineffective and what needs of the individual should be targeted.

We have yet to define what auditory training is doing so that the effective components of auditory training programs can be known. For all of the above-mentioned reasons, we argue that it is important to offer a training paradigm that enables the investigator to define the variables being tested so that effective components of the listening experience can be identified. In this study, we purposefully target spectral resolution as a potential mechanism of action to determine if improved spectral thresholds contribute to improved speech perception. To do this, we propose to use spectral ripple stimuli: a stimulus that contains elements of real-world speech-like signals but is void of linguistic cues so that pre-existing, heterogeneous experiences with language can be avoided.

C. Why Spectral Ripple Stimuli?

There are several ways to approach auditory training in CI listeners, with a range of stimuli that can be represented across a continuum. As illustrated in Figure 1, at one end of the continuum, we have psychoacoustic based measures at the single-channel level, such as frequency or channel discrimination. An advantage of these methods is that we have greater control over what we are testing and training. However, these measures are very time-intensive and relatively removed from real-world listening situations, because people using CIs do not listen through one channel. On the other side, we have speech-language based measures. The advantage of these methods is that the testing is more real-life, with words and sentences using the patient's own device and CI settings. However, because of the large variety of materials it is very difficult to know what aspect of these training materials influenced performance, if a change is seen at all. Further, the use of speech and language materials limits the use of these materials for those without a good understanding of language, such as children and non-native English speakers.

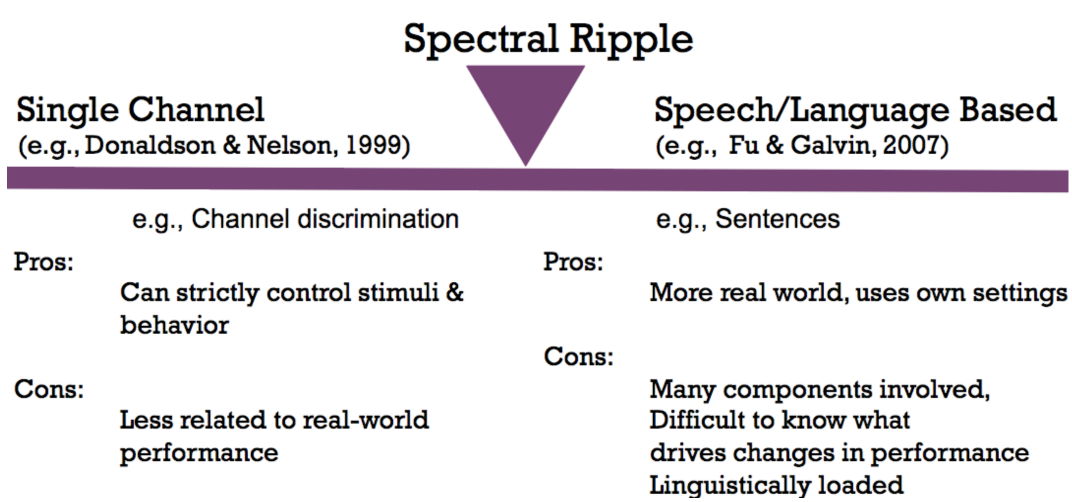

Figure 1. Training stimuli continuum.

Spectral ripple stimuli can be thought of as a middle ground, as the stimuli: 1) are nonlinguistic, and 2) contain a broad frequency range similar to speech that is processed through the patient's own clinical device and settings. The combination of these features make spectral ripple stimuli more relevant to speech perception than single-channel stimulation while at the same time avoiding the linguistic nature of word and sentence materials that would preclude testing young children.

Rippled noise, or comb-filtered noise, is a broadband noise that is sinusoidally shaped in the power spectrum. While ripple noise has a long history (e.g., Glasberg et al., 1984; Houtgast, 1974; 1977; Pick, 1980), Supin et al. (1994) developed the ripple-phase reversal test using these stimuli in a novel way that provided a single-point-measure of the frequency resolving power for a given listener. In this task, a listener indicates whether they can differentiate between two versions of rippled noise, as illustrated in Figure 2.

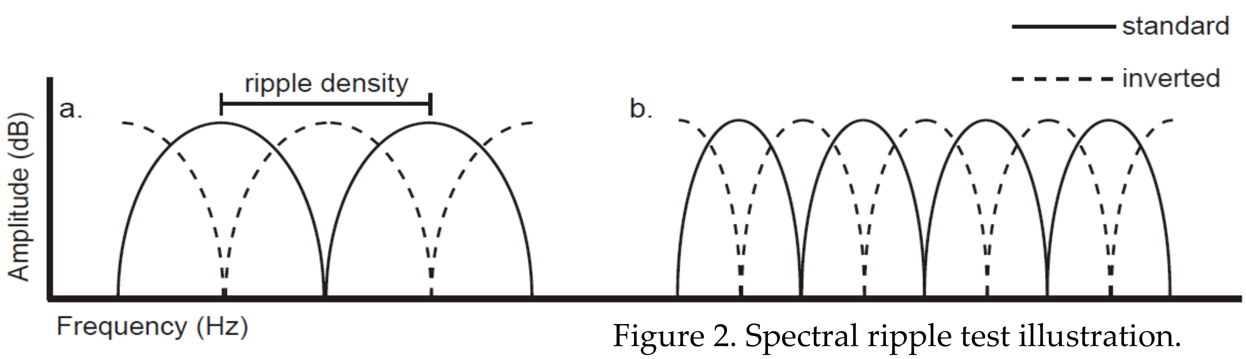

Figure 2. Spectral ripple test illustration.

In this task, a listener is presented with a *standard* ripple noise stimulus at given ripple density, and subsequently presented an *inverted* version where the ripple peaks and troughs are reversed (see Figure 2 – solid lines "standard," dashed lines "inverted"). The densities are increased until the listener can no longer detect that an inversion has occurred, providing a ripple resolution threshold (i.e., the highest resolvable density). Figure 2.a. illustrates widely spaced ripple densities (easier) and 2.b. shows narrowly spaced ripples (more difficult). This test has been used to assess the "frequency resolving power" in normal hearing, hearing impaired and cochlear implant listeners (Supin et al., 1994; 1997; 1998; 1999; Henry and Turner, 2004; 2005; Won et al., 2007; 2010; 2011; Drennan et al., 2010; Anderson et al., 2011). Several of these studies have reported correlations between spectral ripple threshold and vowel and consonant recognition in quiet (Henry et al., 2003; 2005), word recognition in quiet (Won et al., 2007; 2011), speech reception thresholds in noise (Won et al., 2007; 2011), and music perception (pitch-direction discrimination, melody, and timbre identification) (Won et al., 2010). Additionally, while Anderson et al (2011) reported a relationship between ripple thresholds and sentence recognition in quiet, there was no relationship with sentences in noise. What is known, however, is that repeated-testing over time (no auditory training) using spectral ripple stimuli does not result in significant changes in perception, and that individual performance has been shown to be very stable on this task for a year or more (Won et al., 2007; 2010; Anderson et al., 2011). This means that 1) use of spectral information in one's every day environment is insufficient for enhancing use of spectral information, and 2) these particular tools using spectral ripple are stable in individuals. We therefore propose training as a means of intervention to determine if it is possible to improve spectral ripple discrimination thresholds, and whether improvements in spectral ripple threshold transfer to tasks of speech and music perception. The results of this study can help us better understand the relationship between spectral ripple threshold and speech and music perception.

Our approach to spectral ripple training involves a single-interval, yes/no task for three reasons. 1) Ultimately we are interested in a training approach that could be used for CI listeners of any age, and the single-interval, yes/no task can be used with young children who are less capable of performing the more commonly used two-alternative, forced choice procedure. As an example, normal hearing children as young as 6 months and children with CIs at 12 months can perform this task (Horn and

Werner, in prep). 2) The single-interval stimulus design is conducive for recording electrophysiological measures of discrimination (Won et al., 2011). 3) Finally, while the yes/no design is highly dependent on bias, we can monitor bias during testing and training to determine if sensitivity, or bias, changes with training.

D. Case for Single Subject Design

To date, much of what has been studied with respect to training in CI listeners has been reported in terms of group effects. But, as previously mentioned, a common finding in auditory training research is the heterogeneity in performance across listeners. Not only is this true for normal hearing participants, this is particularly relevant to people who wear CIs and have participated in training studies (Fu and Galvin, 2007; Stacy and Summerfield, 2010). One solution to this problem is the single-subject experimental design (SSED). Single-subject experimental designs are highly respected approaches commonly used in behavioral research, especially treatment and intervention studies in education and medicine. Even though they are especially effective when studying heterogeneous populations, single subject designs are underutilized in the field of CI research.

Phase designs are ideal to evaluate whether a functional relationship exists between a behavior and an intervention (Todman, File, and Dugard, 2012; Kennedy, 2005). Sometimes referred to as "time-series" designs, phase designs measure a single behavior of interest repeatedly over time; in this case the behavior is spectral ripple threshold. In a single-subject design, subjects are tested on the measure of interest repeatedly until a stable baseline of performance is reached. Treatment is initiated and the measure of interest is repeatedly tested throughout treatment; in this case, the intervention is focused auditory training. Figure 3 illustrates the multiple baseline single subject research design employed here, using hypothetical data.

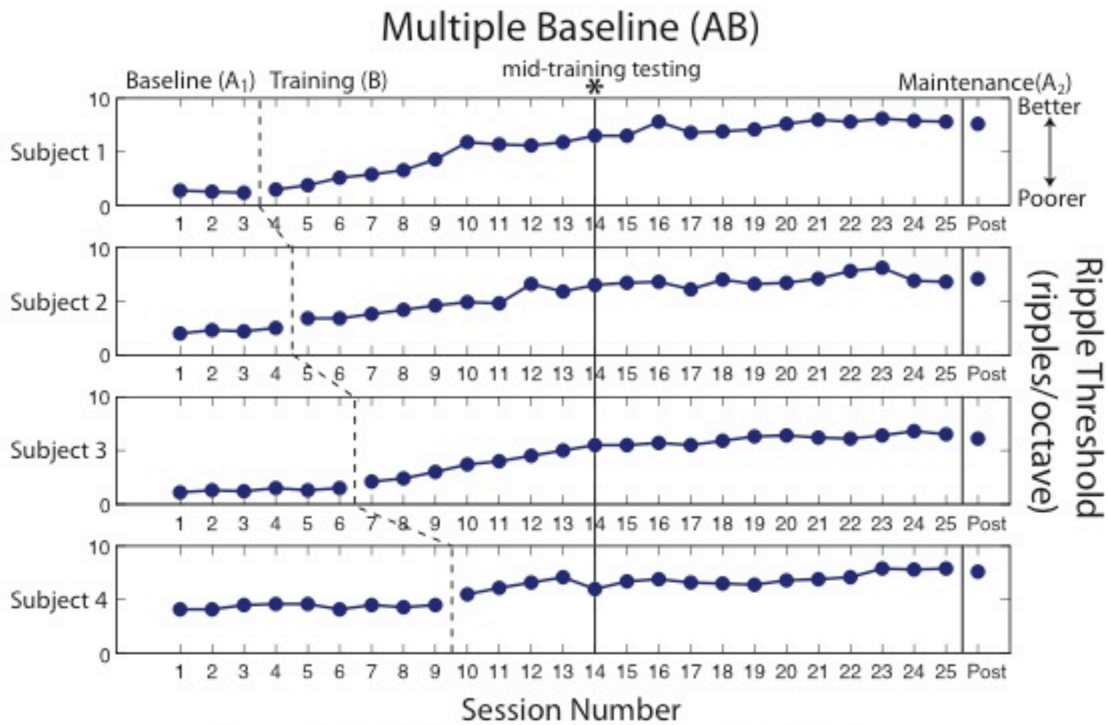

Figure 3. Multiple baseline single subject design.

In Figure 3, each plot represents one subject, the x-axis represents the number of sessions and the y-axis is the individual subject's spectral ripple discrimination threshold (ripples/octave). The points to the left of the dashed lines represent repeated measures of threshold, referred to as baseline. Multiple baseline measures are used to provide a better estimate of typical performance and to assess performance variability. Repeated baseline measures may also reveal improvements that result from repeated stimulus exposure even without training. The dashed line indicates the onset of treatment, in this case, auditory training. The onset of the training protocol is randomly staggered for each subject; a key element to the multiple baseline approach to control for coincident events that might take place on a particular session had all subjects initiated training on the same day. For example, if all subjects showed improvement in performance on session five, regardless of whether it was a baseline or a treatment session, this would indicate that the changes might have occurred without intervention.

However, had these subjects all initiated training on day five, the results may mislead the investigator to conclude that the treatment was effective at changing behavior. The final vertical line indicates the cessation of training, and any points to the right are referred to as retention. In this example, only one immediately post-training session is recorded, but this can be extended for any period of time to evaluate whether the training-related gains are maintained long-term.

Multiple baseline measures are also important because there is evidence that people can learn with repeated testing in the absence of feedback (Moore and Amitay, 2007) and it is important to separate the effects of repeated stimulus exposure, and performance variability, from the effects of training (Tremblay et al., 2010). Each subject serves as his/her own control in this design; the logic is that multiple measurements on the same subject provide a baseline against which the effects of treatment can be evaluated. Phase designs also qualify as randomization designs if the intervention points are chosen at random for each subject, allowing the data to be analyzed statistically using randomization tests (e.g., Dugard, File, and Todman, 2012). In addition to randomization tests, phase designs can be examined using visual analysis to determine if the treatment worked (efficacy), how well did it work (magnitude), and how many sessions did it take (dosage) (Kennedy, 2005; White et al., 2007). Additionally, repeatedly measuring behavior over time, throughout training, provides for closer inspection of patterns of learning that may not be apparent in traditional pre/post-training group designs. As an example, examining individual patterns of performance can help to further define so- called 'mechanisms of action' involved in the training regime. If spectral ripple stimuli are indeed targeting processes involved in spectral resolution, as intended, then one would expect to observe improved performance on outcome measures which rely on spectral resolution. What's more, we would not expect to observe improvement on spectrally-loaded outcome measures for an individual who does not improve their spectral ripple threshold. Exploring individual patterns of learning and retention, therefore, also help to identify sources of variability and identify who may or may not be receptive to training.

E. Outcome Measures

If a training program is intended to address the communication needs of CI users in a clinically relevant way, then it is important that the perceptual gains observed in the research laboratory generalize to behaviors experienced in one's real world listening environment. To do this, ecologically relevant outcome measures are typically included to test the generalization of learned behaviors. Here we propose a training regime aimed at improving spectral resolution. We include relevant outcome measures that are not only ecologically relevant to the communication needs of the CI listener, they also help to define the active mechanism of training. For example, if training participants using spectral ripple stimuli results in improved performance on measures that heavily rely on spectral resolution (e.g., speech perception in noise, music perception) as well as those that do not (intensity discrimination), one could conclude that auditory training is engaging processes that are not specific to spectral resolution.

1. Why Speech in Noise?

Poor spectral resolution is thought to be an important contributor to poor speech perception in the presence of background noise in people with hearing loss (Leek and Summers, 1996). Impaired spectral resolution has been documented in CI listeners (Bierer and Faulkner, 2010; Bierer, 2010; Qin and Oxenham, 2003; Nelson and Jin, 2004; Fu and Nogaki, 2004; Donaldson and Nelson, 2000; Nelson et al., 2003; Fu et al., 1998; Nelson et al., 1995; Henry et al., 2000; Fu et al., 1998). As a result, CI listeners show poor performance on tests of speech perception in the presence of background noise (Firszt et al., 2004; Turner et al., 2004; Stickney et al., 2004, Nelson et al., 2003; Zeng 2004; Zeng et al., 2005; Fu et al., 1998). As previously described, those individuals with better spectral ripple thresholds perform better on tasks of speech in noise. Therefore, we include speech in noise tests as an outcome measure and hypothesize that perceptual gains in spectral ripple threshold will translate to improved speech in noise performance.

2. *Why Music Perception?*

CI listeners perform more poorly than normal hearing listeners on many aspects of music perception, which is thought to be a result of the inadequate representation of the spectral details by the device and the listening strategy employed by CI listeners (e.g., Gfeller et al., 1997; 2000; 2002). The typical CI processor is engineered for speech signals and includes only spectral envelope information. This is a problem for music perception because the temporal fine-structure of the incoming acoustic sound, which is key to music perception, is discarded. It is not surprising then that much of the ongoing research in music perception with CI listeners is aimed at engineering approaches to improve signal processing strategies in order to retain much of the spectral detail important for music (e.g., Drennan et al., 2008). Once again, auditory training cannot change the quality of the signal that is entering the auditory system by the CI device, but it can be used to facilitate better use of the information being delivered by the processor. Significant correlations have been reported between spectral ripple thresholds and performance on tests of music perception, specifically, that subjects with better spectral ripple thresholds performed better on tasks of pitch direction discrimination and also melody and timbre identification (Won et al., 2010). Therefore, we included these three subtests of music perception as outcome measures to determine if improved spectral ripple thresholds results in improved melody, timbre, and pitch direction perception.

3. *Intensity Discrimination*

Spectral ripple discrimination can be thought of as spectral shape discrimination, and performance depends on the ability of the CI listener to integrate the spectral information across channels – therefore the threshold should be influenced by how well the spectral information is encoded by the device and how well listeners can use this information. However, there is debate about the perceptual mechanism responsible for performance on this task (Azadpour et al., 2011; Won et al., 2011). Because the sound spectrum is a function of frequency and intensity, improved spectral resolution could result from improvements in the representation of frequency or intensity, or both. Additionally, the spectral ripple task can be influenced by within-channel and across-channel changes in loudness, particularly at low ripple densities. To ensure intensity cues are taken into consideration, pre- and post- measures of intensity resolution can help to determine if improved spectral ripple thresholds following training might be explained by the improved use of intensity, rather than spectral, information.

4. *Hearing Handicap*

Untreated hearing loss is known to contribute to a poor quality of life; with people avoiding social situations and having feelings of loneliness (e.g., Chia et al., 2007). Research aimed at improving speech understanding in noise, therefore, has the potential to improve the CI user's quality of life. While CIs mostly have a positive impact on the lives of the recipients, many listeners continue to report hearing, communication, and psychological difficulties after implantation with greater speech perception outcomes related with higher reports of quality of life (e.g., Hirschfelder, Grabel and Olze, 2008). Therefore, we included a measure of quality of life as an outcome measure to determine if participating in training resulted in improved self-report of hearing handicap.

F. Timeline and Retention

The time course for learning on spectral ripple resolution training is unknown. Therefore several factors influenced the frequency, number of training trials, and duration of the training program. Because CI listeners performed the training exercises in the laboratory, rather than their own homes, training several days per week for each subject was impractical for most subjects. While many CI training studies have subjects performing training several days per week, Nogaki et al. (2007) suggest that the frequency of training does not reduce the training outcomes with the same total number of training sessions.

Published research involving cross-language training in normal hearing listeners suggests novel word learning can be retained for up to six months (Bradlow et al., 1999). Neural mechanisms associated with word learning in normal hearing adults show even longer effects, up to one year (Tremblay et al., 2010). Little is known about the retention of trained sounds in CI users; therefore, we propose to retest all outcome measures at 1 month and 3 months post training.

RESEARCH QUESTIONS

The proposed experiment addresses four research questions: 1) Does focused auditory training using spectral ripple stimuli result in improved spectral ripple discrimination in adult CI listeners? We hypothesized that speech perception is dependent on spectral resolution and, therefore, improved spectral ripple thresholds would improve performance. To test this hypothesis, we use spectral ripple training, as indicated by improved spectral ripple threshold, to determine if changes in spectral ripple thresholds relate to the way the subjects make use of the frequency information provided to them by their own device. 2) Does improved performance on spectral ripple discrimination transfer to untrained tasks of speech perception in quiet and in noise, music perception, and increment detection, when compared with an untrained control group? If speech perception is dependent in part on improved spectral resolution, we hypothesize that improvements in spectral resolution would transfer to untrained tasks that were directly related to spectral resolution (speech and music), but *not* transfer to untrained tasks that were unrelated to spectral resolution (intensity resolution). 3) Do improvements in spectral resolution and perception impact overall communication abilities and quality of life using standardized outcome measures of CI success? We hypothesize that improved spectral ripple thresholds will in turn impact speech perception performance as well as measures of psychosocial benefits. 4) How long are observed changes in performance retained? 1 month? 3 months post-training? We hypothesize that the observed gains described in the earlier aims will be retained at 3 months post-training.

METHODS

A. Participants

Sixteen adult cochlear implant listeners, seven women and nine men, participated in this study. Subjects were recruited through the University of Washington Communication Sciences Participant Pool. The subjects ranged in age from 22 to 83 and were all native speakers of American English who had become deaf post-lingually, with no known neurological conditions. Ten listeners wore a Cochlear device, and eight wore Advanced Bionics; three were implanted bilaterally. All subjects had at least 6 months experience with their implants. Details about individual subjects can be found in Table 1, including age, etiology, duration of severe hearing loss prior to implantation, implant type, and duration of implant use. No subject had residual hearing thresholds greater than 60 dB HL in their non-implanted ear. Subjects were randomly assigned to the training and control groups. The individuals in the training group made between fourteen and sixteen visits to the laboratory (1 to 3 hour-long visits), and participated in approximately 34 hours of testing/training over the course of six months. All eight subjects in the training group completed pre- and post-training sessions including follow up at 1 month; five returned for the final 3 month post-training session. The individuals in the control group made five visits to the laboratory (2 hour long visits), and participated in approximately ten hours of testing over the course of six months. Each subject provided written consent, and the experiments were conducted in accordance with guidelines set by the Human Subjects Division of the University of Washington.

Table 1. Subject Demographics.

Subj. #	Gender	Age (yr)	Etiology	Duration of Severe Hearing Loss	Implant Type	Duration of Implant Use
TRAINING GROUP N = 8						
S1	F	67	Hereditary	6 yrs	Nucleus 24	2 yrs
S2	M	25	Unknown	6 yrs	(R)Nucleus 22 (L)Nucleus 24	(R) 12 yrs (L) 9 mo
S3	M	77	Unknown	3-4 yrs	Nucleus 22	3 yrs
S4	F	60	Unknown	10yrs	Nucleus 22	10 yrs
S5	F	32	Atypical Meniere's	1 yr	(R)HiRes90k (L)HiRes90k	(R) 1 yr (L) 4 yrs
S6	F	22	Enlarged Aqueduct Syndrome	4 yrs	Nucleus 22 Nucleus 24	(R) 8 yrs (L) 2 yrs
S7	F	39	Hereditary	25 yrs	Clarion CI	10 yrs
S8	M	87	Viral	30 yrs	Nucleus 22	5 yrs
CONTROL GROUP N = 8						
S9	F	69	Hereditary	6 yrs	HiRes90k	2 yrs
S10	M	46	Unknown	16 yrs	HiRes90k	1 yr
S11	M	81	Unknown	2 yrs	Nucleus 22	5 yrs
S12	F	47	Unknown	35 yrs	HiRes90k	2 yrs
S13	M	59	Unknown	5 yrs	Nucleus 24	3 yrs
S14	M	83	Cranial Injury	9 mo	Nucleus 22	5 yrs
S15	M	75	Unknown	1.5 yrs	Nucleus 24	3 yrs
S16	M	81	Unknown	10 yrs	HiRes90k	3 yrs

B. Procedure

In all conditions, subjects were tested in a sound-attenuated booth, one meter in front of a single loudspeaker (JBL Professional LSR25P Bi-Amplified) at 0 degrees azimuth. Subjects used their own device and clinical map for the duration of the experiment. During the first session, subjects were advised to set the volume to a comfortable level. Their selected levels were documented so that subjects could be asked to re-adjust to the same level during all subsequent visits. Bilateral CI users were tested using both implants. To ensure that observed changes in performance over time were not related to changes in CI processor settings, clinical maps were not modified during the participation of this study.

C. Spectral Ripple Discrimination Training

1. Overall Design

The design of this training experiment is the same as shown in Figure 3, consisting of three phases. Each phase will be described in more detail below. A randomized, multiple-participant, staggered-baseline, ABA single-subject research design was used with three successive phases, the baseline phase (A_1), the treatment phase (B), and the maintenance phase (A_2). The proportion of sessions dedicated to training versus baseline testing was intentionally manipulated in order to establish a temporal relationship between a change in behavior and the onset of the treatment; therefore each subject's treatment onset was randomized. Subjects were randomly assigned to complete at least four, but not more than 10 baseline sessions prior to the onset of training. All subjects in the training group participated in 23 total sessions, so that the amount of stimulus exposure was equal across subjects. Because the number of trials per testing and training condition was controlled, subjects were not able to replay a stimulus. Thresholds were measured for each session during three phases of testing and training – baseline (4-10 sessions), training (10-16 sessions), and maintenance (3 sessions). The stimuli and methods used for testing/training are described below.

a. Baseline Testing (A_1)

To determine baseline spectral ripple discrimination thresholds, subjects were randomly assigned to complete a minimum of four and maximum of ten spectral ripple baseline sessions. Thresholds were obtained with the methods previously described without feedback. Each threshold was an average of four runs, for a total of 400 trials per threshold (one session).

b. Training (B)

The training method is identical to the baseline testing, except for the inclusion of trial-by-trial feedback. The session of treatment onset was randomly assigned (sessions 5-11). Subjects returned to the laboratory for training until all sessions were completed (approximately eight weeks). Thresholds were continuously measured during each training session to assess ongoing learning effects.

c. Maintenance (A_2)

Because the training was intended to create a lasting change in performance, the final A phase is referred to as maintenance. The results of this phase were used to determine the degree of immediate learning effects as well as retention, included spectral ripple discrimination tests immediately post-training, one month post-training, and three months post-training. Thresholds were obtained with the methods previously described without feedback.

2. *Stimuli*

Stimuli are those used by Won et al. (2011). Four hundred pure-tone frequency components were summed to generate a rippled noise. Amplitudes of the components were determined by a full-wave rectified sinusoidal envelope on a logarithmic amplitude scale. Spectral peaks were evenly spaced on a logarithmic frequency scale. The stimuli had a bandwidth of 100-5,000 Hz with a peak-to-valley ratio of 30 dB. The starting phases of the components were randomized for each presentation. The stimuli are generated with 100 different "ripple" densities, spaced logarithmically from 1 to 25 ripples per octave. Both a standard and an inverted version of the standard stimuli with the spectral envelope phase reversed were created. For the standard ripple stimuli, the phase of the full-wave rectified sinusoidal spectral envelope was set to zero radians, and for the inverted stimuli it was set to $\pi/2$.

Two 4-s long stimuli were created, one without a spectral change (standard-standard, "*no-change*"), and one with a spectral change at the midpoint (standard-inverted, "*change*"). In these stimuli, the first 2-s consists of the standard stimulus while the final 2-s contains either a standard or an inverted stimulus. For the standard-standard, or "*no-change*" stimulus, a 4-s duration standard ripple stimulus was created and ramped with 150-ms rise/fall times, then filtered with a long-term, speech-shaped filter (Byrne et al., 1994) and with a 5 kHz low-pass filter. Therefore, there was no spectral change within the stimulus.

For the standard-inverted, or "*change*" stimuli, each 2-s of each stimulus was individually created and concatenated. To eliminate the possibility of a loudness change in the standard-inverted stimuli at the 2-s point, the root-mean-square of the first 2-s (standard ripple) and the last 2-s (inverted ripple) were matched. After amplitude matching, the stimuli were ramped with 150 ms rise/fall times, and filtered using a long-term speech shaped filter (Byrne et al., 1994). Finally, a 5kHz filter was applied to each stimulus to eliminate any high frequency components added during concatenation. To prevent a subject from perceiving a cue due to the temporal pattern of transition from standard to inverted ripple at the point of change, the phases of the 400 individual frequency components was randomized for the first 2-s and second 2-s separately. To control for potential variations within the individual stimuli, ten versions of each ripple density were created resulting in a total of 2,000 stimuli (1,000 each standard-standard and standard-inverted).

Figure 4.a shows spectrograms for both the waveforms (upper panels) for the standard-inverted and standard-standard stimuli in the one ripple per octave condition. Figure 4.b shows the amplitude plotted for the duration, and the fine structure at the

transition point (1.98-2.02 seconds) for the one ripple per octave standard-inverted stimulus, showing that there is no temporal cue (such as sudden overall amplitude change) that a subject could use to detect a change within the standard-inverted stimuli.

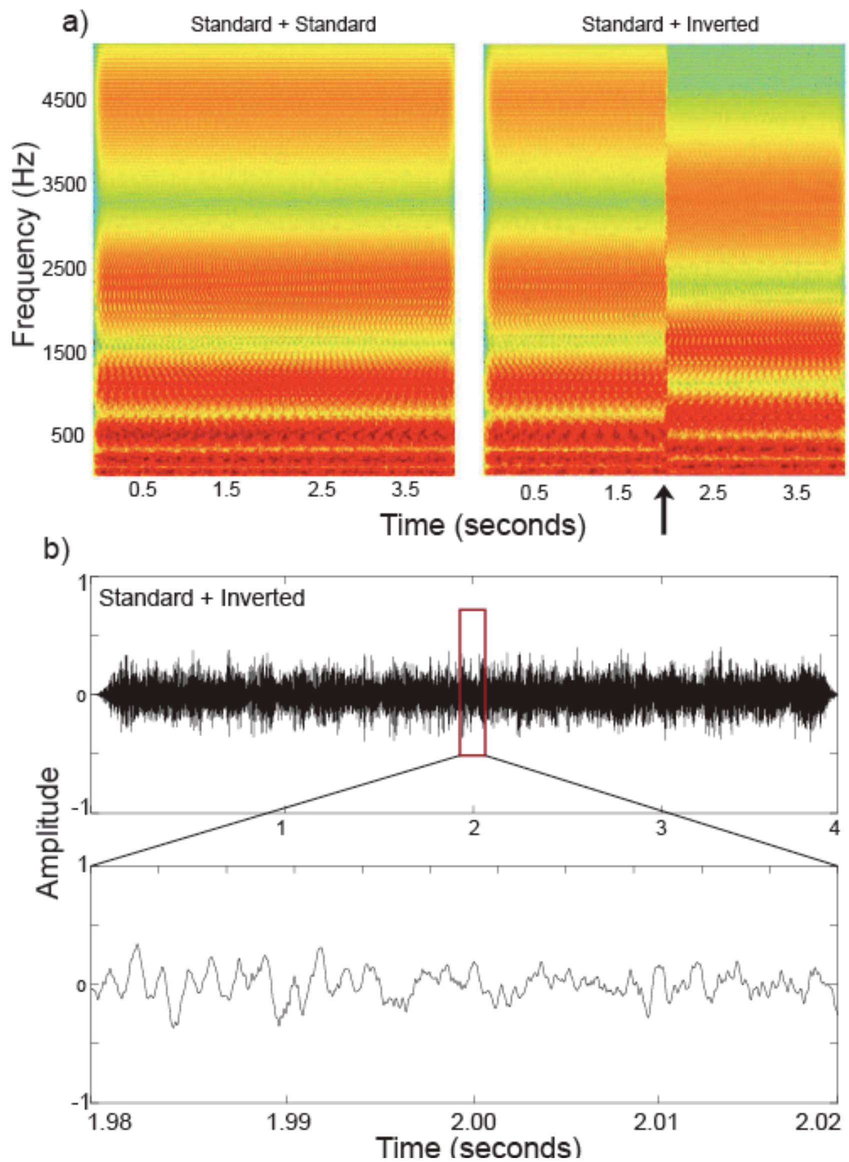

Figure 4. a) Spectrogram; b) Amplitude waveform and fine stricture at transition.

3. *Procedure*

Spectral ripple density thresholds were estimated adaptively with a two-up, one-down, single-interval, two-alternative forced choice (yes/no) procedure (Levitt, 1971). No-change trials were included randomly and with equal probability, with the constraint that not more than three no-change trials were presented consecutively. Only responses to change trials altered the direction of the adaptive track, and step size varied during a run following PEST rules (Taylor and Creelman, 1967). The ripple densities tested in this experiment were 100 log steps from 1 to 25 ripples/octave, and the step sizes were 16, 8, 4, 2, and 1 (based on PEST rules). The initial trial in all runs was a change trial at 1-ripple per octave, a level with the greatest contrast from the standard to inverted, and based on preliminary data, estimated to be suprathreshold for all subjects. Each run included 100 trials, and threshold was calculated as the average of the reversals (i.e., changes in the direction of the ripple density) once the step size reached 2 steps. Criterion location (c) was calculated to evaluate response bias based on the hit and false alarm rates for the change and no-change trials for 5 steps above and below the ripple threshold ($c = -\frac{1}{2}[(z(H) + z(F))]$, Macmillan and Creelman, 2005). Feedback was only provided during training, and subjects could not replay a stimulus. During stimulus presentation, a visual cue was provided at 2 seconds indicating the point in time the change may occur. Practice, testing and training were conducted using a custom Matlab program (The Mathworks, Inc.) with the subject indicating whether they heard a change in the stimulus by clicking on a button labeled "*yes*" or "*no*" using a computer mouse. Subjects then clicked a button marked "next" to initiate the next stimulus trial, and were able to take a break at any time. The stimulus presentation level was 65 dBA. One test session consisted of four runs of 100 trials (400 trials/session) and took approximately 45 minutes to complete. Subjects were given short (2-5 min) breaks after two sequential threshold runs, and a longer (5-10 min) break between sessions; no more than two testing or training sessions were completed during one visit. More breaks were provided at the participant’s request.

To orient subjects in the training group to the test stimuli and procedure, one practice session was completed prior to testing. Subjects used a computer mouse to choose one of 6 buttons on a computer screen corresponding to 6 stimuli ranging from easy to discriminate (1 ripple/octave) to increasingly more difficult (>8 ripples/octave). All stimuli were change trials, and the subject was advised that each contained a “change” but they may or may not be able to detect the change in all stimuli. The purpose of this practice session was to provide the subject with experience of listening to the sound quality changes across stimuli and orient them to the point in the time when they should listen for the change. To move to the baseline testing phase each subject listened to each stimulus at least twice, and no subject listened to more than four examples during this orientation procedure.

D. Outcome Measures

Table 2 shows the approximate timeline and tests administered at each visit for the training and control groups. A description of each outcome measure is below.

Table 2. Timeline.

Training Group	**Pre-Training** Ripple test CNC in Quiet SRT in Noise CAMP Increment HHIA/E	~1 month (5-10 sessions) training →	**Mid-Training** Ripple test CNC in Quiet SRT in Noise CAMP	~1 month (5-10 sessions) training →	**Post-Training** Ripple test CNC in Quiet SRT in Noise CAMP Increment HHIA/E	~1 month no training →	**1-mo Post-Train** Ripple test CNC in Quiet SRT in Noise CAMP	~2 months no training →	**3-mo Post-Train** Ripple test CNC in Quiet SRT in Noise CAMP
Control Group	**Visit A** CNC in Quiet SRT in Noise CAMP HHIA/E	~1 month no training →	**Visit B** CNC in Quiet SRT in Noise CAMP	~1 month no training →	**Visit C** CNC in Quiet SRT in Noise CAMP HHIA/E	~1 month no training →	**Visit D** CNC in Quiet SRT in Noise CAMP	~2 months no training →	**Visit E** CNC in Quiet SRT in Noise CAMP

The subjects in the training group were measured at five points during the training program, pre-training (first visit or during baseline phase), mid-training (~session 15), immediately post-training, one-month post-training, and three-months post-training. At each session, the subjects in the training group completed the following tests: spectral ripple discrimination (no feedback), two CNC lists in quiet, Spondee SRT in noise, and CAMP (pitch, melody, timbre). The Hearing Handicap Inventory was administered during the pre-training session and immediately post-training. Increment detection was assessed (in a subset of subjects) prior to the onset of training and post-training.

The subjects in the control group were measured at five points during a six-month period. These visits approximated the time course for the training group, once per month for four months (visits A, B, C, and D) and at six months (session E). Control subjects were not assessed on spectral ripple discrimination or increment testing at any point during the experiment. At each session (A-E), subjects completed the following tests: two CNC lists in quiet, Spondee SRT in noise, and CAMP (pitch, melody, timbre). The Hearing Handicap Inventory was administered at visit A and C.

1. *Word Recognition in Quiet*

The consonant-nucleus-consonant (CNC) word recognition test (Peterson and Lehiste, 1962) is an open-set monosyllabic speech perception test recorded by a single male speaker and was used as the measure of speech in quiet. CNC word lists were randomly chosen out of ten lists for each subject, and no individual subject was given the same list twice. Two CNC lists (100 words) were presented to subjects during each of the five outcome test sessions (Figure 5). Word lists were presented in sound field at 65 dBA and subjects repeated the word they heard aloud. A total percent correct word score was recorded for each subject.

2. *Speech Reception Threshold in Noise*

The speech reception threshold (SRT) test is an adaptive, closed-set task where listeners identify one of 12 spondees in the presence of steady state, speech-shaped noise (Won et al., 2007; 2011; Turner et al., 2005). The SRT test was applied with the same stimuli and methodology as previously established by Won et al. (2007). The SRT test measures the level of noise required to just mask a closed set of 12 equally difficult spondees (Harris, 1991). The spondees, two-syllable words with equal emphasis on each syllable, were recorded materials presented by a female talker. The same background noise was used on every trial, and began 500ms prior to the onset of the spondee. The level of the speech-shaped noise was calibrated prior to initiating the first test run. A practice run was completed prior to the first experimental run to orient the subjects to the task and familiarize to the words. The subject was presented with a matrix of 12 spondees on a screen, by clicking on each spondee the subject heard the word presented in noise at +10 dB signal-to-noise ratio (SNR; spondee 65 dBA, noise 55 dBA). During the practice run, each subject had to listen to each word twice to move on.

In all test sessions, the level of the spondees was fixed at 65 dBA while the steady-state noise was varied adaptively using a one-down, one-up procedure (Levitt, 1971), in 2-dB steps. Feedback was not provided. Testing continued until 14 reversals were obtained and threshold was calculated as the average of the last 10 reversals. The dependent variable was the mean SNR of four test runs.

3. *Music Perception*

The University of Washington Clinical Assessment of Music Perception (CAMP) was used for music testing (Nimmons et al., 2008; Kang et al., 2009). All stimuli in this task were presented at 65 dBA. The CAMP consists of three subtests: complex-tone pitch direction discrimination, melody recognition, and timbre recognition. The details of stimulus creation and the melodies used are described in Nimmons et al. (2008). Practice sessions preceded every test session, as described below, and written instructions were displayed prior to each subtest, for every test session.

a. Complex-Tone Pitch Direction Discrimination

The pitch direction discrimination subtest was a two-interval, two-alternative forced choice task with a one-up, one-down tracking procedure (Levitt, 1971). In the task, subjects identify which of two tones has the higher fundamental frequency. Synthesized complex tones, 760 msecs in duration, were presented at 65 dBA. To minimize level cues, presentation levels were roved within trials (±4 dB range in 1-dB steps). On each presentation, a tone at the reference fundamental frequency (262, 330, or 392 Hz) and a tone with a higher fundamental frequency were played in random order. Subjects identified which of the two had the higher fundamental frequency. Complex tones of three fundamental frequencies (262, 330, and 392 Hz) were used in the task.

To familiarize the subjects with the task, the first four trials were identified as practice trials, and feedback was provided. If they made an error, they had to change their response to move on to the next trial. Feedback was not provided during testing. The initial interval between the standard and the reference tone was 12 semitones (one octave), a contrast discriminable by all subjects. For the first and second reversals, the signals were adjusted in six and three semitone steps, respectively; for the final six reversals, the interval was adjusted in one-semitone steps. Because 1 semitone was the smallest tested interval, it was considered the best achievable score; therefore a reversal at zero was automatically added by the test algorithm when the subjects answered correctly at 1 semitone. Adaptive tracking was performed simultaneously for the three reference frequencies interleaved in random order until three runs of eight reversals were completed for each reference frequency. The threshold was estimated as the mean difference limen in semitones for three adaptive tracks, each determined from the mean of the final six of eight reversals.

b. Melody Identification

The melody identification subtest requires subjects to identify isochronous melodies from a closed set of 12 popular melodies. The test uses synthesized piano-like tones with identical envelopes. Amplitude of each note in the melody was randomly roved by ± 4 dB and five different versions were pre-synthesized. Rhythm cues were removed such that notes of a longer duration were repeated in an eighth-note pattern. Subjects were asked to identify melodies by selecting the title corresponding to the melodies presented. The dependent variable is a total percent correct score calculated after 36 melody presentations, including three presentations of each melody (Nimmons et al., 2008). To familiarize subjects with the melodies, a practice session preceded testing where subjects had to listen to all test melodies twice. During the testing phase, each melody was played three times in random order across 36 presentations. The score was the percentage of correct responses.

c. Timbre Identification

The timbre recognition subtest requires subjects to identify randomly presented musical instruments based on timbre, from a closed set of eight common instruments. The stimuli were recordings of real instruments playing an identical standardized, connected melodic sequence, and a piano, trumpet, clarinet, saxophone, flute, violin, cello, and guitar. Similar to the melody test, a practice session preceded testing and subjects had to listen to each instrument twice. During testing, each instrument was presented three times in random order across 24 presentations. The score was the percentage of correct responses.

4. *Quality of Life*

Hearing loss related quality-of-life was assessed using the Hearing Handicap Inventory scale (Newman et al., 1990; Ventry and Weinstein, 1982). There are two versions of this questionnaire, one for adults and one for the elderly (>65 years of age), the age of the subject determined which version was used. These are 25-item questionnaires with two subscales: social function and emotional function as they relate to hearing loss with a maximum score of 100. The questionnaires were administered in a paper-pencil format, and instructions were reviewed with each subject prior to completion.

5. *Intensity Discrimination*

Intensity difference limens (DL) were measured in a subset of subjects (five of the eight training subjects: S1, S2, S3, S6 and S8) to determine if training related gains in performance resulted from an improved ability to discriminate intensity rather than frequency. Three experimental electrodes were chosen for each subject and were those that reflected the largest intensity difference, based on the electrode outputs, in response to spectral ripple stimuli at the highest ripple density each subject could discriminate (i.e., subject-specific spectral ripple threshold).

Stimuli were biphasic pulse trains delivered in a direct-stimulation mode via a specialized CI interface (Cochlear Corporation). The stimulation mode was monopolar (MP), with a pulse duration of 25 μs/phase. The intensity DLs were measured on three single electrodes for each subject. The initial phase of each pulse was always cathodic (negative). Intensity DLs were measured with a "pseudo-continuous" paradigm, adapted from the continuous paradigm described in Wojtczak et al. (2003). The pedestal (650 ms) is played continuously throughout the 1700-ms trial, and beginning 650 ms after trial onset, the increment (ΔI) is presented in one of three final 100 ms intervals, separated by 250 ms. The onset of the pedestal and the increment were ramped with a 10 ms rise/fall time to minimize possible onset cues. Using a custom Matlab program, increments were generated by specifying higher amplitude pulses, relative to the pedestal level, randomly in one of the three final intervals. Thresholds for detecting increments were measured for two amplitude levels of the pedestal, corresponding to 50% and 75% of the subject's dynamic range measured in clinical level steps. T and C levels from the subject's clinical map determined the range of testing, and C levels were never exceeded. The three intervals were visually indicated on a computer screen in front of the subject. The increment (ΔI) occurred randomly in one of the three intervals, and the subject had to indicate which of the three intervals contained the increment. The initial increment was set to a level large enough to be easy to detect. The step size was one clinical level step, corresponding to approximately 2.046% or 0.176 dB of current. A run was terminated after twelve reversals, and threshold was calculated as the mean of the final eight reversals. Intensity DLs were obtained by converting the threshold amplitude increments into Weber fractions (WF) in dB ($10 \log(\Delta I/I)$). The final WF in dB for each channel at each base level was obtained by averaging two to four WFs obtained from separate runs.

RESULTS

A. Does focused auditory training using spectral ripple stimuli result in improved spectral ripple discrimination in adult CI listeners?

Group mean spectral ripple thresholds improved relative to baseline for all training phases (Figure 5). A repeated measures ANOVA confirmed that mean spectral ripple thresholds differed between training phases ($F(4, 16) = 5.521$, $P = 0.001$). Planned comparisons, using paired sample t-tests, showed statistically significant change in spectral ripple thresholds from baseline to mid-training, immediately post-training, 1 month post-training, and up to 3 months post-training (Table 3). Figure 5 shows the mean spectral ripple threshold recorded for each phase of testing on the left, and the mean change in ripple threshold on the right (positive values mean better thresholds), error bars indicate 95% confidence intervals, with each individual subject being identified by a unique symbol (see key). A randomization test also confirmed a significant change from pre-training to post training (p = 0.022).

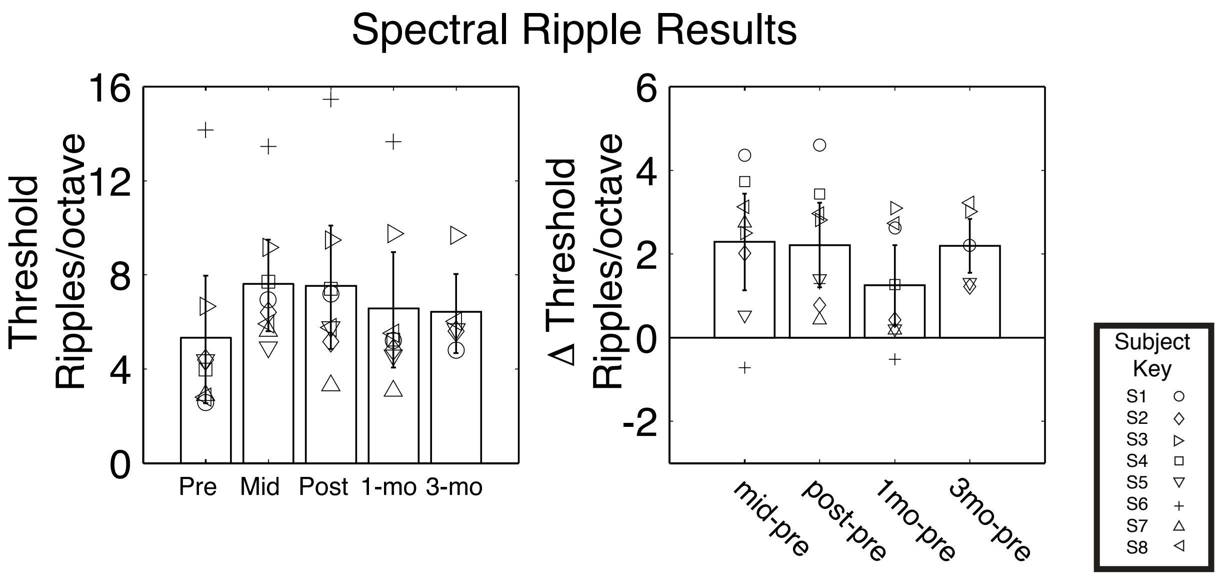

Figure 5. Spectral ripple group results.

Table 3. Spectral ripple group results.

Phase	Improvement	Sig
Mid-training – Baseline	2.29 ±1.67 RPO	p = 0.006
Post-training – Baseline	2.21 ±1.45 RPO	p = 0.004
1mo-post – Baseline	1.25 ±1.38 RPO	p = 0.038
3mo-post – Baseline	2.19 ±0.93 RPO	p = 0.006 (n=5)

Group mean response bias (criterion location – c) values changed relative to baseline for all training phases (Figure 6). A repeated measures ANOVA confirmed that mean c values differed between training phases ($F(4, 16) = 5.788$, $P = 0.004$). Planned comparisons, using paired sample t-tests, showed statistically significant decreases in c values from baseline to mid-training, immediately post-training, 1 month post-training, and up to 3 months post-training (Table 4). Figure 6 shows the mean c values recorded for each phase of testing on the left, and the mean change in c on the right, error bars indicate 95% confidence intervals, with each individual subject being identified by a unique symbol. Positive values of c indicate a conservative response bias, while negative values indicate a liberal response bias. On average, subjects changed from being conservative at baseline to being less conservative or unbiased at subsequent time points.

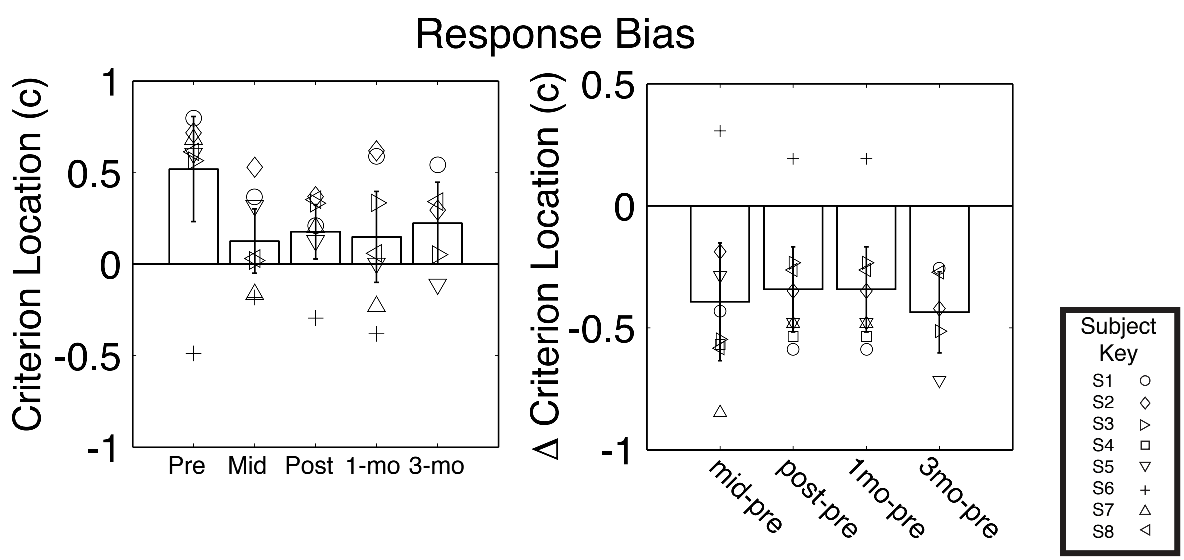

Figure 6. Response bias (c) changes.

Table 4. Response bias (c) changes.

Phase	Change	Sig
Mid-training – Baseline	-0.393 ±0.35	p = 0.015
Post-training – Baseline	-0.342 ±0.25	p = 0.006
1mo-post – Baseline	-0.371 ±0.32	p = 0.014
3mo-post – Baseline	-0.436 ±0.19	p = 0.007 (n=5)

Because changes in bias coincide with improvements in spectral ripple thresholds, the influence of bias makes it difficult to determine whether apparent improvements in spectral ripple thresholds reflect changes in sensitivity, changes in response bias, or both. Therefore, we cannot conclude that improved spectral ripple thresholds reflect an improvement in spectral resolution.

While the group results show that the spectral ripple training was effective in improving spectral ripple thresholds, it is also important to examine individual differences in performance. Historically, multiple approaches have been used to determine which subjects improved and who did not. Here we chose to use 'a method of projection' that is based upon an individual's baseline performance (White et al., 2007). This modeling approach incorporates all sessions during the baseline phase, and projects the performance that would be expected in the absence of intervention (White et al., 2007). If actual performance scores during the training phase surpass the projection, the individual improved as a result of the treatment. As an example, in Figure 7, the individual data for two subjects are shown with projected lines of progress. In this example, S4 improved, and S5 did not.

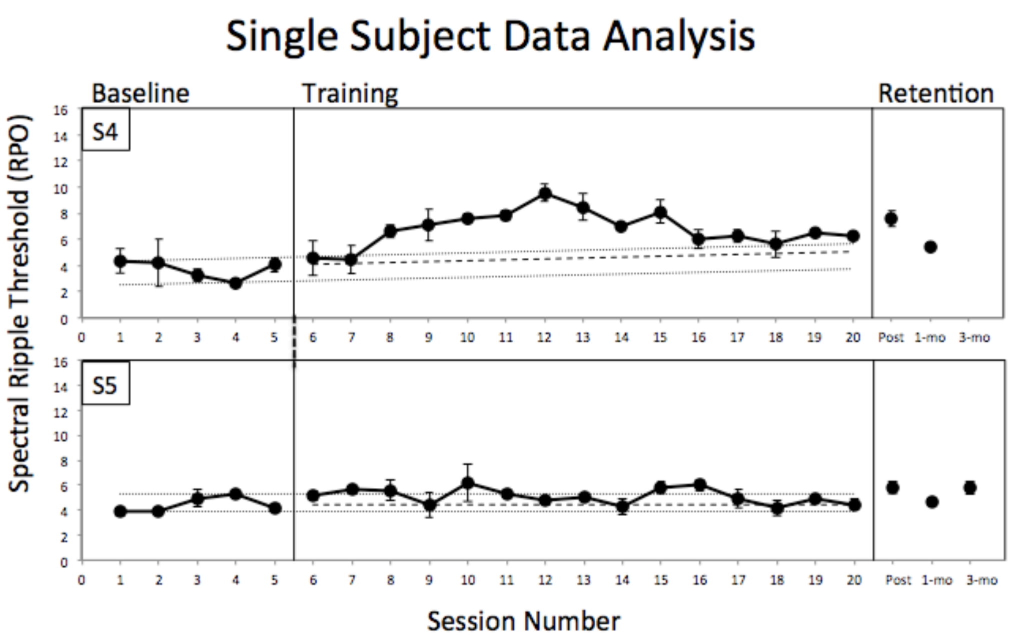

Figure 7. Single-subject data analysis.

Using these objective criteria, of the eight trained subjects, four improved, and four did not. Table 5 shows the means for the baseline, training and the resultant change in ripple threshold for each subject, with those who improved indicated in italics. It appears that at the minimum, a change of 1.3 ripples per octave was required to show improvement, with a range of 1.387-4.07 ripples per octave improvement. Table 5 also shows the response bias (c) means for the baseline, training, and the change in c for each subject. Shifts in bias occurred for all subjects from the baseline to the training phase, becoming more unbiased (0 = unbiased) during the training phase for all but one subject (S6). However, that subject shifted from a liberal response bias to essentially unbiased.

Table 5. Single subject results: ripple and criterion location (c).

	Spectral Ripple			Criterion Location (c)		
Subject	Mean Baseline	Mean Training	Change	Mean c Baseline	Mean c Training	c Change
S1	***2.6456***	***6.0566***	***3.4110***	***0.87***	***0.45***	***-0.42***
S2	4.4528	5.2954	0.8426	0.67	0.35	-0.32
S3	6.7348	7.8518	1.1170	0.47	0.17	-0.30
S4	***4.0417***	***6.8491***	***2.8075***	***0.63***	***0.14***	***-0.49***
S5	4.4447	5.1089	0.6642	0.53	0.05	-0.82
S6	14.4679	12.5525	-1.9155	-0.46	-0.02	+0.57
S7	***3.0261***	***4.4137***	***1.3877***	***0.32***	***-0.16***	***-0.48***
S8	***2.8623***	***6.9388***	***4.0764***	***0.69***	***0.07***	***-0.62***

Figure 8 shows all of the individual data for each subject across each session (spectral ripple threshold in ripples per octave on the ordinate, and session number on the abscissa). Each plot has three phases, Baseline, Treatment, and Retention. The vertical line indicates the onset of training (varies for each subject, session 4-9) with the points to the left representing baseline testing and the points to the right indicating thresholds measured during training. The second vertical line indicates the cessation of training (following session 20), and the points to the right are post-training thresholds: immediately post-training, 1 month post-training, and 3 months post- training.

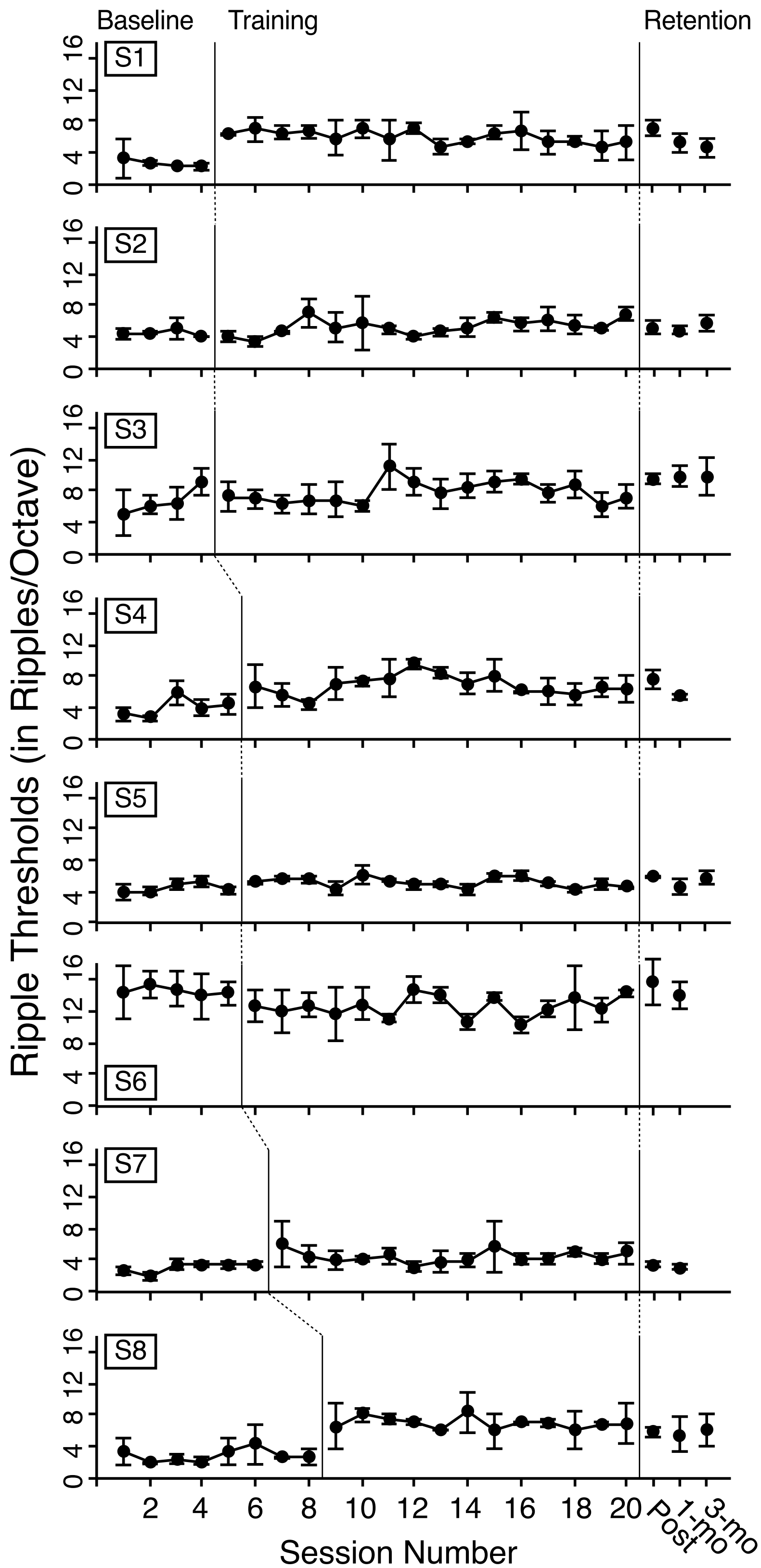

Figure 8. Single-subject data.

B. Does training on spectral ripple discrimination transfer to untrained tasks of speech perception in quiet and in noise, music perception, and increment detection, when compared with an untrained control group?

1. Speech Perception in Quiet

There were no significant improvements in CNC word recognition from pre-training to any subsequent condition in either the trained or control group. The upper panel of Figure 9 shows the percent correct for each phase of the training for the training group (left) and the control group (right), the lower panel shows the mean change in percent correct on CNC words, relative to baseline for the training group (left) and the control group (right). Error bars indicate 95% confidence intervals. Individual subjects are indicated by a unique symbol (see key). In the training group, three individual listeners improved (none identified in the control group) based on critical difference tables for word recognition (Carney and Schlauch, 2007).

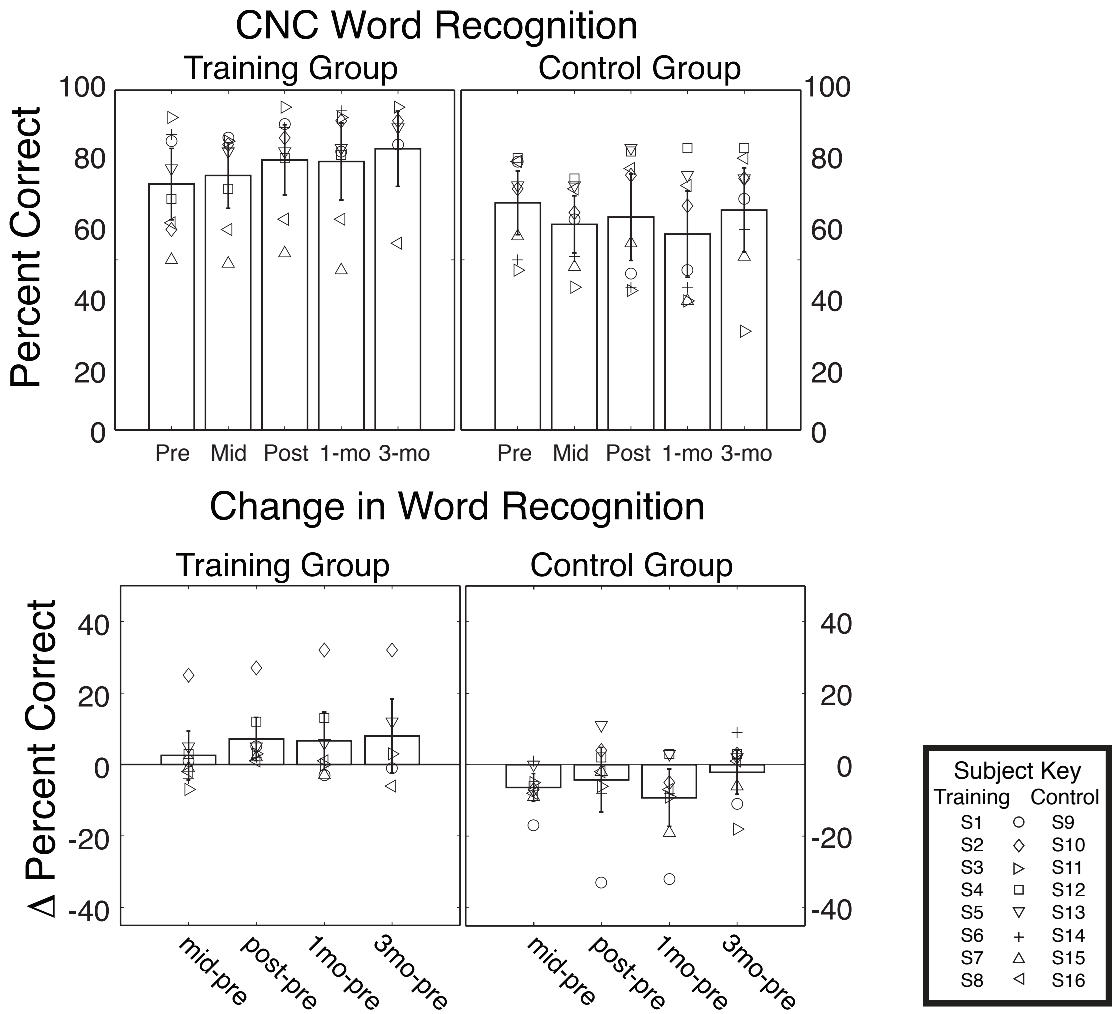

Figure 9. Speech in quiet results.

2. *Speech Perception in Noise*

Speech reception threshold in noise improved relative to baseline for the training group, but not for the control group. The upper panels of Figure 10 show the mean speech reception threshold (in SNR) for the training group (left) and control group (right), and the lower panels show the mean change in signal to noise ratio (SNR) recorded at each phase relative to baseline for the training group (left) and the control group (right). Error bars indicate 95% confidence intervals. Paired samples t-tests indicate a significant improvement in SNR from pre-training to 1-month post-training (-

1.68dB SNR ± 0.68, p = 0.001), with the improvement in the training group greater than the control group (independent samples t-test, p = 0.006). Based on published literature, all eight subjects improved (change exceeding 1 dB SNR) in the training group, and two subjects improved from the control group (Won et al., 2007).

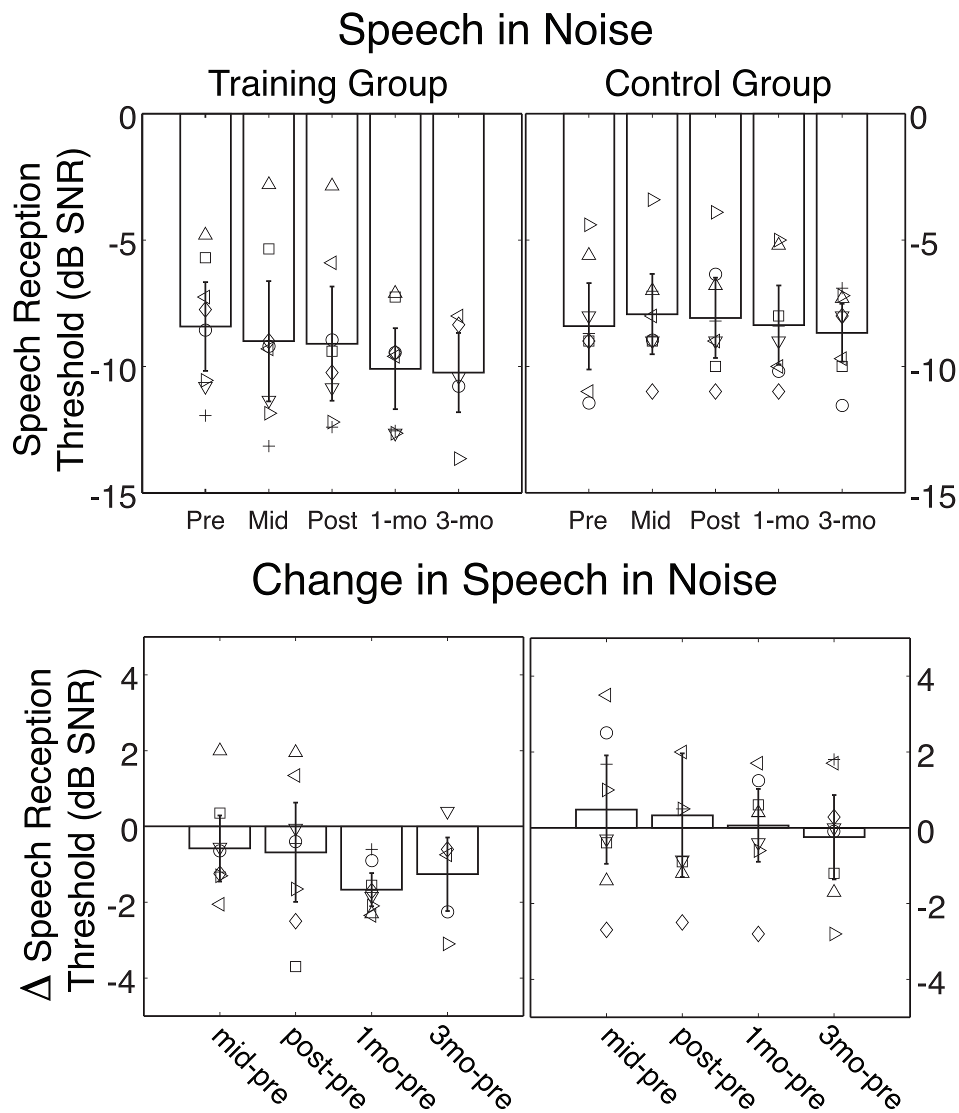

Figure 10. Speech in noise results.

3. *Music Perception*

a. Pitch Direction Discrimination

A significant improvement in pitch perception was noted for 330 Hz for the 1-month post-training phase relative to baseline (paired-sample t-test: -0.659 ±1 0.75 Semitones). However, this improvement was not statistically significantly greater than in the control group (independent samples t-test, p = 0.52). No significant improvement was noted for 226 or 391 Hz in any phase. The left panels of Figure 11 shows the thresholds for pitch direction discrimination for the training group (left) and the control group (right) and the right panels show the mean change in pitch direction detection for each phase, relative to baseline for the training group (left) and the control group (right) for each frequency tested, indicated in each plot (262 Hz, 330 Hz, and 391 Hz). Error bars indicate 95% confidence intervals. Based on published literature, four subjects improved (change exceeding 2 semitones for any of the three pitch tests) in the training group, and two subjects improved from the control group (Kang et al., 2008).

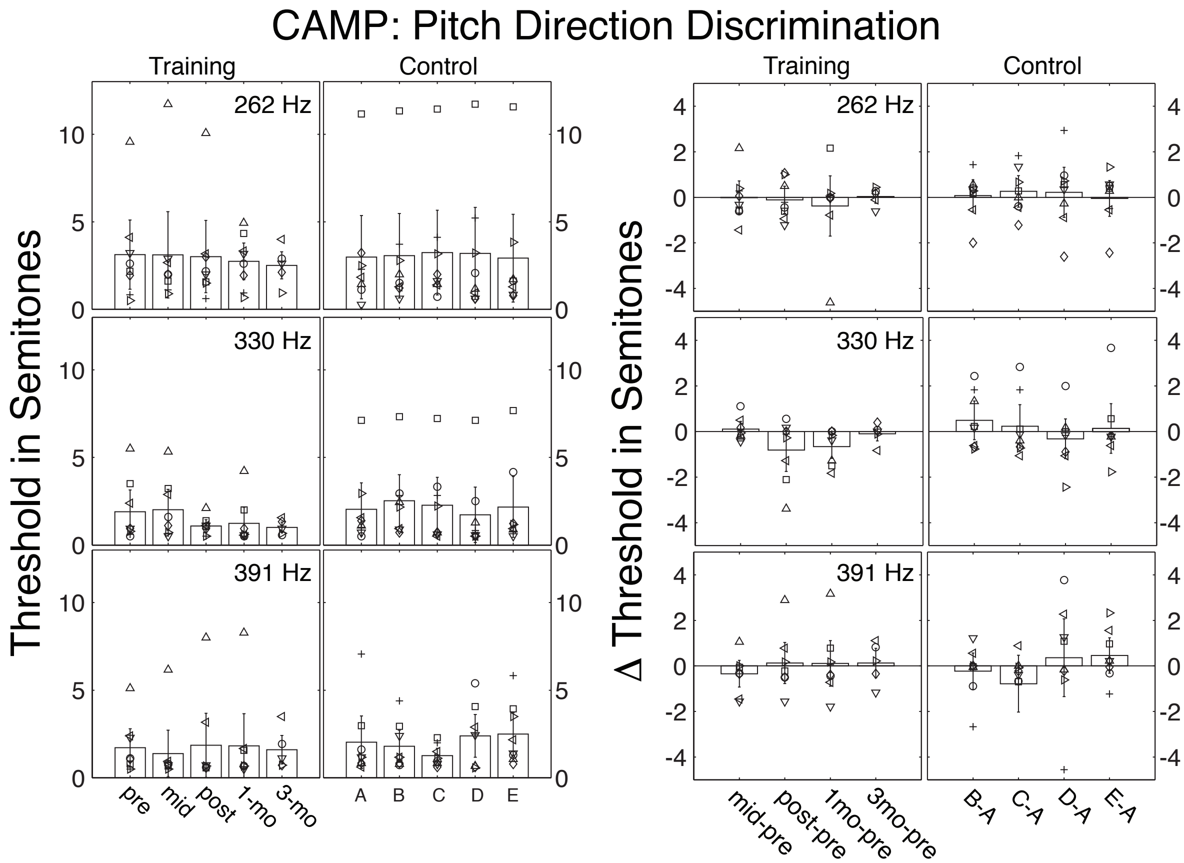

Figure 11. CAMP: pitch results.

b. *Melody and Timbre Identification*

For the trained group, significant improvement in melody identification is noted for both the post-training (11.46 ± 10%, p = 0.015) and 1 month post-training (10.07 ± 9.5 %, p = 0.02) phases relative to baseline and were significantly different than the control group (independent samples t-tests: post-pre p = 0.006; 1 month-pre, p = 0.004). No significant improvement was noted for timbre identification from pre-training to any subsequent condition in either the trained or control group. Figure 12 shows the mean change in melody (12.a) and timbre (12.b) identification, relative to baseline for the training group (left panels) and the control group (right panels). Error bars indicate 95% confidence intervals. Based on published literature, four subjects improved on melody identification (change exceeding 12%), in the training group (none in the control group), and four improved on timbre identification (change exceeding 14%) in the training group, and one subject in the control group (Kang et al., 2008).

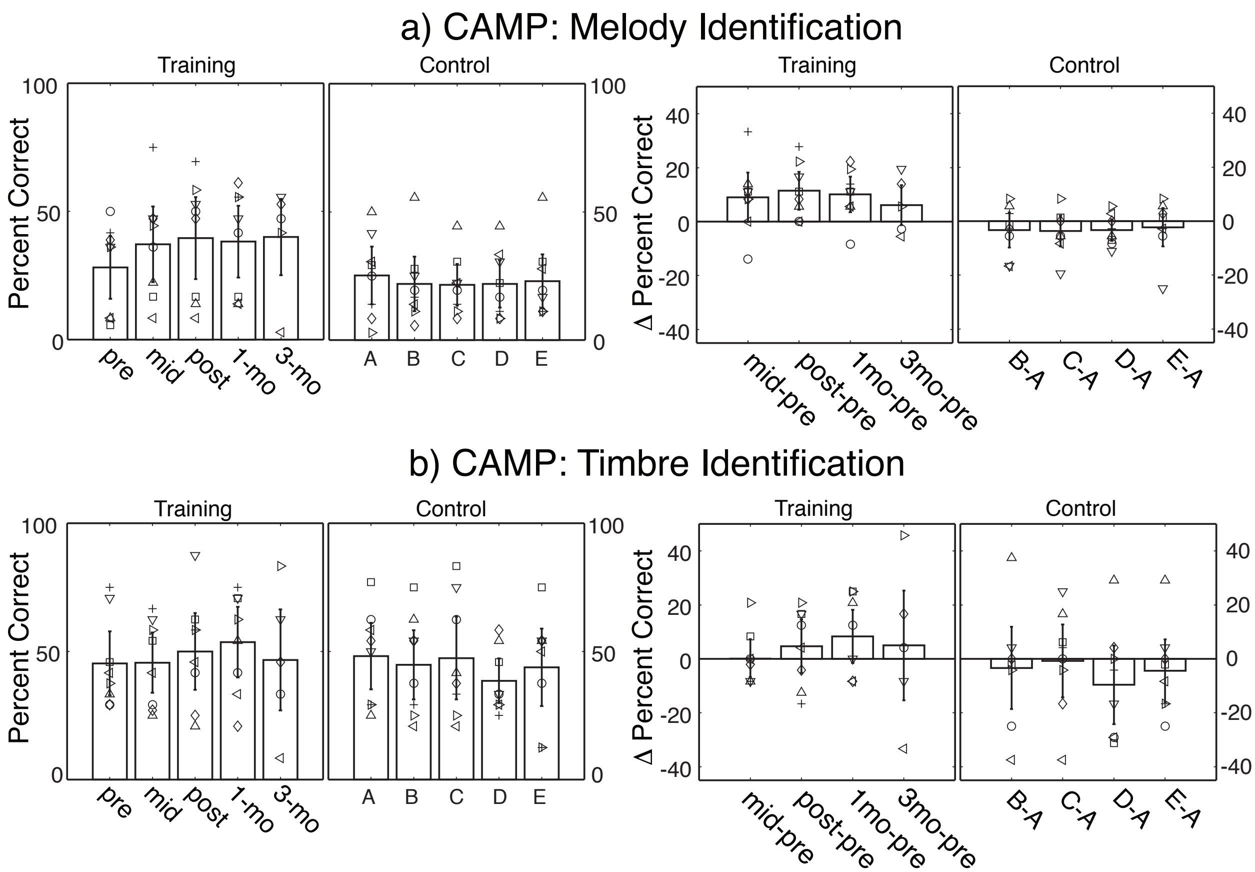

Figure 12. CAMP: melody and timbre results.

4. *Increment Detection*

There were no significant improvements in increment detection thresholds from pre-training to post-training (paired samples t-test: p = .902). Figure 13 shows the increment results for all subjects. Each panel represents an individual subject's pre-training (filled) and post-training (unfilled) increment thresholds, for the 50% and 75% DR conditions, for the three channels tested (see key in figure for channels: indicated by symbol: circle, square, and triangle).

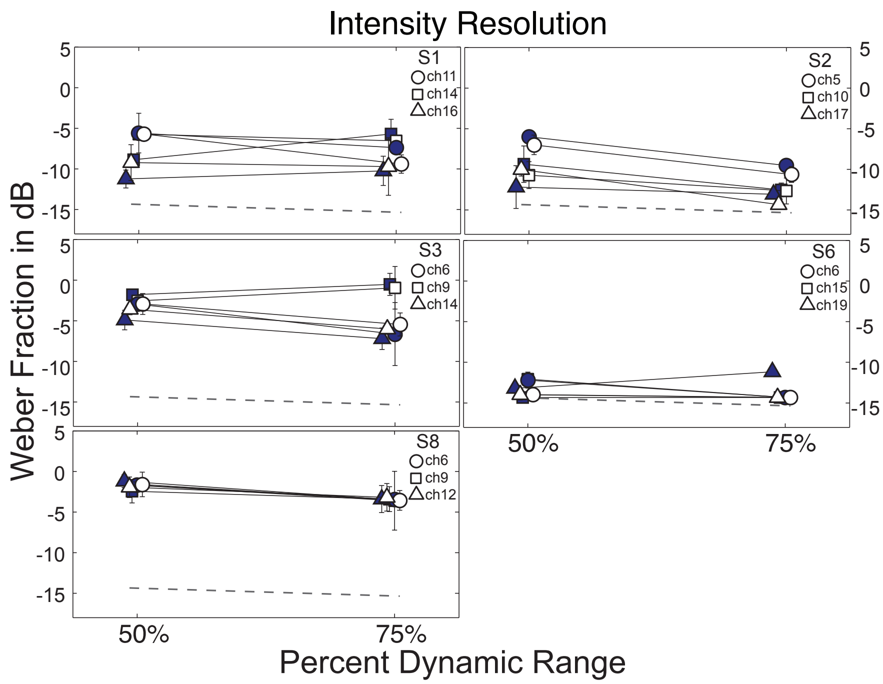

Figure 13. Increment detection results.

5. *Summary*

Table 6 provides a summary of the effects of spectral ripple training on outcome measures more and less dependent on spectral resolution. The only consistent pattern that emerges is that all trained subjects appear to improve their speech in noise performance following training. Individuals who exhibit improved spectral ripple thresholds with training do not demonstrate corresponding changes on other spectrally dependent tests such as speech perception in quiet, speech perception in noise, music perception, or other non-spectral outcomes such as hearing handicap or intensity resolution.

Table 6. Summary.

	Training	**Outcome Measures**						
Subject	Ripple	Speech in noise	Speech in quiet	Music Perception Pitch	Melody	Timbre	Hearing handicap	Intensity resolution
S1	✔	✔						
S4	✔	✔	✔	✔		✔		
S7	✔	✔		✔				
S8	✔	✔		✔				
S2		✔	✔		✔	✔		
S3		✔			✔	✔		
S5		✔		✔	✔	✔		
S6		✔			✔			

C. Does training on spectral ripple discrimination impact overall communication abilities and quality of life using standardized outcome measures of CI success?

There was no significant improvement on perceived hearing handicap from pre-training to post-training in either the trained or control group. The left panel of Figure 14 shows the mean self-reported hearing handicap score for the trained group (left) and the control group (right), and the right panel shows the mean change in self-reported hearing handicap score relative to baseline for the training group (left) and the control group (right). Error bars indicate 95% confidence intervals.

Figure 14. Hearing handicap results.

D. How long are observed changes in performance retained? 1-month? 3-months post-training?

All subjects returned for retention testing one month post training. For those who improved on spectral ripple thresholds, thresholds remained significantly improved relative to baseline. Five of the eight trained subjects returned for 3-month post training testing and improved spectral ripple thresholds were sustained up to 3 months post-training. The observed changes in bias, as indicated by c, were also maintained at those time points. When outcome measures were examined, for those trained subjects with significant improvement, these improvements were sustained up to 3 months post training.

DISCUSSION

A. Overall Findings

The purpose of this experiment was to improve speech in noise and music perception in CI listeners. Our hypothesis was that deficits in speech in noise and music perception arise from degraded spectral resolution. Therefore, our approach was to use auditory training focused on improving spectral resolution with the expectation that these improvements would transfer to tasks dependent upon spectral resolution, and not transfer to non-spectral tasks. In other words, this pattern of transfer of learning would point to a spectral 'mechanism of action' responsible for the gains, and would reflect that the CI user is better able to make use of the spectral information provided by their device. There are two main reasons that we believe that changes in spectral ripple thresholds may not reflect improved spectral resolution, 1) the pattern of transfer of learning, and 2) the influence of response bias on spectral ripple thresholds.

First, as summarized in table 6, there was no clear pattern of transfer of learning. We were successful in improving speech in noise perception for all trained listeners, as well as several subtests of music perception. A single subject design was used to help identify individual patterns of improvement. Several outcome measures and a control group were included to provide evidence of the efficacy of the training program and to assess transfer of learning. Overall, as a group, our results demonstrated that it is possible to improve mean spectral ripple thresholds with training and threshold gains ranging from 1.3 to 4 ripples per octave can be made. However, when single subject data are analyzed, it appears that only four of the eight subjects that were trained improved their spectral ripple thresholds. Further, because not all of the trained subjects showed improvement in spectral ripple thresholds, the resulting improvements in the outcome measures cannot be entirely explained by improved spectral resolution as the mechanism of action. Therefore, questions remain about what actually improved in our training.

Finally, the influence of bias makes it difficult to determine if a change in spectral ripple threshold reflects a change in spectral ripple discrimination, or merely a change in the subject's decision-making process. In this experiment we used a single-interval, yes/no forced-choice task, which can be influenced by listener bias, or tendency of a listener to favor one response over another (Macmillan and Creelman, 2005). Therefore, response bias (criterion location) was tracked along with thresholds (Table 4 and 5). However, because our measure is influenced by bias, a change in threshold might indicate that the subject is becoming more liberal or more conservative in their decision-making. We observed significant changes in bias for all listeners, even for those who did not improve their ripple threshold (Figure 6 and Table 4). The changes in bias are likely related to the introduction of the trial-by-trial feedback provided during the training phase. Therefore, bias appears to be an important factor in this task, suggesting that thresholds involve more than sensitivity, and can reflect changes in the decision-making process. Further data analysis is necessary to make any definitive statements about the relationship between thresholds and sensitivity in this dataset.

B. Alternative Explanations

While the goal of this training program was to target spectral resolution through focused auditory training, the pattern of generalization suggests that these improvements cannot be entirely attributed to improvements in spectral resolution. It is unclear whether the training program is improving perception or if the training merely improves attention on the training task, or overall listening strategy, or reflects changes in response bias. Moore and colleagues (2009; 2010) suggest that learning is influenced by motivation, intelligence, and cognitive factors such as attention and memory. For example, Amitay and Moore (2006) demonstrated that normal hearing subjects trained in frequency discrimination, as well as control subjects actively engaged in a non-auditory task (video game), both showed improvements on a frequency discrimination task relative to baseline testing. The authors conclude that there are likely more broadly-based cognitive mechanisms that are engaged and can influence perception. While cognitive measures, such as working memory, have long been a topic of interest in children with CIs (e.g., Pisoni 2008), less is known about cognition in adults with CIs. Very few studies have explored the role of cognition on outcomes with post-lingually deafened adults with CIs (Heydebrand et al., 2007; Gantz et al., 1993; Knutson et al., 1991). It is possible that the perceptual gains observed in our study reflect the enhancement of more cognitive based mechanisms, such as attention and auditory working memory. Why some people responded and others did not is an important question to consider. It might be that the individuals who improved were more responsive to training because of underlying working memory deficits.

Even though we did not observe a convincing effect of improved spectral resolution with training in our experiment, it does not necessarily mean that improving spectral resolution using spectral ripple stimuli and training is not possible. Our findings might be task- and stimulus-specific. For example, Sabin and Wright (2011) studied perceptual learning using a spectral modulation task, and involved young normal hearing and older hearing-impaired listeners. The spectral modulation task is a variation on the spectral ripple task. In contrast to the spectral ripple discrimination task used in this experiment where the ripple density is varied with a fixed modulation depth, the spectral modulation task employs ripple stimuli with a fixed ripple density and varies the modulation depth. Threshold is defined as the smallest depth (in dB) where the listener is able to differentiate between a modulated and a steady noise signal. For young normal hearing listeners, learning was robust and specific to the trained modulation frequency, meaning that learning did not generalize to untrained modulation frequencies. However, for older hearing-impaired listeners, learning was not as robust and learning generalized to untrained spectral modulation frequencies. Further, their choice of stimulus duration also influenced the outcome (Sabin, personal communication), with less of a learning effect with longer duration stimuli. These results suggest that there might be different mechanisms responsible for the learning patterns observed in these two groups, and that stimulus characteristics and training task might influence outcomes. Even with relatively similar stimuli, the influence of the task, duration, and population, may result in entirely different outcomes (e.g., different time courses and generalization of learning for ITD and ILD: Wright and Zhang, 2009).

Training studies with CI listeners are largely focused on validating clinically available word and sentence based training programs using home-based auditory training (Fu et al., 2005; Wu et al., 2007; Stacey et al., 2010). For example, Fu et al. (2005) show improvements in vowel discrimination (n = 10), consonant discrimination (n = 10), and sentence perception (n = 3) following training. Stacey et al., (2010) replicated this experiment with 11 CI listeners; in contrast to Fu et al.'s study, their participants only showed an improvement of 8% in consonants. However, the program used in Stacey et al. (2010) was not adaptive, and the listeners received less training overall. In these studies, subjects perform the training at home, generally multiple times per week over the course of several weeks, with subjects returning to the lab for outcome measure testing. A problem with this approach is that unsupervised training could lead to differences in performance related to subject engagement, distraction level, and compliance. Also, as previously described by Boothroyd et al. (2010) and Tremblay (2006, 2009, 2011), in the end we do not know what contributed to the perceptual gains or what those who did not improve were missing. Collectively, the examples described here exemplify the complexities of auditory training and the ability to facilitate perceptual change will depend on the population and procedures used.

C. What Did We Train?

If we do not believe that the improvements in spectral ripple thresholds reflect only a change in spectral resolution, then what did the four subjects who improved become better at? It does not appear as if we trained intensity cues because no improvements were noted on intensity resolution following training. Because the sound spectrum is a function of intensity and frequency, improvements gained in one domain (frequency) could have carried over to another domain (intensity). Additionally, the spectral ripple task can be influenced by within-channel and across-channel changes in loudness, particularly at low ripple densities. A significant correlation was found between spectral ripple thresholds and intensity difference limens (mean for all three channels tested; $r = -.910$, $p = 0.032$). It is difficult to know whether the significant correlation reflects a common underlying mechanism. Although spectral ripple discrimination thresholds appeared to improve with training and intensity difference limens did not, the spectral ripple discrimination thresholds were influenced by changes in bias and the intensity difference limens were not. One caveat, however, is that Wright and Zhang (2009) have reported that for normal hearing listeners, training in frequency resolution does not generalize to intensity resolution, yet training in intensity resolution transfers to frequency resolution. In addition, because only a subset of listeners were evaluated with intensity resolution, it is possible that a greater number of subjects may have different outcomes.

Could we have observed incidental learning related to repeated testing? It is difficult to distinguish between changes in behavior that result from improved sensitivity as a result of the auditory training task, versus changes in behavior as a result of repeated-exposure to testing materials (referred to as "incidental-learning"), training-task (independent of the stimuli), procedural-learning, and the contribution of bias (discussed below). To address repeated-exposure to testing materials, we included a control group that received no auditory training, and only returned for the outcome testing at intervals similar to the trained group. We purposefully chose not to expose our control group to our training stimuli to reduce the influence of repeated exposure. Therefore, the inclusion of the 'outcome-measures only' group provided some control over repeated exposure but it is difficult to determine the effect of the training task independent of the training stimuli used. The outcome measures were not repeatedly tested prior to the initiation of training and it is possible that improvements noted on the outcome measures could be a result of repeated testing. However, these tests were chosen because of their high reliability and lack of learning effects. If improvements were noted during a repeated baseline testing phase, some might argue that this reflects purely procedural learning, while others suggest a perceptual component (Moore et al., 2009).

CONCLUSIONS AND FUTURE DIRECTIONS

This single subject design assessed the effects of focused auditory training using spectral ripple stimuli on ripple thresholds and other outcomes measures in adult cochlear implant listeners. The results support the hypothesis that ripple thresholds can be improved through training, with four out of eight trained subjects showing improvements. However, changes in ripple thresholds may not only reflect changes in sensitivity, and may be influenced by several factors, including intensity resolution, cognition (attention, working memory), and the decision-making process (response bias). Subjects in the training group also showed improvements in speech in noise and music perception, while those in the control group did not. Interestingly, even those subjects who did not improve their spectral ripple thresholds appeared to benefit from the training because all of the trained subjects improved in speech in noise, and those who improved on the music subtests were mixed between those who improved and those who did not. It is a function of single subject designs to identify the areas that would require further research. On the basis of the results of this study, we suggest further research into cognitive measures and, task- and stimulus- dependent features, such as learning specificity, to determine the mechanism of action that results in these improvements. Also, future work is necessary to develop approaches to remove the bias from spectral ripple discrimination thresholds.

In conclusion, the results of this study show that the spectral ripple training method is feasible. To better define mechanisms of action that contribute to improved speech perception in noise, a next step will be to apply these methods to a much larger sample. With a greater number of subjects, we can explore individual factors that may influence outcomes. Also we can address the influence of cognition by incorporating pre- and post-training measures of cognition, such as auditory working memory.

CONCLUSION

To summarize, the results of the first experiment suggest that channels with poor electrode-neuron interface show poor spatial selectivity as indicated by the tuning properties. Further, it may not be necessary to use time-intensive techniques, such as PTCs, to arrive at this conclusion because it may be that thresholds measured with a focused stimulus configuration are sufficient to identify these potentially problematic channels. In addition, the results of the second experiment suggest that the slope of the auditory brainstem function is also sensitive to channels with poor electrode-neuron interface. This information helps further support the need to look at alternative mapping strategies to better fine tune the information that is delivered to the auditory system for people who wear cochlear implants with the hope that it might generalize to overall improved perception in speech in noise, music, and quality of life. Results from the third experiment suggest that it is possible to alter someone's perception through the use of auditory training such that the ability to improve ripple thresholds is entirely possible. It appears that everybody benefits from this approach in performance on speech perception in noise tests. However, improvement didn't appear to generalize to tests of speech in quiet, music perception, or quality of life. Moreover, it appears that the patterns of learning appear to be different across individuals. Despite the encouraging results of training using a single-subject research design, unanswered questions remain about what aspects of the training were effective. If the intended goal was to improve spectral resolution, then we might have expected to see generalized improvement to other tasks that depend on the perception of spectral information (e.g., music). This was not the case. One unanswered question is: Are we changing how CI users listen by activating broader non-sensory components, such as attention or memory? Future directions aimed at disentangling these questions will require testing a larger population of CI users. What can be said, however, is that two viable approaches to improving perception in cochlear implant listeners have been identified and that such patient- and device-specific approaches could be refined to better understand the relationship of frequency encoding to perception in adult users of cochlear implants.

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CURRICULUM VITAE

Kathleen F. Faulkner, Ph.D., CCC-A

EDUCATION

2012 Ph.D. Speech and Hearing Sciences
University of Washington
Dissertation: Understanding frequency encoding and perception in adult users of cochlear implants.
Mentor: Kelly Tremblay, Ph.D.

2004 M.S. Speech and Hearing
Washington University in St Louis
Thesis: Afferent innervation of the quail utricle.
Mentor: David Dickman, Ph.D.

2002 B.S. Communication Sciences and Disorders
Saint Louis University

PROFESSIONAL EXPERIENCE

2011 Clinical Supervisor
University of Washington Speech and Hearing Clinic
Department of Audiology, Seattle WA
Worked as a clinical supervisor for first and second year AuD students serving adult and pediatric populations:
Clinical Audiometry, Employee Hearing Screening Program, and Hearing Aids

2004-2005 Clinical Audiologist
University of Colorado Hospital
Department of Audiology, Denver, CO
Worked as a clinical audiologist serving adult and pediatric populations:
Clinical Audiometry, Electrophysiological Assessment, Hearing Aids, and Vestibular Assessment

TEACHING EXPERIENCE

2011 University of Washington, Seattle
Teaching Assistant
SPHSC 570 "Assessment of Auditory Dysfunction I"
SPHSC 576 "Otoacoustic Emissions"

2009 University of Washington, Seattle

Co-taught SPSHC 593 "Electrophysiologic Assessment II"

2006-2008 University of Washington, Seattle
Lead Teaching Assistant for the Department of Speech and Hearing Sciences

2006-2007 University of Washington, Seattle
Teaching Assistant
SPHSC 570 "Assessment of Auditory Dysfunction I"
SPHSC 576 "Otoacoustic Emissions"
SPHSC 563 "Instructional Development Forum"

2005-2006 University of Washington, Seattle
Teaching Assistant
SPHSC 571 "Assessment of Auditory Dysfunction II"
SPHSC 573 "Physiologic Assessment of Auditory Function"
SPHSC 574 "Evaluation of Balance Function"
Guest Lecture: Vestibular Evoked Myogenic Potentials (VEMP)

2004-2005 University of Colorado, Boulder
Teaching Assistant
SLHS 7540 "Vestibular Evaluation I"

LABORATORY EXPERIENCE

2005-2012 Research Assistant to Kelly Tremblay, Ph.D.
University of Washington, Seattle
Brain and Behavior Lab
Performed basic laboratory duties, including data collection and analysis, Matlab programming, and processing of EEG data.

2003-2004 Research Assistant to J. David Dickman, Ph.D.
Washington University, St. Louis
Vestibular Function, Regeneration, & Spatial Orientation Laboratory
Performed basic laboratory duties, including multiple histological tissue preparations and tracing of stained neural tissue using a high-resolution microscope.

2002-2003 Research Assistant to Kevin Ohlemiller, Ph.D.
Washington University, St. Louis
Mechanisms of Cochlear Injury Laboratory
Performed basic laboratory duties, including embedding mice cochlae in plastic and recording ABR in anesthetized mice.

PEER-REVIEWED PUBLICATIONS

Bierer, J.A., **Faulkner, K.F.** and Tremblay, K.L. (2011) Identifying Cochlear Implant Channels With Poor Electrode-Neuron Interfaces: Electrically Evoked Auditory Brain Stem Responses Measured With the Partial Tripolar Configuration. *Ear and Hearing*, July/August;32(4): 436-444.

Huss, D., Navaluri, R., **Faulkner, K.F.**, Dickman, D. (2010) Development of otolith receptors in Japanese Quail. *Developmental Neurobiology*, May;70(6):436-55.

Bierer, J.A. and **Faulkner, K.F**. (2010) Identifying cochlear implant channels with poor electrode-neuron interface: partial tripolar: single-channel thresholds and psychophysical tuning curves. *Ear and Hearing*, April;31(2):247-58.

PEER-REVIEWED POSTER PRESENTATIONS

Faulkner, K.F., Tremblay, K.L., Rubinstein, J.T., Werner, L.A., Nie, K. Auditory Learning: Can Adult Cochlear Implant Listeners Learn to Hear Changes in Spectrally-Rippled Noise Stimuli? Aging and Speech Communication Research Conference (October, 2011).

Faulkner, K.F., Tremblay, K.L., Rubinstein, J.T., Werner, L.A., Nie, K. Auditory Training in Adult Cochlear Implant Listeners using Spectrally-Rippled Noise Stimuli in an Adaptive, Single-Interval, Paradigm. Conference on Implantable Prostheses (July, 2011).

Faulkner, K.F., Tremblay, K.L., Bierer, J.A. Characterizing channels with high thresholds using electrically-evoked auditory brainstem responses. Objective Measures in Auditory Implants Sixth International Symposium (September 2010).

Faulkner, K.F., Tremblay, K.L., Rubinstein, J.T., Werner, L.A., Nie, K. Auditory Training in Adult Cochlear Implant Listeners using Spectrally-Rippled Noise Stimuli in an Adaptive, Single-Interval Paradigm – Preliminary Results. Association for Research in Otolaryngology MidWinter Meeting (February 2010).

Faulkner, K.F., Tremblay, K.T., Bierer, J.A. Electrically-evoked auditory brainstem responses with the partial tripolar configuration. NIH-sponsored mentored student poster presentation at the American Auditory Society Meeting (April 2009).

Bierer, J.A., **Faulkner, K**., Tremblay, K.T. Probing the cochlea with partial-tripolar cochlear implant stimulation: The feasibility of electrically-evoked auditory brainstem measures. Association for Research in Otolaryngology MidWinter Meeting (February 2009).

Faulkner, K., Tremblay, K.T., Bierer, J.A. Psychophysical tuning curves and electrically-evoked auditory brainstem responses with the partial tripolar electrode configuration. Northwest Auditory Meeting, Portland, OR (October 2007).

Faulkner, K., Tremblay, K.T., Bierer, J.A. Psychophysical tuning curves and electrically-evoked auditory brainstem responses with the partial tripolar electrode configuration. Conference on Implantable Auditory Prostheses (July 2007).

Faulkner, K., Huss, D., Dickman, J.D. Afferent innervation of the quail utricle. American Academy of Audiology, National Convention (March 2005).

Ortmann, A.J., **Faulkner, K.F.,** Gagnon, P.M., Ohlemiller, K.K. Removal of the Ahl Allele from the C5Bl/6 Background Does Not Improve Noise Resistance. Association for Research in Otolaryngology MidWinter Meeting (February 2004).

NON-PEER REVIEWED PUBLICATIONS

Faulkner, K., Clinard, C., Tremblay, K. Moment of Science: A goldilocks perspective on auditory training: How much is just right? *Audiology Today*, publication of the American Academy of Audiology, 19:5 (Sept/Oct) 2007, p.31.

Faulkner, K., Tremblay, K. Moment of Science: Babies Beware: Noise exposure might affect your hearing later in life. *Audiology Today*, publication of the American Academy of Audiology,19:4 (May/June) 2007, p.53.

Faulkner, K., Tremblay, K. Moment of Science: Is it possible to improve binaural hearing? *Audiology Today*, publication of the American Academy of Audiology, 19:1 (Jan/Feb) 2007, p.53.

Faulkner, K., Friesen, L., Tremblay, K. Moment of Science: A neural perspective on cochlear implants. *Audiology Today*, publication of the American Academy of Audiology, 18:5 (Sept/Oct) 2006, p.46.

Taleb, M., **Faulkner, K.**, Cunningham, L. Moment of Science: New concepts in hair cell regeneration. *Audiology Today*, publication of the American Academy of Audiology, 18:4 (July/Aug) 2006, p.42.

INVITED PRESENTATIONS:

Faulkner, K.F. "Auditory training to improve spectral resolution in cochlear implant listeners." Indiana University Speech Perception Laboratory. October 2011.

Faulkner, K.F. "Auditory Learning in Individuals who wear Cochlear Implants." Center for Integrative Neuroscience Spring Symposium – Learning and Memory: *mechanisms, function, applications.* May 2011.

CERTIFICATION AND LICENSING

ASHA Certification: Audiology
Washington State Department of Health: Audiologist License

PROFESSIONAL ASSOCIATIONS

American Speech Language and Hearing Association
Academy of Rehabilitative Audiology
Acoustical Society of America
American Auditory Society
Association for Women in Science

AWARDS AND HONORS

2011	Student Travel Award to attend the Indiana Aging and Speech Communication Research Conference (October, 2011).
2011	Student Travel Award to attend the Conference on Implantable Auditory Prostheses (July 2011).
2010	New Century Scholars Program Doctoral Scholarship awarded by the American Speech Language Hearing Foundation.
2010	Olswang Endowed Graduate Student Conference Fund Award to attend the Association for Research in Otolaryngology MidWinter Meeting, Anaheim, CA (February 2010).
2010	University of Washington Graduate Student Conference Travel Award to attend the Association for Research in Otolaryngology MidWinter Meeting, Anaheim, CA (February 2010).
2010	Audiologist travel award for the Association for Research in Otolaryngology (ARO) MidWinter Meeting by the American Academy of Audiology Research Committee funded by the Collegium Oto-Rhino-Laryngologicum Amicitiae Sacrum – US Group, Inc. (CORLAS), Anaheim CA (February 2010).
2009	NIH-sponsored student poster presentation at the American Auditory Society (AAS) Meeting, Scottsdale, AZ (March 2009).
2008	Best of Audiology Literature 2007 – for the publication "Moment of Science," Noise exposure might affect your hearing later in life. Acknowledged in *The Hearing Journal*, June 2008, Vol. 41, No.6.
2007	Travel Scholarship for the Conference on Implantable Auditory Prostheses (CIAP).
2007-2011	Ad Hoc reviewer for Ear and Hearing.
2006-2009	NIH-NIDCD T32 training grant trainee.
2005	Top Scholar Scholarship Program (Autumn Quarter) University of Washington, Seattle

RESEARCH SUPPORT

NIH NIDCD F31 DC010309 (Faulkner, PI) 9/2009 – 3/2011
"Auditory Training to Improve Spectral Resolution in Cochlear Implant Listeners"
To examine the use of focused auditory training exercises to improve perception of spectral details, as well as speech in noise and music perception.
Sponsor Kelly Tremblay, Ph.D.; Co-sponsor Jay T. Rubinstein, M.D., Ph.D.

NIH NIDCD T32-DC005361 (Covey, PI) 6/2006 – 8/2009
Predoctoral Trainee on Auditory Neuroscience Training Grant.