Factors Affecting Binaural Unmasking in Listeners with Cochlear Implants

By

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To my father, Leon

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Table of Contents

Abstract

Many listeners with bilateral cochlear implants show sensitivity to binaural information when stimulation is provided with single electrodes in both ears. However, there is wide variability in binaural hearing performance with single-electrode stimulation, and performance with multi-electrode stimulation can degrade relative to performance with single-electrode stimulation. Two studies were conducted to further our understanding of the binaural hearing performance of listeners with cochlear implants.

In Study 1, binaural unmasking i.e., the improvement in signal detection between diotic and dichotic stimulus conditions, was examined in children with bilateral cochlear implants. Diotic and dichotic signal detection thresholds with multi-electrode stimulation, using three electrode-pairs (three electrodes in each ear) spaced widely along the electrode array, were examined relative to performance with each of the three electrode-pairs individually. Of interest was whether multi-electrode performance was better or worse than the best single-electrode performance. Additionally, Study 1 sought to determine whether interaural time difference sensitivity is advantageous for binaural unmasking in listeners with cochlear implants by comparing the magnitude of binaural unmasking between children who showed interaural time difference sensitivity and those who did not show interaural time difference sensitivity.

In Study 2, the relationship between binaural and monaural hearing performance was examined in adults with cochlear implants using single-electrode stimulation. It was hypothesized that binaural sensitivity is affected by characteristics at the auditory periphery and may show a relationship with monaural hearing performance. Binaural measures including

dichotic signal detection and interaural time difference discrimination were examined. Monaural measures including dynamic range and amplitude modulation detection, were obtained in each ear. In addition, differences in loudness growth between ears were examined in relationship to dichotic signal detection, to investigate whether asymmetries in loudness growth interfere with processing of binaural information in stimuli which vary in intensity over time.

These two studies attempted to improve understanding the binaural hearing performance of listeners with cochlear implants with both single- and multi-electrode stimulation. A greater understanding of binaural hearing performance of listeners with cochlear implants may lead to interventions which allow listeners with cochlear implants to receive greater advantages from binaural hearing.

Chapter 1

Background and Introduction

BINAURAL HEARING

Binaural hearing plays a major role in sound source localization and understanding speech in noise. Binaural hearing depends on sensitivity to interaural time differences (ITDs) and interaural level differences (ILDs). ITDs and ILDs are determined by the location of a sound source in the horizontal plane relative to the listener's head. ITDs depend on the difference in distance between the source and each of the ears, with the maximum ITD being approximately 800 µs in humans (Abbagnaro *et al.*, 1975). Normally hearing (NH) listeners are able to process ITDs in temporal fine structure at low frequencies up to approximately 1.3 kHz (Zwislocki and Feldman, 1956; Yost, 1977). At frequencies above 1.3 kHz, listeners are able to use ITDs in temporal envelope (Bernstein and Trahiotis, 1994). The difference in distance between the sound source and each of the ears affects ILDs to a small extent, but the majority of the ILD results from the interference of the listener's head with the sound, which reduces the level of the sound at the ear farther from the source. ILDs depend on frequency, with higher frequencies generally causing greater ILDs (Abbagnaro *et al.*, 1975).

ITDs and ILDs are processed differently in the brainstem. The processing of ITDs is believed to occur in the medial superior olive where firing rates of neurons are affected by the extent of coincidence in excitatory inputs (Yin and Chan, 1990; Joris *et al.*, 1998). The processing of ILDs and envelope ITDs is believed to occur in the lateral superior olive where the firing rates of neurons are affected by the integration of information from ipsilateral excitatory inputs and contralateral inhibitory inputs (Joris and Yin, 1995; Tollin, 2003).

Spatial Release from Masking

Having two ears improves speech reception in the presence of interfering sounds when sound sources are spatially separated compared to when sound sources are co-located. This effect is referred to as spatial release from masking. Three factors are thought to contribute to this effect (Ihlefeld and Litovsky, 2012).

First, when sounds are spatially separated, there are spatial configurations of sound sources which result in one of the ears having a higher signal-to-noise ratio (SNR) than the other, specifically, the ear closer to the target source or farther from the masker source. This effect, referred to as head shadow, can result in 12 dB of improvement in speech reception thresholds (Jones and Litovsky, 2011). Head shadow does not involve binaural hearing per se but rather the ability of the listener to attend to the ear with the higher SNR.

Second, we consider cases when informational masking occurs. Informational masking is masking that occurs centrally rather than peripherally due to for example, when the listener is distracted by the masker due to masker uncertainty (Watson *et al.*, 1976; Lutfi *et al.*, 2003). Speech maskers are an example of maskers which are thought to produce informational masking (Freyman *et al.*, 1999). Hawley *et al.* (2004) reported that spatial separation with speech maskers resulted in 6 to 7 dB of improvement in speech reception thresholds which could be attributed to binaural hearing. In this situation, spatial separation of sound sources is thought to provide the listener with interaural cues associated with each sound source, which help the listener segregate the target speech from the masker and attend to the target (Freyman *et al.*, 1999).

Third, when there is spatial separation between sound sources and spectral overlap i.e., energetic masking, between the target and masker, interaural de-correlation (interaural dissimilarity) is created in the combination of the target and maskers, which is thought to improve reception of the target speech (Bronkhorst and Plomp, 1988; Akeroyd and Summerfield, 2000; Hawley *et al.*, 2004). This can be demonstrated by presenting speech in broadband noise and comparing speech reception thresholds between a condition in which the speech and noise have the same interaural parameters, the diotic condition, and a condition in which the speech and noise have different interaural parameters, the dichotic condition (Schubert, 1956; Levitt and Rabiner, 1967a). The difference between the diotic and dichotic condition has been referred to as the binaural intelligibility level difference. The binaural intelligibility level difference has been found to be approximately 3 dB when there is a non-zero ITD in the target speech and a zero ITD in the noise. An interaural-phase delay of π radians increases the binaural intelligibility level difference to 6 dB (Levitt and Rabiner, 1967a). It has been suggested that the binaural intelligibility level difference arises from the listener's integration of a binaural spectrum (de-correlation as a function of frequency) with the monaural spectrum from each ear to help improve reception of the target speech (Akeroyd and Summerfield, 2000). Evidence for this idea was provided by Akeroyd and Summerfield (2000) who showed that listeners were able to identify synthetic vowels created by introducing an interaural correlation of zero at frequencies corresponding to each vowel's first and second formant in broad-band noise which was otherwise diotic.

Binaural Masking Level Differences

This dissertation focuses on measures associated with the binaural intelligibility level difference. Similar to the binaural intelligibility level difference, the binaural masking level difference (BMLD) refers to the measured difference in signal detection thresholds between a diotic and a dichotic condition. The BMLD is of interest, because it is thought to reflect listeners' sensitivity to binaural cues that are relevant for the binaural intelligibility level difference (Levitt and Rabiner, 1967b; a; Culling *et al.*, 2001). A common way to measure a BMLD is to compare a listener's signal detection thresholds between a diotic condition (NoSo) and a dichotic condition (NoSπ). In these conditions, *No, So, and Sπ* refer to noise with 0 interaural phase difference, a target signal with 0 interaural phase difference, and a target signal with a π radians interaural phase difference, respectively. The size of the BMLD can be as large as 25 dB SNR (van de Par and Kohlrausch, 1997). For any particular set of interaural parameters, the BMLD depends on multiple factors including the center frequency of the signal, the bandwidth of the noise, and the level of the noise (Wilbanks and Whitmore, 1968; Yost, 1988; van de Par and Kohlrausch, 1999). For stimuli presented over headphones, interaural de-correlated sounds (e.g., NoSπ) have been described as having a larger perceived width in the head than interaurally correlated sounds (e.g., NoSo) (Whitmer *et al.*, 2012). Figures 1 and 2 show NoSo and NoSπ waveforms, respectively; lower SNRs are shown in lower panels.

Interaural correlation (or interaural coherence) of a signal can be measured by calculating the cross-correlation between the signals presented to the left and right ears such that one calculates the correlation of the left and right signals at all interaural time delays between the two signals. The interaural correlation is the maximum correlation resulting from the cross-correlation of the left and right signals (Goupell and Hartmann, 2006; Whitmer *et al.*, 2012). For a variety of stimuli, the calculated interaural correlation can reasonably describe the size of BMLDs produced by listeners (Osman, 1971). For an NoSπ stimulus, interaural de-correlation occurs in both the temporal fine structure and temporal envelope of the stimulus (van de Par and Kohlrausch, 1995). Therefore, listeners can show BMLDs for high frequency stimuli with which the listener does not have access to temporal fine structure. However, listeners show larger BMLDs at low frequencies compared to high frequencies suggesting that sensitivity to interaural correlation in temporal fine structure results in better performance for binaural signal detection (Schubert and Schultz, 1962; Durlach, 1964; Eddins and Barber, 1998). As a function of SNR, the interaural correlation of the NoSπ stimulus is near -1 (anti-correlated) at high SNRs, near 0 (de-correlated) at 0 dB SNR, and near 1 (correlated) at low SNRs. The interaural correlation[1] of the temporal envelope of an NoSπ stimulus (in which the target (S) is a sinusoid) is 1 at both low and high SNRs and approximately .75 at 0 dB SNR (van de Par and Kohlrausch, 1995). Both the temporal envelope of the noise (No) and the envelope of the tone (Sπ) are correlated which is the reason for high temporal envelope correlation at both low and high SNRs. Interaural de-correlation of temporal fine structure and temporal envelope has also been described as time-varying ITDs and ILDs, respectively. The variance of ITDs and ILDs in a stimulus can be used to predict listeners' performance in dichotic signal detection and interaural correlation discrimination (Zurek, 1991; Goupell and Hartmann, 2006; van der Heijden and Joris, 2010).

COCHLEAR IMPLANTS

The populations of interest for this dissertation are children and adults with cochlear implants (CIs). CIs are devices that can allow individuals to hear through electrical stimulation of the auditory nerve. There are typically 12 to 22 electrodes on an array which is placed in the cochlea (Loizou, 2006). The target location for the electrode array is the scala tympani of the cochlea. The cochlea is a spiral structure consisting of approximately 2.6 turns (Erixon *et al.*, 2008). Electrode arrays typically reach insertion depths of approximately 1.25 turns (Finley *et al.*, 2008). Therefore electrode arrays typically sit more in the basal portion of the cochlea. The insertion depth and proximity of the electrode array to the modiolus is in part affected by the type of electrode array that the listener uses (Cohen *et al.*, 2003; Kos *et al.*, 2005). The return electrode(s) can either be one or more electrodes in the cochlea and/or one or more electrodes outside of the cochlea (Bierer, 2010). The use of return electrodes farther from the active electrode produces greater spreads of excitation and lower audibility thresholds and maximum comfort levels (Chatterjee, 1999; Chatterjee *et al.*, 2006; Bierer and Nye, 2014).

Typically, CIs provide stimulation through biphasic current pulses. Pulsatile stimulation was introduced for multi-channel stimulation so that electrodes could be activated non-simultaneously (Wilson *et al.*, 1991). Non-simultaneous activation is desirable because simultaneous activation of multiple electrodes is thought to result in interaction of current from different channels (Shannon, 1983b). Rates of stimulation for everyday listening are typically 900 to 2500 pulses per second (Loizou, 2006).

Loudness

Loudness can be manipulated for listeners with CIs by changing the current amplitude (in microamperes) of pulses, the pulse rate (number per second), or the phase duration (in microseconds) of pulses. Higher current amplitudes, rates, and longer phase durations all produce louder percepts (Chatterjee *et al.*, 2000; Kreft *et al.*, 2004). Typically, loudness is manipulated by changing the current amplitude while the pulse rate and phase duration of the pulses are held constant (Loizou, 2006). Loudness as a function of current in microamperes grows more quickly than loudness as a function of acoustic amplitude. Loudness growth with electric hearing has been described as an exponential function or as a power function in which the power is approximately 3, whereas loudness growth with acoustic hearing has been described as a power function in which the power is .6 (Stevens, 1955; Fu and Shannon, 1998; Chatterjee *et al.*, 2000).

When loudness is manipulated by changing the current amplitude, the dynamic range of listeners with CIs is approximately 6 to 15 dB, which is smaller than the dynamic range of listeners with normal hearing which is greater than 100 dB (Viemeister, 1988; Zeng *et al.*, 2002; Kreft *et al.*, 2004). However, detection of changes in intensity with electric hearing can be better than it is with acoustic hearing (Nelson *et al.*, 1996). At moderate to high levels of stimulation many listeners show ceiling level performance for intensity discrimination and amplitude modulation detection with the intensity resolution of the CI limiting performance (Galvin and Fu, 2009; Chatterjee and Yu, 2010).

Pitch

Pitch can be manipulated by changing the stimulating electrode or changing the pulse rate or the amplitude-modulation rate of high-rate pulse trains (McKay *et al.*, 1994; McKay *et al.*, 2000). Stimulation of different electrodes should ideally take advantage of the tonotopic organization of the cochlea i.e., electrodes located more basally should produce higher pitch percepts and electrodes located more apically should produce lower pitch percepts. Listeners with CIs have shown the ability to discriminate pitch based on changes in the electrode which produces stimulation (Donaldson *et al.*, 2005). Listeners with CIs are also able to discriminate pitch based on pulse rate at reference pulse rates below approximately 300 pulses per second (Shannon, 1983a; Townshend *et al.*, 1987).

BINAURAL HEARING IN ADULTS WITH COCHLEAR IMPLANTS

Bilateral CIs are being provided in attempts to provide listeners with the advantages that come from hearing with two ears, namely sound localization along the azimuth and improvements in understanding speech in noisy environments. Although having two CIs has been shown to improve sound source localization (Litovsky *et al.*, 2004; Verschuur *et al.*, 2005), listeners with bilateral CIs have poorer sound localization than NH listeners (Verschuur *et al.*, 2005; Majdak *et al.*, 2011). In addition, the contribution of binaural hearing to speech understanding in the presence of interferers is poor for listeners with bilateral CIs (van Hoesel *et al.*, 2008; Loizou *et al.*, 2009; Van Deun *et al.*, 2010; Misurelli and Litovsky, 2012). Improvements in understanding speech in noise shown by listeners with two CIs have largely been attributed to head shadow (Schleich *et al.*, 2004; Loizou *et al.*, 2009). Because a major difficulty for listeners with CIs is understanding speech in complex acoustic environments (Friesen *et al.*, 2001; Nelson *et al.*, 2003; Stickney *et al.*, 2004), better access to binaural information is likely to benefit listeners with bilateral CIs.

An initial step in determining why listeners with CIs generally underperform NH listeners on tasks involving binaural hearing, has been to investigate listener sensitivity to interaural cues with electrical stimulation. Since there are aspects of CI signal processing that prevent the transmission of interaural information, direct electrical stimulation has been used to bypass clinical speech processors to send controlled signals to select left-right electrode pairs. With direct electrical stimulation to single pairs of electrodes, adults with CIs have shown sensitivity to ITDs in low-rate pulse trains (van Hoesel, 2007; van Hoesel *et al.*, 2009; Litovsky *et al.*, 2012). The highest pulse rate at which listeners are able to discriminate ITDs is listener and stimulation site dependent but is typically well below the 1300 Hz limit found with acoustic temporal fine structure (van Hoesel and Tyler, 2003; Majdak *et al.*, 2006; van Hoesel *et al.*, 2009). Listeners with CIs can show ITD discrimination with high-rate pulse trains when the amplitude of the pulse trains are modulated or when there is jitter in the pulse timing (Laback and Majdak, 2008; van Hoesel *et al.*, 2009). Litovsky *et al.* (2010) tested three adults who were pre-lingually deaf, and none demonstrated ITD sensitivity; however, nearly every adult with childhood-onset and adult-onset of deafness demonstrated ITD sensitivity. These data suggest that either long-term deprivation or deprivation early in life may be detrimental for ITD sensitivity. Sensitivity to ILDs is studied less often than ITDs with direct electrical stimulation, because ILD stimuli, by design, involve monaural cues. However, when ILD sensitivity has been examined, most adults with CIs show the ability to discriminate ILDs and lateralize ILDs (i.e., indicate changes in intracranial position of a sound that are consistent with changes in ILDs) (Litovsky *et al.*, 2010; Kan *et al.*, 2013).

Binaural Masking Level Differences in Adults with CIs

In addition to ITD sensitivity, listeners with CIs have also shown BMLDs with direct electrical stimulation. Using stimulation of a single pair of electrodes, Long *et al.* (2007) found an average BMLD of approximately 12 dB in 7 adults with CIs. Lu *et al.* (2010) used a similar approach, and found a smaller average BMLD of 4.6 dB in 5 adults with CIs. Goupell and Litovsky (2015) found a somewhat larger average BMLD of 8.5 dB for 10 adults with CIs. Goupell and Litovsky (2015) also showed that listeners with CIs can discriminate changes in interaural correlation.

Most studies that have examined BMLDs of listeners with CIs used transposed stimuli in which the electrical pulse trains are modulated by an envelope which was calculated through half-wave rectification. With this method of calculating the temporal envelope, the temporal fine structure (in addition to the temporal envelope) of the unprocessed stimuli appears as temporal envelope and can be presented at higher center frequencies to NH listeners. van de Par and Kohlrausch (1997) showed that BMLDs of NH listeners were larger with transposed stimuli than with non-transposed high-frequency stimuli. However, listeners with CIs have also demonstrated BMLDs with non-transposed stimuli (Goupell and Litovsky, 2015). For listeners with CIs, BMLDs with non-transposed stimuli depend on detection of interaural de-correlation in slower modulations of the electrical pulse train, which are due to the temporal envelope as opposed to the fine structure of the unprocessed stimulus. There is evidence which suggest that listeners with CIs rely on these slower modulations for binaural unmasking even with transposed stimuli (Long *et al.*, 2006).

To date, much of the research on BMLDs (and interaural correlation discrimination) of listeners with CIs has examined the effects of different aspects of signal processing. Long *et al.* (2006) found that non-linear compression (which is typical for CI signal processing) as opposed to linear compression between the acoustic stimuli and the electric stimuli resulted in smaller BMLDs. Goupell and Litovsky (2015) found poorer interaural correlation discrimination at lower pulse rates (100 vs. 1000 pulses per second) which they suggested was due to poorer representation of temporal envelope at lower pulse rates or smaller dynamic ranges with lower pulse rates. Todd *et al.* (2014) found that timing the electrical pulses with the temporal fine structure of the unprocessed stimulus resulted in an improvement in BMLDs only at low pulse rates (i.e., 125 pulses per second) such that performance at low pulse rates with temporal fine structure encoded was similar to that of performance at high rates without temporal fine structure encoded.

While single-electrode stimulation provides excellent information about binaural processing when one area of the cochlea is stimulated, this approach is unrealistic, as speech processors rely on multi-electrode stimulation to provide CI users with access to the spectral structure of sounds. To date, two studies have examined BMLDs under multi-electrode stimulation. Lu *et al.* (2011) found that greater overlap in excitation patterns from neighboring electrodes resulted in larger NoSo and NoSπ thresholds as well as smaller BMLDs when NoSo or NoSπ stimuli were presented from one electrode pair and diotic noise was presented from one or more neighboring electrode pairs. Van Deun *et al.* (2011) showed that BMLDs were greatly reduced during multi-electrode stimulation when the electrodes (in each ear) presented different samples of noise compared to when the electrodes presented the same sample of noise. These studies suggest that neighboring electrodes can produce energetic masking which reduces the BMLD. One question addressed in Chapter 2 of this dissertation is how listeners with CIs perform on BMLD tasks with multi-electrode stimulation when energetic masking is reduced. It is unknown whether having multiple channels of information can improve performance and whether poorer performing stimulation sites can negatively affect performance.

Interaural Pitch Matching

In attempts to optimize binaural sensitivity, a number of studies have examined binaural sensitivity with pitch-matched electrode pairs (van Hoesel and Tyler, 2003; Majdak *et al.*, 2006; Laback and Majdak, 2008; van Hoesel *et al.*, 2009; Lu *et al.*, 2010). Choosing electrode pairs for measures of binaural sensitivity by pitch matching is based on models of central binaural processing which suggest better binaural sensitivity when input arises from corresponding places in the cochleae (which would produce the same pitch percepts) (Tollin, 2003). Furthermore, pitch-matching is based on data from NH listeners that show that binaural sensitivity deteriorates with increasing interaural mismatch in the center frequency of the stimuli (Henning, 1974; Nuetzel and Hafter, 1981). Since electrode insertion depth and positioning can vary between different ears, there is no certainty that electrodes in left and right ears with corresponding numbers will stimulate corresponding locations on the auditory nerves (Finley *et al.*, 2008). There is some evidence that interaural pitch-matching increases the likelihood of stimulating at corresponding locations in the cochleae. Both van Hoesel and Clark (1997) and Long *et al.* (2003) (3 participants in total) found that differences between the ears (left vs. right) in pitch judgments for individual electrodes corresponded well with interaural offsets in the positioning of the left and right electrode arrays. Furthermore, similar to NH listeners, ITD sensitivity of listeners with CIs has been found to systematically decrease as one of the electrodes of a pair moves away from the electrode which produces the best ITD sensitivity (Poon *et al.*, 2009). Kan *et al.* (2013) also found that the proportion of interaurally "fused" percepts reported by participants decreased as the interaural mismatch increased relative to the pitch-matched pair. Furthermore, Kan *et al.* (2013) also found that the range of intra-cranial positions produced by ITDs that participants reported decreased as interaural mismatch increased. Both the perception of interaural fusion and the range perceived intra-cranial positions can be understood to be relevant for binaural hearing performance. Therefore, we may expect that choosing electrode pairs based on pitch-matching may improve binaural sensitivity. However, it should be noted that the electrode pair which produces the best ITD sensitivity is not always the pitch-matched electrode pair (van Hoesel and Clark, 1997; Long *et al.*, 2003; Poon *et al.*, 2009).

BINAURAL HEARING OF CHILDREN WITH COCHLEAR IMPLANTS

Similar to adults with CIs, children show benefits from bilateral CIs for sound source localization and speech reception in noise (Litovsky *et al.*, 2006b; Litovsky *et al.*, 2006c; Godar and Litovsky, 2010; Grieco-Calub and Litovsky, 2010). However, similar to adults with CIs, performance on sound source localization is poorer than that of NH children and benefits to speech reception in noise appear to be due to head-shadow (Litovsky *et al.*, 2006c; Grieco-Calub and Litovsky, 2010; Misurelli and Litovsky, 2012).

Children with CIs are different from adults with CIs not just in that they are less experienced and mature, but also in that many children with CIs have never experienced a period of natural hearing whereas adults with CIs are often post-lingually deaf (Litovsky *et al.*, 2012). Both animal-model research and research in children with CIs suggests that early-life auditory deprivation may negatively affect binaural sensitivity. Hancock *et al.* (2013) found that congenitally deaf white cats showed fewer neurons in the inferior colliculus that were ITD sensitive compared to cats which experienced 6 months of deafness in adulthood. Studies of children with bilateral CIs have found that many children with CIs do not demonstrate ITD sensitivity (Salloum *et al.*, 2010; Ehlers *et al.*, 2013). Ehlers *et al.* (2013) found that 1 child out of 8 with bilateral CIs was able to discriminate ITDs with direct stimulation to a mid-array electrode-pair. This is in contrast to NH infants and children who can discriminate ITDs of 75 μs and lower (Ashmead *et al.*, 1991; Van Deun *et al.*, 2009b). In contrast to ITD discrimination, children with CIs show BMLDs (Van Deun *et al.*, 2009a). Van Deun *et al.* (2009a) found BMLDs for all but 1 child out of 7 tested. BMLDs ranged from –3.6 to 15.2 dB, with a mean of 6.4 dB.

This suggests that the binaural processing required to show BMLDs does not depend on ITD sensitivity. This is notable given that research in adults with CIs has shown that there is positive relationship between ITD sensitivity and NoSπ detection thresholds (Goupell and Litovsky, 2015).

CHAPTERS 2 & 3

In Chapters 2 and 3, I examine factors that affect BMLDs (or diotic and dichotic signal detection thresholds) in children and adults with CIs. Motivation for these studies was largely derived from that finding that listeners with CIs generally show variability in performance across different stimulation sites (i.e., electrodes). In Chapter 2, I examine diotic and dichotic signal detection thresholds in children with CIs using single-electrode-pair stimulation and multi-electrode-pair stimulation to examine whether thresholds with multi-electrode-pair stimulation are better or at least as good as the single-electrode-pair thresholds at the site which performs the best when stimulated by itself. In addition, I address the question of whether having ITD sensitivity is beneficial for binaural release from masking with electrical stimulation. In Chapter 3, I examine whether monaural measures made in each ear at different stimulation sites can predict diotic and dichotic signal detection thresholds (as well as ITD discrimination) across stimulation sites in attempts to understand if variability is due to the characteristics at the auditory periphery in either ear. Throughout this dissertation, the improvement in signal detection (or speech reception) due to binaural hearing is referred to as binaural release from masking or binaural unmasking, interchangeably.

Listeners in these study all had implants manufactured by Cochlear Ltd. because of the research interface that was available. Therefore, listeners had arrays with 22 electrodes spaced .75 mm apart. Higher numbered electrodes corresponded to more apical locations and lower pitch percepts. For each active electrode, stimulation was provided with two extra-cochlear return electrodes (MP1+2). Current amplitude could be manipulated in clinical-unit steps. An increase of one clinical unit was an increase of either .157 or .176 dB of current depending on the specific internal device type.

ENDNOTES

[1]The interaural correlation, calculated with the normalized cross correlation as opposed to the normalized covariance (van de Par and Kohlrausch, 1995).

Figure 1

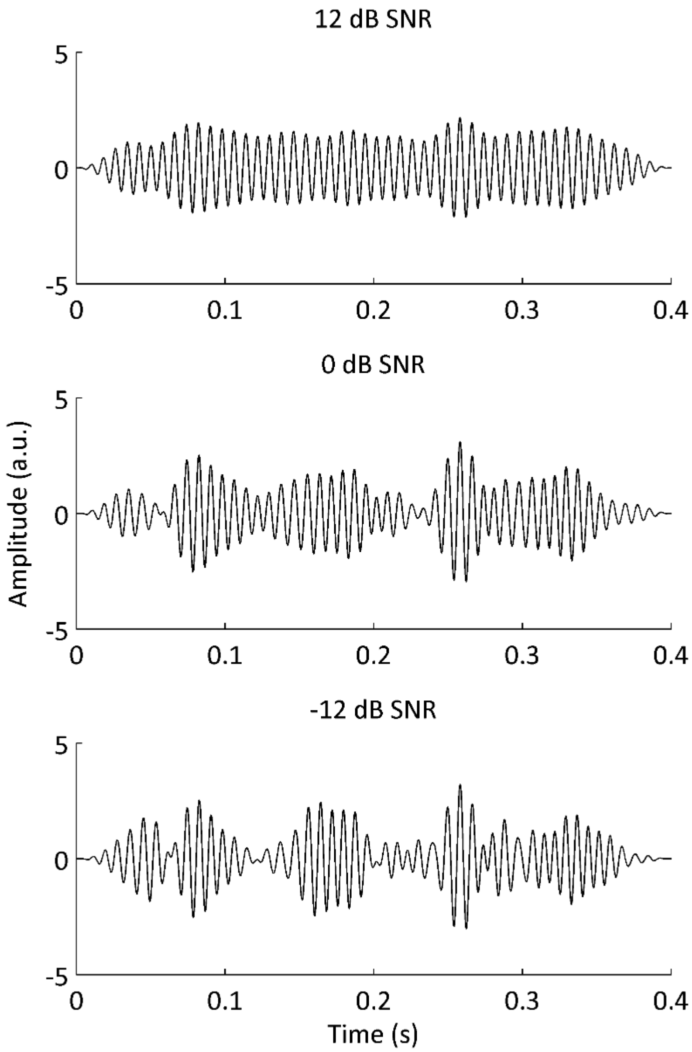

Figure 1. Amplitude (a.u.) as a function of time (s) for an NoSo stimulus made from 50-Hz-bandwidth noise centered at 125 Hz. The target was a 125 Hz tone. The upper, middle, and lower panels shows SNRs of 12, 0, and -12 dB. With this stimulus, both left and right ears receive the same information.

Figure 2

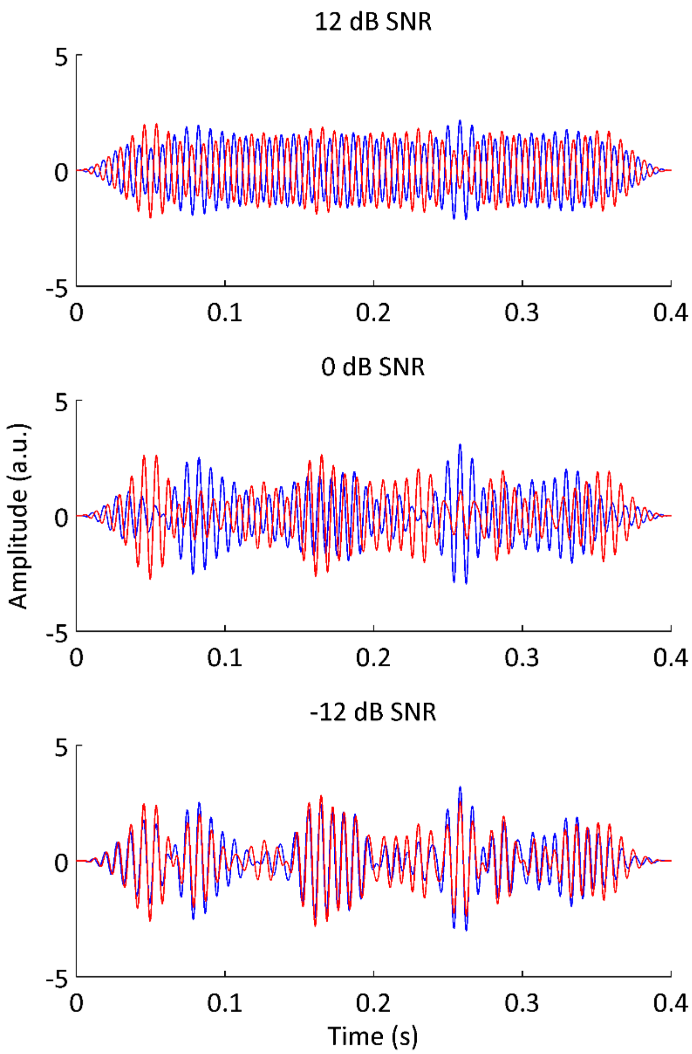

Figure 2. Amplitude (a.u.) as a function of time (s) for an NoSπ stimulus made from 50-Hz-bandwidth noise centered at 125 Hz. The target was a 125 Hz tone. The upper, middle, and lower panels shows SNRs of 12, 0, and -12 dB. The channels presented to the left and right ears are shown in blue and red, respectively.

Chapter 2

Binaural release from masking with single- and multi-electrode stimulation in children with cochlear implants

ABSTRACT

Children with bilateral cochlear implants have shown binaural unmasking for signal detection with single electrodes in both ears. Little is known about how they use binaural information with multi-electrode stimulation. Given that binaural hearing performance can vary across stimulation sites, this study sought to determine how performance with multi-electrode stimulation compared to the best performance with single-electrodes. Using direct electrical stimulation, diotic and dichotic signal detection thresholds were measured with stimulation at three-electrodes (in each ear) and at single-electrodes at each of the three stimulation sites. Results showed that performance with three-electrodes was better than the worst performance with single-electrodes and was similar to the best performance with single-electrodes, providing no evidence of interference from stimulation at sites associated with poor performance. This was similar to the pattern of performance shown by adults with normal hearing listening to a cochlear implant simulation. Children who were older showed better diotic thresholds but the effect of age was less apparent for dichotic thresholds. The magnitude of binaural unmasking was not related to whether the children had good interaural time difference sensitivity. The results provide further evidence for the potential of children with cochlear implants to benefit from binaural hearing for speech reception in noise.

INTRODUCTION

For normally hearing (NH) listeners, binaural hearing provides advantages for speech reception in noise and for signal detection (Webster, 1951; Schubert, 1956; Carhart *et al.*, 1967). Binaural release from masking can be observed when one compares listeners' performance in a diotic condition in which the target has the same interaural parameters as the noise, to a dichotic condition in which the target and noise have different interaural parameters. Listeners have better thresholds for speech reception and signal detection in the dichotic condition compared to the diotic condition. The benefits of binaural hearing arise in part due to the creation of interaural de-correlation in the combination of the target and masker which is thought to provide the listener with information about the spectral structure of the target or the presence of the signal to be detected (Akeroyd and Summerfield, 2000).

Cochlear implants (CIs) are provided to children and adults with severe-to-profound deafness in order for these individuals to have hearing. It is now common for individuals to receive bilateral CIs, which have the potential to provide the binaural hearing benefits that NH listeners receive from having two ears. Bilateral CIs have been shown to improve reception of speech in noise, but for most listeners the benefit occurs mainly from "head shadow" which results in a better signal-to-noise ratio at one of the ears (Schleich *et al.*, 2004; Litovsky *et al.*, 2006a; Litovsky *et al.*, 2009; Loizou *et al.*, 2009; Misurelli and Litovsky, 2012). Most listeners with bilateral CIs show limited to no advantages from binaural hearing for speech reception in noise (van Hoesel *et al.*, 2008; Loizou *et al.*, 2009). Since most auditory signals are transmitted to CI users via stimulation with multiple electrodes, it is important to understand how binaural hearing operates under multi-electrode conditions, especially if engineering efforts are to be made towards improving the transmission of binaural information.

Studies on binaural masking release in adults with bilateral CIs have focused on adults with post-lingual onset of deafness (Long *et al.*, 2006; Lu *et al.*, 2010; 2011). Many bilateral CI recipients are children who are pre-lingually deaf and thus, have had limited to no access to binaural cues prior to receiving their CIs. Furthermore, the binaural cues they receive through their clinically fit CIs are not preserved with fidelity. Therefore, children with CIs cannot be expected to perform as adults on measures of binaural hearing. The aim of this study was to examine binaural hearing for release from masking in children with bilateral CIs.

Children with bilateral CIs have shown binaural release from masking for signal detection with stimulation provided to single-pairs of electrodes i.e., one active electrode in each ear (Van Deun *et al.*, 2009a). This suggests that children with CIs are sensitive to interaural de-correlation of temporal envelope in dichotic listening conditions, as the stimuli they received were temporal envelope modulations of electrical pulse trains. This is noteworthy given that many children with bilateral CIs do not show sensitivity to interaural time differences (ITDs). Thus, measures of binaural unmasking make use of a binaural cue (i.e., interaural de-correlation of temporal envelope), which could potentially be used by children with CIs to obtain better speech reception in noise. While studies using single-electrode pairs demonstrate listeners' binaural sensitivity in the absence of interference from neighboring electrodes, performance with multi-electrode-pair stimulation provide information regarding more realistic stimulation modes which make use of multi-electrode stimulation.

It is unknown as to whether performance with multi-electrode stimulation will be enhanced or reduced, relative to single-electrode stimulation. On one hand, it is reasonable to assume that there may be improvement in signal detection with multi-electrode stimulation compared to single-electrode stimulation if listeners are able to integrate information across different stimulation sites (i.e., electrodes). Studies on NH listeners suggest that listeners are able to integrate information across separate critical bands. Diotic signal detection accuracy has been found to improve when the number of signals increases even when the signals occur in separate critical bands (Green, 1958; Buus *et al.*, 1986). Similarly, loudness is reported to increase as the bandwidth of a constant-energy signal is increased passed the critical band (Zwicker *et al.*, 1957). There is also evidence of integration with dichotic signal detection. Langhans and Kohlrausch (1992b) found that dichotic thresholds improved as the number of spectral components outside of any one critical band increased. The ability of listeners to integrate information across different critical bands has also been suggested as an explanation for the apparently larger binaural critical band compared to the monaural critical band for NoSπ (noise interaurally in-phasic, signal interaurally phase-inverted) detection (van der Heijden and Trahiotis, 1998; van de Par and Kohlrausch, 1999). That is, increasing noise bandwidth while maintaining overall noise energy levels does not improve signal detection thresholds until bandwidths are 2 to 4 times the monaural critical bandwidth, which can be explained with the idea that increasing bandwidths beyond the monaural critical band reduces the opportunity for listening to information in separate critical bands.

Alternatively, performance may suffer with multi-electrode stimulation for a number of reasons. First, there may be channel interactions i.e., electrodes can stimulate overlapping regions of auditory nerve fibers which may result in information presented from one electrode masking information from a different electrode (Shannon, 1983b; Abbas *et al.*, 2003; Cohen *et al.*, 2003). Lu *et al.* (2011) examined binaural release from masking with multi-electrode stimulation and found poorer binaural release when there was greater overlap in stimulated neural populations by neighboring electrodes which suggests that there was energetic masking between neighboring electrodes. Furthermore, asymmetries between ears in the regions of overlap could distort binaural information.

Additionally, performance could be reduced with multi-electrode stimulation by stimulation at sites that have poor binaural sensitivity. Binaural sensitivity is variable across different stimulation sites (van Hoesel *et al.*, 2009; Litovsky *et al.*, 2010). This across-site variability may be related to differences in neural survival and the proximity of the electrodes to the neural elements at different locations along the electrode array (Saunders *et al.*, 2002; Khan *et al.*, 2005; Fayad *et al.*, 2009). In addition, binaural sensitivity depends on stimulation being provided to interaurally place-matched neural fibers (Tollin, 2003). Across site variability may be related to the extent to which stimulated neural fibers are interaurally matched (van Hoesel and Clark, 1997). Regardless of the reason for the existence of poorer performing stimulation sites, the presence of stimulation at poorer performing sites may interfere with listeners using information from the better performing stimulation sites. Studies on NH listeners have found that the presence of a signal in one spectral region can interfere with dichotic signal detection in a separate spectral region (Bernstein, 1991b), suggesting that binaural sensitivity in one stimulation region can be influenced by stimulation in another spectral region. There is some evidence, however, that at least for adults with CIs, poorer performing sites do not interfere with performance during multi-electrode stimulation. Ihlefeld *et al.* (2014) measured ITD discrimination using amplitude modulated pulse trains in adults with bilateral CIs. Discrimination with two electrode-pairs was found to be similar to discrimination with the single electrode pair at the better performing site. This indicates that for adults with CIs, performance in ITD discrimination of amplitude modulated stimuli does not suffer from the presence of stimulation at poorer performing sites.

The aim of this study was to examine the relationship between single- and multi-electrode stimulation for both diotic and dichotic signal detection in children with CIs. Diotic and dichotic signal detection thresholds were measured using three stimulation sites individually and combined. Stimulation sites were spaced widely and presented identical stimuli to minimize the influence of energetic masking between stimulation sites. Van Deun *et al.* (2011) found that diotic and dichotic detection thresholds were similar between single- and multi-electrode stimulation when stimuli presented across adjacent electrodes were identical in the multi-electrode condition. In contrast, when different samples of noise were presented to the different electrodes, thresholds increased in the multi-electrode condition. This supports the idea that identical stimuli presented to different electrodes results in less energetic masking between stimulation sites.

Similar to Ihlefeld *et al.* (2014), we hypothesized that (1) performance would improve with multi-electrode stimulation compared to single-electrode stimulation for both diotic and dichotic signal detection if information is integrated across sites, (2) performance with multi-electrode stimulation would be worse than the best performance with single-electrode stimulation if stimulation sites produce interference, or (3) performance with multi-electrode stimulation would be similar to the best performance with single-electrode stimulation if listeners can fully use information from the best site and no integration occurs. Performance of NH adults using a CI simulation was also assessed in order to examine the pattern of performance between single- and multi-site stimulation using stimuli that were similar to what was presented to the children with CIs.

METHODS

Participants and Equipment

Participants included 11 children with bilateral CIs between the ages of 11 and 17 years. All of the children had Nucleus device types manufactured by the Cochlear Ltd. Table I shows demographic information and the electrode pairs used for the current study. Higher numbered electrodes are located more towards the apical end of the array. Stimuli were delivered from a personal computer using the Nucleus Implant Communicator and L34 processors (Cochlear Ltd.).

Eight NH adults also participated in this study. The ages of the NH participants ranged from 19 to 25. Table II shows ages and pure tone thresholds (in dB SPL) of the NH listeners. All participants had thresholds below 31 dB SPL. Stimuli were delivered by a personal computer connected to a Tucker-Davis Technologies system (System 3 with RP2.1, HB7, PA5 units) and ER-2 insert earphones (Etymotic Reseach, Inc.).

Stimuli Used with CI Listeners

Electrical stimuli were trains of biphasic pulses presented in MP1+2 (monopolar) mode. Each phase was 25 µs with an 8 µs inter-phase gap. Electrodes in the right and left ears were selected based on interaural pitch matching which was conducted for a different study. The stimuli used for pitch matching were 300 ms constant amplitude pulse trains presented at a rate of 100 pulses per second (pps), and at levels that were determined by the participants to be comfortable. The stimuli used for pitch matching were the same pulse rate and duration as the stimuli used to measure ITD discrimination. Stimuli used for loudness mapping were 400 ms pulse trains at 1000 pps matching the duration and pulse rate of the stimuli used for examining diotic and dichotic signal detection. Diotic and dichotic signal detection were measured at 1000 pps, because dichotic signal detection has been shown to be poorer at low rates (Todd *et al.*, 2014). ITD discrimination was measured at 100 pps because performance has been shown to be poorer at high rates (van Hoesel *et al.*, 2009).

The electric stimuli used to measure diotic and dichotic signal detection were based on waveforms. The waveform stimuli for examining diotic and dichotic signal detection were generated at a sampling rate of 44100 Hz. Samples of Gaussian noise were generated with a center frequency (CF) of 500 Hz, bandwidth of 50 Hz, and a duration of 400 ms created in the frequency domain. The target signal was a 300 ms 500 Hz tone which when presented, was temporally centered in the noise. Both the tone and the noise had 50 ms onset and offset ramps created with a Hann window. The tone and noise were either interaural in-phase (NoSo) or the noise was in-phase and the tone was interaurally phase-inverted (NoSπ). The signal-to-noise ratio (SNR) of the tone and noise varied between 20 dB and -32 dB in steps of 2 dB. Stimuli were pre-generated and consisted of 35 independent noise samples.

The Hilbert envelopes of the waveform stimuli were calculated and were normalized to the average amplitude. The envelopes were then resampled at a rate of 1000 Hz. The envelopes were compressed between listeners' thresholds and maximum levels (found during mapping with the multi-electrode stimulation) using the compression function used by Long *et al.* (2006) and were used to modulate the amplitude of 400 ms electrical pulse trains at 1000 pps. Pulses on left and right sides were synchronized. For one listener, CIDQ, envelopes were resampled at 1800 Hz and were used to modulate pulse trains at 1800 pps which was a rate used in the listener's clinical map. This was done because the listener had a considerably small dynamic range at 1000 pps and difficulty with the signal detection task during the familiarization with the stimuli at 1000 pps.

In the single-electrode conditions, stimuli were presented to a single bilateral pair of electrodes, located at either the basal, middle, or apical regions of the electrode arrays. In the multi-electrode conditions stimuli were presented to all three electrode pairs. When all three electrode pairs were active, electrodes on each side (left or right) were activated sequentially with 333 μs between pulse onsets from different electrodes. Identical information was presented to each of the 3 electrodes. That is, for any given stimulus interval, the same sample of noise was presented, and when the target was presented, it was presented with the same interaural phase relationship (NoSo or NoSπ) to all 3 electrodes. The three panels of Figure 1 show an example of an NoSπ stimulus at 10, 0, and -10 dB SNR from top to bottom. The left and right channels are shown in black and gray, respectively. The range of units on the ordinate was arbitrarily chosen and typically would not be the same between left and right channels.

Stimuli Used with NH Listeners

The acoustic stimuli used with the NH listeners were designed to simulate the electrical stimuli used with the CI users. The waveform stimuli were the same as those of the listeners with CIs except created at a sampling rate of 50000 Hz. The Hilbert envelopes of these stimuli were used to modulate the amplitude of trains of Gaussian-shaped pulses. Each pulse train had a CF of either 3650, 6922, or 13014 Hz to simulate implant stimulation at the apical, middle, and basal regions of the electrode array. These CFs were calculated to have a spacing of 4.5 mm along the basilar membrane according to the Greenwood function (Greenwood, 1990). This spacing is somewhat smaller than the approximate 6 mm (.75 mm between electrodes × 8 electrodes) between electrodes used for this study along the electrode array. Using the Greenwood function to calculate -3 dB bandwidths of 1.5 mm, the -3 dB bandwidths of the pulse trains were set to 788, 1467, and 2732 Hz, respectively. A pulse rate of 300 pps was used in order to control the spread of excitation while maintaining a modulation depth of >99% between pulses. Pulse trains were normalized to have equal spectral peak energy. Like the listeners with CIs, stimuli were presented to either a single CF or to all three CFs. When stimuli were presented to all three CFs, the normalized pulse trains were summed together with a 333 μs delay between the three pulse trains of different CFs. Pulse trains at a CF of 3650 Hz were presented at a level of 66 dBA. The pulse train at 6922 Hz, 13014 Hz, and the multi-site pulse train had levels of 58 dBA, 43 dBA, and 67 dBA, respectively. Interaurally uncorrelated pink noise was presented from DC to 20 kHz at 60 dB SPL to mask possible combination tones. Figure 2 shows individual pulses at each of the three CFs and the three pulses combined in the multi-site condition. Figure 3 shows the spectrum of the multi-site condition.

Procedure

Interaural pitch matching

Interaurally pitch-matched electrode pairs were sought out for stimulus presentation using methods similar to those of previous studies (Litovsky *et al.*, 2010; Litovsky *et al.*, 2012; Kan *et al.*, 2013). This was done under the assumption that electrodes in left and right ears which elicit similar pitch percepts deliver information to the same binaural processing units centrally. First, children with CIs completed a pitch-rating task in which all active, even-numbered electrodes from both sides were stimulated individually in a random order. The participants rated the perceived pitch of each stimulus by selecting a location on a visual analog scale. Responses to 10 trials were collected for each of the electrodes tested (approximately 10 electrodes in each ear, and 20 total). If this task had been performed at a previous visit to the lab, it was not repeated at the visit at which the other measures of the study were collected. Second, the results from the pitch-rating task were used to select electrodes in the two ears, for a direct interaural pitch comparison task. The pitch comparison task was completed in order to find 3 pitch-matched pairs, spaced along the electrode array at apical, middle, and basal regions. Six electrodes (typically even-numbered) on the right were chosen for each of 3 comparison electrodes on the left (typically L4, L12, and L20), thus a total of 18 comparisons. A randomly chosen electrode on the left was stimulated, followed (in random order) by one of 6 electrodes in that region on the right, at which the comparison was made. The participant indicated whether the second sound was *much higher*, *higher*, *the same*, *lower*, or *much lower* in pitch than the first sound. Responses to 20 trials were collected for each comparison.

Loudness mapping and centering

Thresholds and maximum acceptable loudness levels were measured through experimenter adjustments for each of the six electrodes (3 pitch-matched pairs) to be used for subsequent testing. Thresholds were levels that provided a consistent response from listeners on ascending tracks. Maximum acceptable loudness levels were measured by slowly and carefully increasing the stimulus level, until the participant indicated that it was the highest level still within the comfortable range. At least two measures of maximum acceptable loudness levels were obtained, and the average was the final value used for each electrode.

Using the diotic noise stimuli of the signal detection task, levels were adjusted using the following procedure. For each of the left and right sides, maximum levels for the three electrodes were lowered relative to the maximum acceptable loudness level by approximate 25 clinical units. The three electrodes were then stimulated concurrently, and maximum levels were raised in small steps until the participant indicated the loudness was at the high end of the comfortable range. Maximum levels were adjusted by changing the maximum levels for each electrode by the same number of clinical units. Each side (stimulating 3 electrodes concurrently) was then stimulated sequentially and the maximum levels of one side were further adjusted so that the two sides were as close to equal loudness as possible.

Following the loudness balancing, each individual left-right pair was stimulated on its own to check whether the auditory image was approximately centered in the listener's head. If the participant indicated that the auditory image was not approximately centered, small adjustments in the maximum levels were made to bring the image towards center.

NoSo and NoSπ signal detection

The signal detection task consisted of a 3-interval 2-alternative forced-choice task in which the target interval (NoSo or NoSπ) occurred in either the second or the third interval, and was randomly chosen on each trial. Non-target intervals consisted of diotic noise. Each interval contained a different noise sample, which was randomly selected without replacement. Inter-stimulus intervals were 300 ms. Listeners were instructed to select the interval (2^{nd} or 3^{rd}) in which the stimulus was different. Based on prior research and pilot testing, participants were given some information about how the sounds might be perceived. For the NoSo stimuli, participants were told that the stimulus that was different would likely have a percept of a sound that was "smoother" in nature. For the NoSπ stimuli, they were told that the stimulus that was different would likely have a perceived "width" or "movement" in the head. Correct answer feedback was always provided.

The SNR of the tone and noise were varied using a 2-down 1-up adaptive procedure beginning at 20 dB SNR. Initially the step size was 8 dB and changed to 4 dB after 1 turnaround and 2 dB after 3 turnarounds. The adaptive track stopped after 10 turnarounds. Stimuli were presented in blocks in which the target was either NoSo or NoSπ. This was done to reduce the number of times listeners would have to switch the cue(s) to which they were attending. Within each block, one track for each of the 4 stimulation conditions (3 single-electrode-pair conditions + 1 multi-electrode-pair condition) was presented in a new order for each block. The total number of conditions was 8 (4 stimulation conditions × 2 interaural phase conditions). Typically four complete tracks were completed for each condition.

A probe stimulus was played if the track called for a third or greater up-step uninterrupted by a down-step, and the up-step would have resulted in a stimulus at or above 12 dB SNR. The probe stimulus was at 20 dB SNR. If the listeners gave three incorrect responses at 20 dB SNR the track was stopped and was not used in the analysis. If the response to the probe was correct, the track continued, otherwise another probe was played. This procedure was modified when a participant showed consistent poor performance, such that the probe was still presented but the participant needed a greater number (i.e., 12) incorrect responses at 20 dB SNR for the program to end.

Prior to measuring signal detection thresholds, listeners were familiarized with the stimuli. Familiarization typically consisted of the signal detection task at 12 dB SNR for the NoSo condition and 0 dB SNR for the NoSπ condition. Listeners completed at least 10 trials for each condition prior to testing. The testing procedure for the NH listeners followed that of the listeners with CIs.

On a few occasions, CI participants needed a track restarted more than once in order to obtain 10 turnarounds. CIAW had difficulty completing a full track for the base pair (NoSπ), and only 3 complete tracks were obtained on that condition. CIDX had difficulty completing full tracks for the mid pair (NoSo) and the base pair (NoSo) mid-way and near the end of testing, respectively; however, 4 thresholds for this participant were obtained for all conditions. It may be the case that these conditions were actually difficult for the listeners, and therefore the thresholds may in fact be higher than what is reported. For CIDQ, the criteria for a track to end early was modified because the participant showed consistent performance near ceiling. For this participant, tracks were sometimes limited by the highest SNR available in the adaptive track (20 dB SNR) and therefore thresholds reported can be considered underestimates.

The NoSo and NoSπ thresholds were fit to linear mixed-effects model which had random intercepts for participants and stimulation conditions. F-tests were conducted by comparing models with and without the fixed effect of interest. Binaural masking level differences (BMLDs) were calculated by subtracting NoSπ thresholds from NoSo thresholds.

Interaural time differences

In addition to measuring NoSo and NoSπ thresholds, interaural time difference (ITD) just-noticeable-differences (JNDs) were measured in the listeners with CIs for a different study but on the same visit on which the BMLD measures were made. Stimuli consisted of constant amplitude pulse trains of 100 pps presented at comfortable levels, and were presented using a method of constant stimuli. The ITDs tested were adjusted based on the listener's sensitivity in order to create psychometric functions with at least four points, each point consisting of at least 40 trials. ITDs greater than 1600 µs were not tested. Listeners heard 2 intervals and responded by indicating the location (left vs. right) of the second sound relative to the first sound. The ITD of the second sound was equal in magnitude but opposite in direction to the first sound. Psychometric functions were calculated based on the data collected (Wichmann and Hill, 2001a; b). ITD JNDs were the estimated ITDs at which listeners' responses were 70.7% correct based on the psychometric function. If ITD sensitivity could not be determined or was estimated to be 1600 µs it was classified as ≥1600 µs.

Pure tone thresholds

For the NH listeners, tone detection thresholds were measured at the CFs of the stimuli (3650, 6922, and 13014 Hz) using a 2-down, 1-up adaptive procedure, which ended after 10 reversals. The step size of the adaptive procedure changed from 3 dB to 1.5 dB after the first reversal and to .5 dB after the second reversal. Typically a single track was collected per CF for each ear. Thresholds were calculated by averaging the last 6 reversals of each track. Thresholds are shown in Table II.

RESULTS

NoSo and NoSπ thresholds

CI listeners

Figure 4 shows NoSo and NoSπ thresholds for each of the children with CIs for each of the single-electrode conditions and the multi-electrode condition. NoSπ thresholds were lower than NoSo thresholds [$F_{1,298}$ = 123.91, P < .0001]. The mean NoSo threshold was 2.49 dB SNR

(S.D. = 7.44 dB) and the mean NoSπ was -3.82 dB SNR (S.D. 4.7 dB), with a mean BMLD of 6.31 dB (S.D. = 6.09 dB). If the data are considered without the results of participant CIDQ, who had considerable difficulty with the task, then the mean NoSo threshold was 1.65 (S.D. = 3.99 dB) and the mean NoSπ threshold was -5.5 dB SNR (S.D. = 5.23) with a mean BMLD of 7.15 dB (S.D. = 5.73). The effect of track order was examined to determine whether the results were affected by learning or fatigue. The effect of order was not significant [$F_{1,298} = .26$, $P = .609$] nor was the phase × order interaction [$F_{1,297} = .046$, $P = .83$]. The effect of stimulation site (apex, mid, base) was not significant [$F_{2,19} = .31$, $P = .73$] nor was the phase × stimulation site interaction [$F_{2,17} = .82$, $P = .45$].

Performance was compared between single- and multi-electrode stimulation conditions. Comparisons were made between multi-electrode stimulation thresholds and the best, 2nd best, and worst single-electrode thresholds. For both NoSo and NoSπ, thresholds in the multi-electrode condition were better than the worst single-electrode thresholds [NoSo: $t_{26} = 4.086$, $P = .00037$; NoSπ: $t_{26} = 5.097$, $P < .0001$] and the second best single-electrode thresholds [NoSo: $t_{26} = 2.47$, $P = .020$; NoSπ: $t_{26} = 2.12$, $P = .044$]. The multi- electrode thresholds were not different from the best single-electrode thresholds [NoSo: $t_{26} = 1.26$, $P = .56$; NoSπ: $t_{26} = .59$, $P = .56$]. Figure 5 shows the multi-electrode thresholds as a function of the worst single-electrode threshold for each child with NoSo and NoSπ conditions shown in black and gray, respectively. All points fall at or below the identity line, indicating that, in general, the multi-electrode thresholds were better than at least one of the single-electrode thresholds.

Figure 6 shows the multi-electrode thresholds as a function of the best single-electrode threshold with points falling on both sides of the identity line. For the NoSo condition, there were 6 children whose thresholds were nominally better in the multi-electrode condition than in the best single-electrode condition. There were 3 children whose multi-electrode thresholds were poorer than the best single-electrode thresholds, but this was only outstanding for 1 child. The other 2 children showed multi-electrode thresholds that were quite similar to their best single-electrode thresholds. For the NoSπ condition, all children had performance in the multi-electrode condition that was similar to or better than the best single-electrode threshold except one child. Four children had NoSπ thresholds in the multi-electrode condition that were nominally better than their best single-electrode threshold, and six children had multi-electrode thresholds that were similar to their best single-electrode threshold.

NH listeners

Figure 7 shows the NoSo and NoSπ thresholds of each of the NH adults as a function of stimulation site. NoSπ thresholds were lower than NoSo thresholds [$F_{1,220}$ = 141.39, P < .0001]. The mean NoSo threshold was 3.11 dB SNR (S.D. = 2.16 dB) and the mean NoSπ was -6.00 dB SNR (S.D. = 8.19 dB), with a mean BMLD of 9.11 dB (S.D. = 7.21). For both NoSo and NoSπ, multi-site thresholds were better than the worst single-site thresholds [NoSo: t_{17} = 2.97, P = .008; NoSπ: t_{17} = 5.67, P < .0001]. For NoSπ but not NoSo, the multi-site thresholds were better than the 2nd best single-site threshold [NoSo: t_{17} = 1.39, P = .18; NoSπ: t_{17} = 3.23, P = .0049]. The multi-site thresholds were not significantly different from the best single-site thresholds [NoSo: t_{17} = .49, P = .63; NoSπ: t_{17} = 1.11, P = .28].

The effect of place (CF) was significant [$F_{2,13}$ = 6.93, P = .0089]. The phase × place interaction was not significant [$F_{2,11}$ = 2.61, P = .12]. Post-hoc pairwise contrasts using a Holm correction (for 6 contrasts) showed that the middle CF produced higher NoSπ thresholds than the apical CF [t_{11} = 4.047, P = .011]. There was no significant difference in NoSπ thresholds between the apical and basal CFs [t_{11} = 1.64, P = .52] or the middle and basal CFs [t_{11} = 2.4, P = .18]. There were no significant differences between places for the NoSo condition or for BMLDs.

Effect of ITD Sensitivity

Figure 8 shows ITD JNDs as a function of place of stimulation for the children with CIs. ITD discrimination was not measured for CIAW or CIAG and was only tested on the mid electrode pair for CIDX due to time limitations. Five children (out of 9 tested) showed ITD sensitivity (<1600 μs) for at least one electrode pair.

Figures 9 and 10 show the relationship between ITD sensitivity and BMLDs and ITD sensitivity and NoSπ thresholds, respectively, for each stimulation site of each child with CIs. We were interested in whether binaural unmasking was greater in listeners with ITD sensitivity given that ITD processing may aid in binaural unmasking if interaural differences in the temporal modulations of the NoSπ stimuli can be processed as temporal envelope ITDs. If ITD processing is involved with binaural unmasking, we would expect less unmasking from children who are insensitive to ITDs.

It can be seen in Figure 9 that the BMLDs of the children with ITD sensitivity are within the range of those of the children without ITD sensitivity. For the children without ITD sensitivity, BMLDs ranged from -8.33 to 18.91 dB or 1.08 to 18.91 dB without CIDQ. The BMLDs of the children with ITD sensitivity ranged from 3.00 to 10.5 dB. Similarly, it can be seen in Figure 10 that the NoSπ thresholds of the children with ITD sensitivity are within the range of those of the children without ITD sensitivity. For the children without ITD sensitivity, NoSπ thresholds ranged from -12.58 to 17.33 dB SNR or -12.58 to 3.25 dB SNR without CIDQ. Similarly, the NoSπ thresholds of the children with ITD sensitivity ranged from -12.83 to 6.33 dB SNR.

Effect of Age

One of the questions addressed in this study was the effect of age of children with CIs on diotic and dichotic signal detection thresholds. Studies of NH children have found that NoSo thresholds, NoSπ thresholds, and BMLDs and binaural release from masking for speech reception improve with age (Hall and Grose, 1990; Summerfield *et al.*, 1994). Figure 11 shows NoSo and NoSπ thresholds averaged across conditions for each child as a function of age. Thresholds were lower for children with higher ages [$F_{1,9}$ = 7.14, P = .026]. The interaction between phase and age was not significant [$F_{1,8}$ = 2.8203, P = .12]. However, for the NoSo data alone, the effect of age was significant [$F_{1,10}$ = 21.23, P < .00097], but age was not significant for the NoSπ data alone [$F_{1,10}$ = 2.061, P = .18].

DISCUSSION

Research on children with bilateral CIs has shown that binaural release from masking can be observed when a limited number of electrode pairs in the cochlear array is stimulated (Van Deun *et al.*, 2009a; Van Deun *et al.*, 2011). However, everyday listening with CIs involves multi-electrode stimulation in order for spectral information of signals to be transmitted to the listener. Listeners with bilateral CIs have demonstrated limited benefit from binaural processing for speech reception in noise (van Hoesel *et al.*, 2008; Loizou *et al.*, 2009; Misurelli and Litovsky, 2012), indicating a need to understand binaural hearing under multi-electrode stimulation in listeners with CIs. In this study, we were interested in examining binaural unmasking of children with bilateral CIs with single- and multi-electrode stimulation. Children with bilateral CIs are a unique population in that many of them have only received degraded auditory input throughout their lives.

In this study, we examined diotic (NoSo) and dichotic (NoSπ) signal detection with three-electrode-pair stimulation compared to signal detection with single-electrode-pair stimulation at each of the three stimulation sites individually. Prior research on binaural sensitivity of listeners with CIs has indicated that there is variability in performance across stimulation sites (van Hoesel *et al.*, 2009; Litovsky *et al.*, 2010), and it is unknown whether stimulation of poorer performing sites interferes with processing of information from better performing sites. One of three patterns of performance was expected: (1) information would be integrated across stimulation sites producing better performance in the multi-electrode condition than in any single-electrode condition; (2) the presence of poorer performing sites would interfere with listeners' use of the better performing sites, producing performance that was worse in the multi-electrode condition compared to the best single-electrode performance; (3) there would be no integration or interference and listeners would show performance in the multi-electrode condition that was similar to their best single-electrode performance.

NoSo and NoSpi Detection Thresholds

Consistent with previous research, most of the children with CIs showed BMLDs i.e., better signal detection thresholds in dichotic conditions than in diotic conditions (Van Deun *et al.*, 2009a). Van Deun *et al.* (2009a) found an average BMLD of 6.4 dB with single-electrode stimulation with a group of 7- to 15-year-olds with bilateral CIs, which is quite similar to the BMLD of 6.3 dB found in this study. The stimuli in this study differed from those used in most previous studies on binaural unmasking of listeners with CIs in that the stimuli were not "transposed" (van de Par and Kohlrausch, 1997; Long *et al.*, 2006; Van Deun *et al.*, 2009a; Lu *et al.*, 2010) i.e., the half-wave-rectified temporal fine structure of the unprocessed stimuli did not appear as temporal envelope modulations of the electric pulse trains. The interaural differences that occurred in the dichotic stimuli were due to the effect of the target signal on the temporal envelope of the noise, which depends on the phase of the signal. This type of interaural information would be presented with most envelope detection strategies, not just those involving half-wave-rectification. Therefore the use of non-transposed stimuli is quite relevant for understanding how to improve realistic stimulation. Presumably, sensitivity to the interaural differences present in the dichotic stimuli in this study would depend largely on the processing of interaural level differences. This idea is supported by the finding that children without ITD sensitivity showed BMLDs. In fact, they showed BMLDs and dichotic detection thresholds that were similar to the children who were sensitive to ITDs as shown in Figures 9 and 10, providing no evidence that ITD sensitivity aids in binaural unmasking when only the temporal envelope of the unprocessed stimulus is presented.

Most of the listeners' thresholds varied between sites of stimulation to some extent. There was no systematic effect of place of stimulation for the children with CIs, which is consistent with studies in adults which have not found an effect of stimulation site on ITD discrimination (van Hoesel *et al.*, 2009; Litovsky *et al.*, 2010). The lack of a place effect may in part be due to the fact that with electrical stimulation, identical stimuli can be presented to different places of stimulation unlike with acoustic hearing. Contrary to the lack of a place effect in listeners with CIs, NH listeners have shown an effect of CF for high frequency interaural correlation discrimination. Goupell and Litovsky (2014) found that interaural correlation discrimination was worse at a CF of 8 kHz compared to a CF of 4 kHz. Poorer performance at higher CFs (4, 8, and 12 kHz were examined) has also been observed with envelope ITD discrimination in NH listeners (Bernstein and Trahiotis, 1994). In the current study, NH listeners showed poorer NoSπ thresholds at the 6922 Hz CF compared to the 3650 Hz CF. Given that previous studies have found poorer binaural sensitivity at higher CFs and that the stimuli were quieter at higher CFs in this study, it was not surprising that the 6922 Hz thresholds were higher than the 3650 Hz thresholds. However, the reason that this effect was apparent with 6922 Hz but not with 13014 Hz was surprising and requires further investigation. The observation that listeners with CIs do not show an effect of stimulation site could be interpreted as an indication that poorer performance at higher CF in NH listeners is not due to central binaural auditory processing. However, due to uncertainty about which regions of the auditory nerve are stimulated by any particular electrode and variability in neural survival and the characteristics of the electrode-neural interface, this conclusion cannot be made with any certainty.

For the children with CIs, performance with single-electrode-pair stimulation in this study may have been poorer than what it would have been had higher levels of stimulation been used. The maximum level for the single-electrode conditions was typically lower than what it could have been, because for each electrode the same levels were used for both the single-electrode and the multi-electrode conditions. Since multi-electrode simulation is louder than single-electrode stimulation lower levels were necessary. Studies on NH listeners have shown that sensitivity to envelope ITDs and ILDs decreases at lower levels; however, the effect is greater for ITDs than ILDs (Dietz *et al.*, 2013). Furthermore, it is possible that had we conducted a loudness balancing procedure across the different stimulation sites prior to the experimental task, there would have been less across-site variability. However, loudness balancing between stimulation sites that produce different pitch and quality percepts cannot be assumed to be straightforward, and it is uncertain by what criteria the children would have adjusted the level of the stimuli.

Thresholds were found to be related to age, at least for the NoSo stimuli. Children in this study ranged in age from 11 years to 17 years. From this sample of children, it appeared that NoSo thresholds decreased as a function of age from 11 to 14 years and thereafter almost plateaued. This is in contrast to the findings of Hall and Grose (1990) who found that NoSo (and NoSπ) thresholds of NH children reached the adult range by 6 to 7 years of age. Hall and Grose (1990) held the level of their noise constant and varied the level of the signal. In the current study, we held the overall level of the combination of noise and signal constant and varied the SNR which may have led to a more difficult task. Examination of the performance of NH children using a CI simulation (as was used in this study with NH adults) would help to determine whether there is a relationship between NoSo signal detection and age for 11- to 17-year-olds with NH. It is possible that children with CIs show an effect of age at older ages than NH children. This could be due to the fact that the auditory input they receive is more degraded than that of children. It is reasonable to think that degraded auditory input could lead to slower auditory development. NoSπ thresholds showed a much less consistent pattern with age. In fact, some of the younger children showed the best NoSπ thresholds which suggests that the binaural cues were more salient to some of the children, such that advantages of being older were less apparent for the NoSπ signal detection. It should be noted, however, that some children with CIs have shown improvements over time with performance on localization accuracy and acuity which involve binaural hearing suggesting that experience may help to improve binaural hearing performance (Litovsky *et al.*, 2006b; Grieco-Calub and Litovsky, 2010).

Single- vs. Multi-site Thresholds

Children with CIs showed BMLDs in the multi-site condition in addition to the single-site conditions. Van Deun *et al.* (2011) also found that three out of three children with CIs showed BMLDs with stimuli presented from three adjacent electrodes. However, diotic and dichotic signal detection thresholds were higher and BMLDs were smaller in the multi-electrode condition compared to the single-electrode condition likely due to energetic masking between electrodes. In contrast, in this study the three stimulation sites were spaced widely along the array and presented identical stimuli. Presenting identical stimuli was expected to reduce energetic masking because any neural populations which were stimulated by more than one electrode would receive the same information from each electrode. Furthermore, wide spacing of electrodes was expected to reduce the number of neural fibers which were stimulated by more than one electrode as was found by Lu *et al.* (2011).

In this study, for both diotic and dichotic thresholds, the children with CIs as a group showed performance in the multi-site condition that was better than their worst single-site thresholds and was similar to their best single-site threshold. This finding is similar to those of Ihlefeld *et al.* (2014) who examined envelope ITD discrimination in adults with CIs. There were instances in which multi-site thresholds were notably worse than any of the three single-site thresholds; however, this was relatively rare. This suggests that listeners did not generally have difficulty perceiving information from the better performing sites in the presence of stimulation from the poorer performing sites suggesting that there was no interference from stimulation at poorer performing sites. A possible alternative explanation is that there was a combination of integration across sites which improved performance and interference between sites which degraded performance. The combination of the two effects could have led to the appearance of neither integration nor interference. Either way, the results suggest that with the stimulation configuration used in this study, little benefit in signal detection would be derived from removing the stimulation sites that produce poorer thresholds. The removal of stimulation sites with poor performance has been suggested as a means of improving speech reception of listeners with CIs (Garadat *et al.*, 2012). Further investigation could determine if no interference is found with greater numbers of electrodes and other types of binaural hearing measures. Bernstein (1991b) found interference for NH listeners on a dichotic signal detection task with the presence of both diotic and dichotic interferers. However, interference was larger with dichotic interferers. We might expect that if any stimulation site or group of stimulation sites produces interaurally de-correlated internal representations due asymmetries between the ears, interference might be more likely.

Performance in the multi-electrode condition was not found to be better than performance with the best single electrode-pair. This was also the finding with the NH adults suggesting perhaps that the stimuli used in this study did not lend themselves well to being integrated. It has been observed that with stimuli which are variable from trial to trial (as they were in this study, since the sample of noise changed from trial to trial) listeners do not show integration of stimuli presented between the two ears (Langhans and Kohlrausch, 1992a). Therefore, it may also be the case that integration is not observable for stimuli presented at different stimulation sites when the stimuli vary from trial to trial.

Despite the general pattern of thresholds in the multi-electrode condition being similar to thresholds in the best single-electrode condition, there were a number of children whose best performance was in the multi-electrode conditions. However, this pattern of results did not typically occur for both the diotic and dichotic conditions, as can be seen in Figure 4. For example, CIBK's performance was best in the multi-electrode condition only for NoSo, which eliminated the BMLD in the multi-electrode condition. CIAW's performance was best in the multi-electrode condition, but only for NoSπ; thus, a BMLD was only observed in the multi-electrode condition. CIEV was the only child that demonstrated a BMLD in the multi-electrode condition and that had better multi-electrode performance for both NoSo and NoSπ. The variable performance of the listeners with CIs suggests there may be certain circumstances in which integration of information across stimulation sites is more likely, and the reason for this requires further investigation

For NH listeners, we can expect that in some situations integration of information across different frequency regions occurs due to the presence of combination tones which can increase in level when more components are added to the stimuli (Wiegrebe and Patterson, 1999). Combination tones are thought to be due to cochlear mechanisms which would be absent for listeners with CIs for whom cochlear processing is bypassed (Shera and Guinan, 1999). Therefore, for listeners with CIs we would expect that integration would be due solely to central mechanisms. It is still a question as to whether listeners with CIs show integration for other types of stimuli. If listeners with CIs can receive a benefit from having redundant information presented at multiple stimulation sites, this might present a way to improve binaural hearing performance of listeners with CIs.

Summary

The results provide further support that children with bilateral CIs are able to show binaural release from masking for signal detection. This was demonstrated despite some children being insensitive to ITDs. In both diotic and dichotic conditions, stimulation with three-electrode pairs presenting identical stimuli did not result in evidence of interference suggesting that children with CIs are able to use information from better stimulation sites in the presence of information from the poorer performing sites.

ACKNOWLEDGMENTS

This work was supported by NIH Grant R01DC008365 to R.Y.L and NIH Grant F31DC013238 to A.E.T. Support was also provided by NIH P30HD03352 to the Waisman Center. We would like to thank Shelly Godar for help with recruitment and testing of the children with cochlear implants, and Matt Goupell for input on creating the stimuli for the normally hearing listeners. Thanks also go to Brianna Vandyke for help with testing the normally hearing participants. We would also like to thank the children with CIs and their parents for participating in the study.

Table I. Demographics of the participants with cochlear implants.

Participant	Age (yrs)	Hearing History	Etiology	Age at 1^{st} activation (yrs)	Age at 2^{nd} activation (yrs)	Internal device (L,R)	Base pair	Mid pair	Apical pair
CIDX	11.6	Pro.[a] HL[b] ID at birth	Connexin 26	1.43	2.59	CI24R(CS), CI24R(CS)	L4,R4	L12,R12	L20,R18
CIDQ	12.2	ID at birth	Connexin 26	.82	4.34	CI24RE, CI24R(CS)	L4,R4	L12,R12	L20,R20
CIDJ	13.1	ID at 12 mo	Hereditary	1.62	5.04	CI24RE, CI24R(CS)	L6,R6	L12,R12	L20,R18
CIEV	13.2	Progressed from sev.[c] to pro. by 2-3 yrs	Hereditary	2.67	10.95	CI24RE, CI24R(CA)	L4,R6	L12,R14	L20,R20
CIAG	13.3	Progressed from mod[d]-sev. (L) and sev-pro. (R) to pro. by 15 mo	Connexin 26	1.72	3.12	CI24R(CS), CI24R(CS)	L4,R4	L12,R12	L20,R20
CIAW	13.6	Pro. HL ID at 3 mo	Congenital CMV	1.21	5.46	CI24RE, CI24R(CS)	L4,R4	L12,R8	L20,R22
CIBO	14.2	Progressed from mod. to pro. HL by 28 mo	EVAS/ Pendred syndrome	2.83	3.90	CI24R(CS), CI24R(CS)	L4,R4	L12,R12	L20,R18
CIAP	14.7	Progressed from mild to sev-pro. by 3 yrs	Unknown	3.47	5.10	CI24R(CA), CI24R(CA)	L4,R4	L12,R10	L20,R16
CIBK	15.2	ID at 17 mo	Connexin 26	2.14	7.11	CI24RE, CI24R(CS)	L4,R4	L12,R12	L20,R18
CIEU	16.2	Progressed from sev. to pro. by age 4 (R) and 8 (L) yrs	Hereditary	4.28	10.45	CI24RE, CI24R(CS)	L4,R4	L12,R12	L18,R18
CIAQ	17.5	Pro. HL ID at 13 mo	Connexin 26	4.00	8.21	CI24R, CI24R	L4,R4	L12,R13	L20,R19

[a]Profound, [b]Hearing loss, [c]Severe, [d]Moderate

Table II. Age and pure tone thresholds in dB SPL of the normally hearing listeners

Participant	Age (years)	3650 Hz (L, R)	6922 Hz (L, R)	13014 Hz (L, R)
TDP	19	15.8, 20.5	16.6, 2.7	10.0, 9.7
TDY	19	15.2, 22.8	2.5, 7.1	6.8, 10.6
TEA	19	2.6, 1.1	-0.3, -0.9	3.2, 7.4
TEB	19	10.8, 5.9	6.1, 6.6	3.4, 6.9
TDZ	22	12.4, 8.2	-6.3, 5.7	11.7, 30.8
TAF	23	5.5, 17.3	7.3, 8.2	6.7, 10.5
TAW	23	14.7, 15.2	2.2, 9.2	8.5, 11.3
TDQ	25	19.6, 28.0	2.1, 10.8	0.9, 2.3

Figure 1

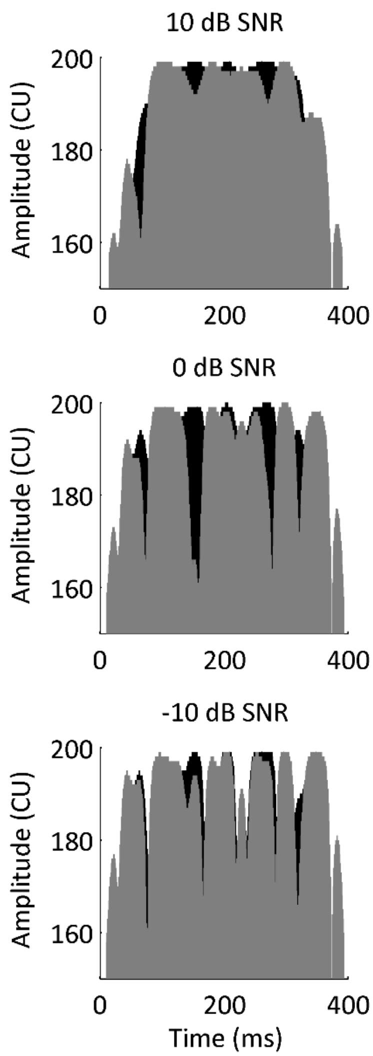

Figure 1. An example of an electrical NoSπ stimulus at 10 dB SNR (top), 0 dB SNR (middle), and -10 dB SNR (bottom). The left and right channels are shown in black and gray, respectively. The current units of the ordinate are in clinical units and the range of the scale was arbitrarily chosen.

Figure2

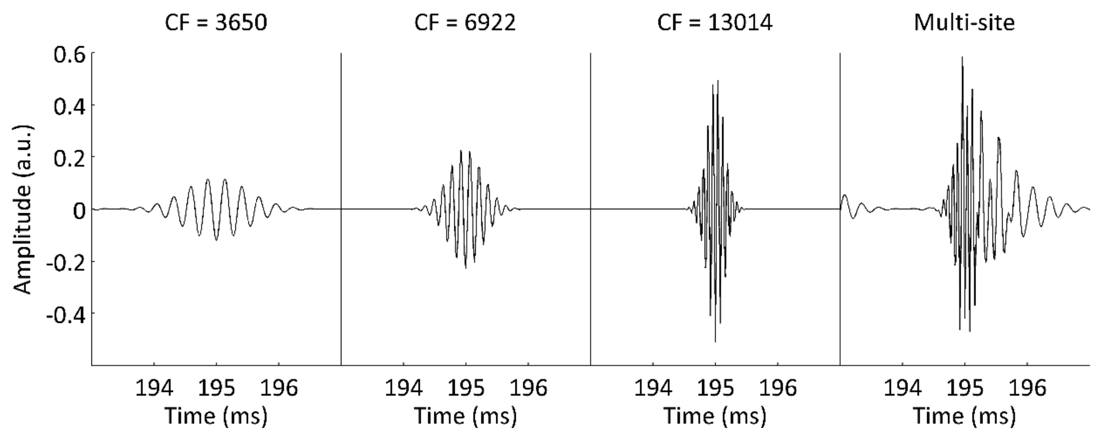

Figure 2. Individual Gaussian-shaped pulses of the CI simulation. The first three panels show pulses centered at 3650, 6922, and 13014 Hz, respectively. The fourth panel shows the pulses of different center frequencies summed together with a 333 μs delay for the multi-site condition.

Figure 3

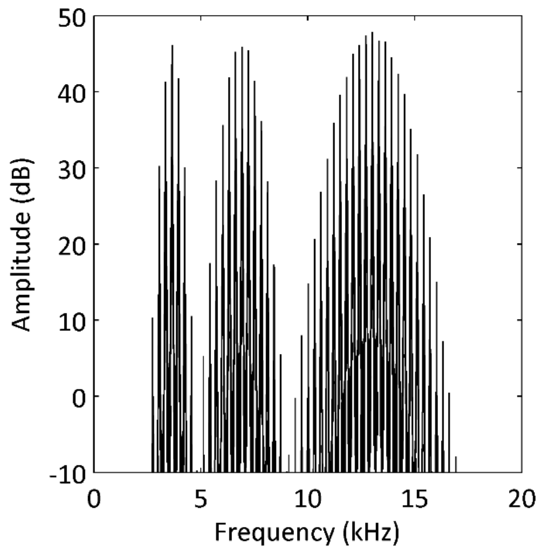

Figure 3. The spectrum of diotic noise (No) in the multi-site condition of the CI simulation.

Figure 4

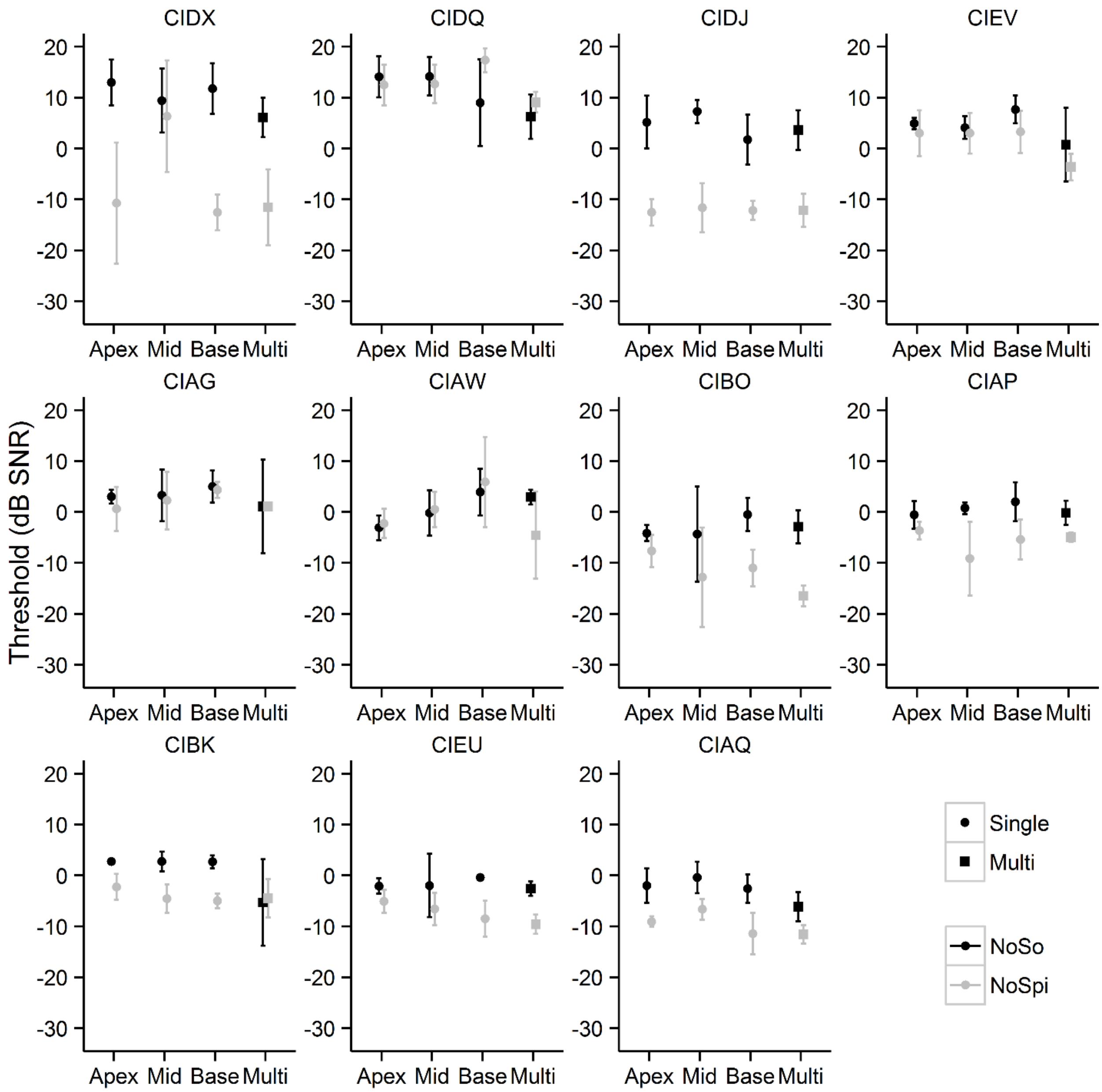

Figure 4. NoSo (black) and NoSπ (gray) thresholds (dB SNR) of the children with CIs for each stimulation condition. Each panel shows the data of an individual child. Single-electrode-pair

stimulation is shown by circles and multi-electrode-pair stimulation is shown by squares. Error bars represent standard deviations.

Figure 5

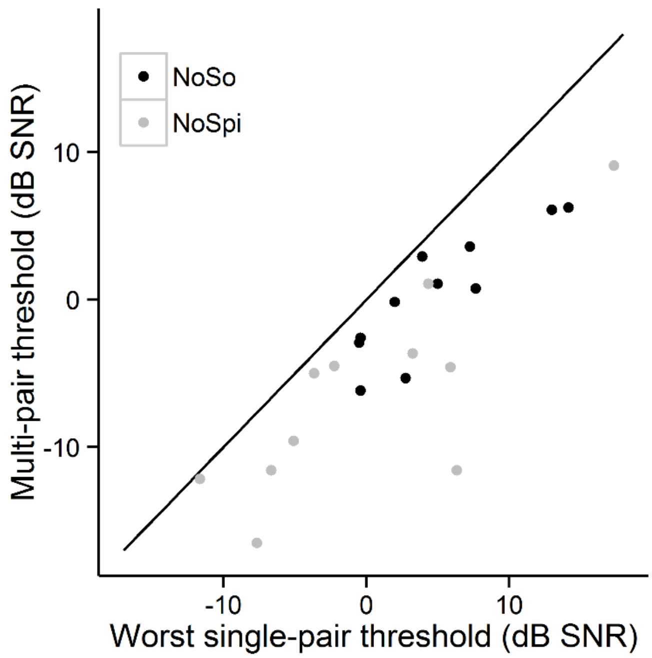

Figure 5. Multi-electrode-pair NoSo and NoSπ thresholds (dB SNR) as a function of the worst single-electrode-pair threshold (dB SNR) of the children with CIs. There are two points per child: one for the NoSo condition (black) and the other for the NoSπ condition (gray).

Figure 6

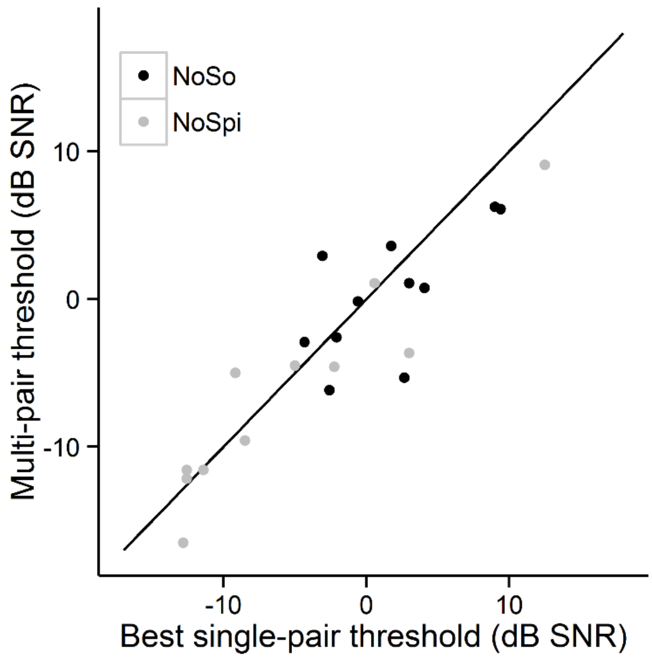

Figure 6. Multi-electrode-pair NoSo and NoSπ thresholds (dB SNR) as a function of the best single-electrode-pair threshold (dB SNR) of the children with CIs. There are two points per child: one for the NoSo condition (black) and the other for the NoSπ condition (gray).

Figure 7

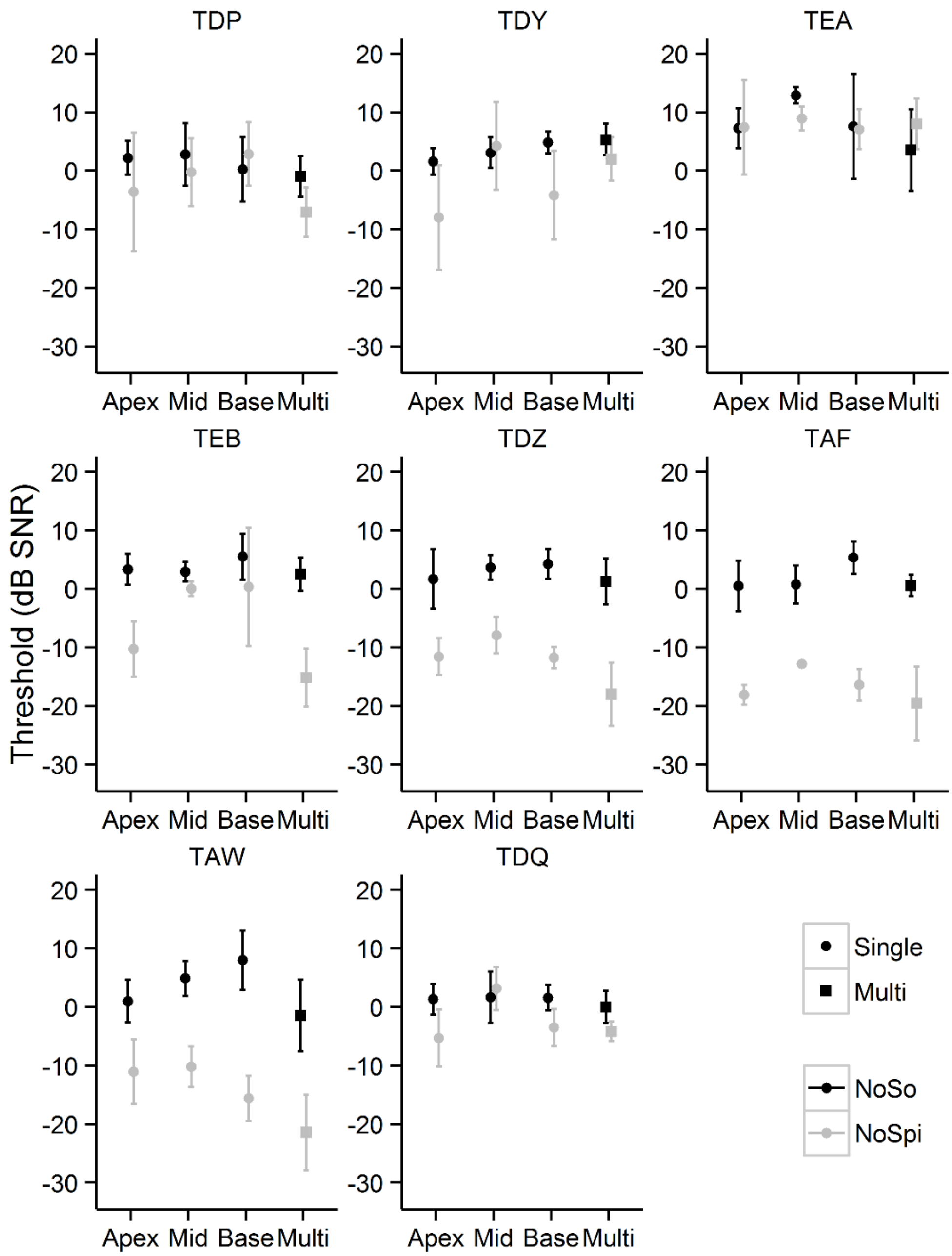

Figure 7. NoSo (black) and NoSπ (gray) thresholds (dB SNR) of the NH adults for each stimulation condition. Each panel shows data from an individual listener. Single-site stimulation

is shown by circles and multi-site stimulation is shown by squares. Error bars represent standard deviations.

Figure 8

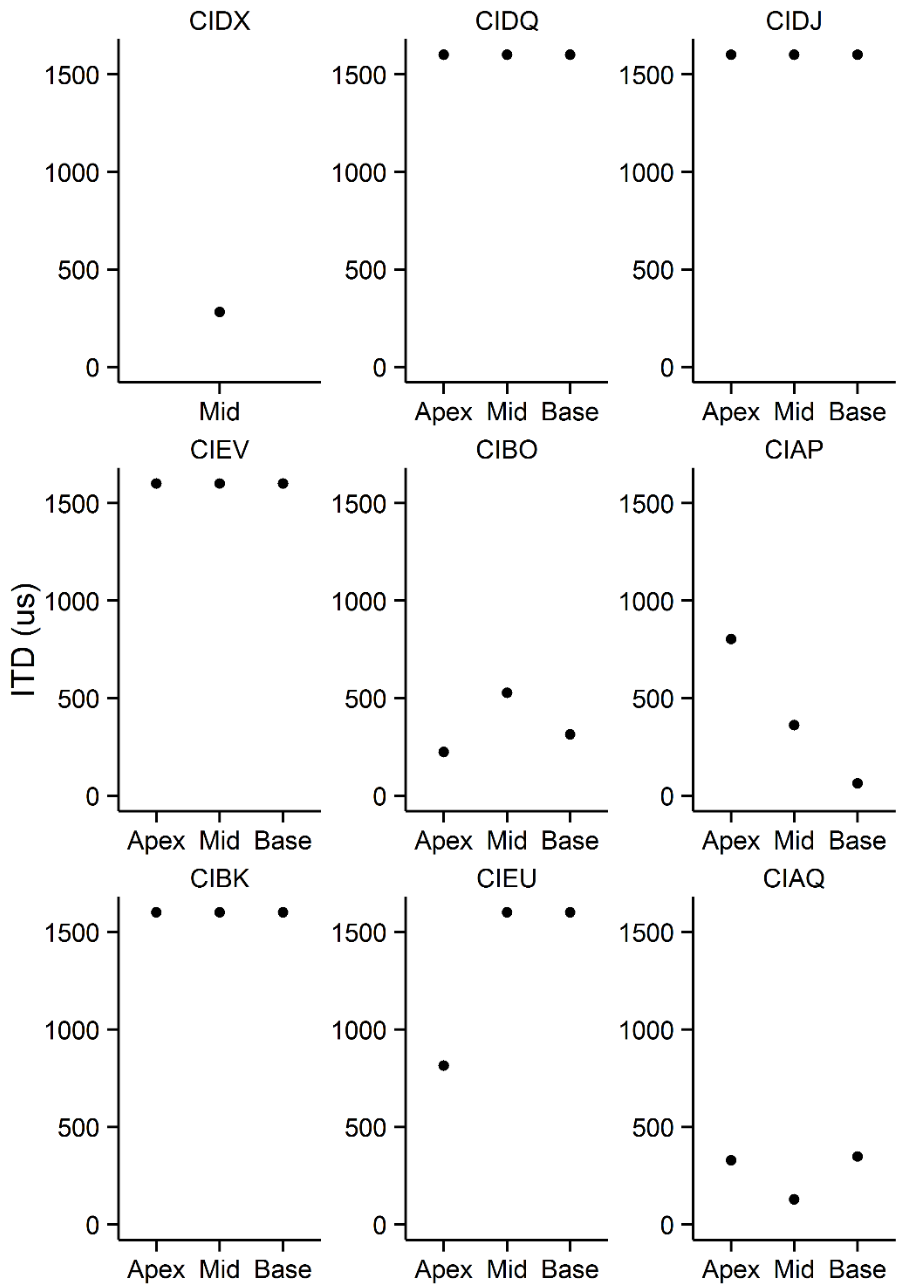
CIDX
CIDQ
CIDJ
CIEV
CIBO
CIAP
CIBK
CIEU
CIAQ
ITD (us)
1500
1000
500
0
Apex
Mid
Base

Figure 8. ITD JNDs (μs) of the children with CIs. Each panel shows data from an individual child. Points at 1600 μs indicate that the ITD JND was not measurable or was estimated to be 1600 μs. (ITDs > 1600 μs were not measured.)

Figure 9

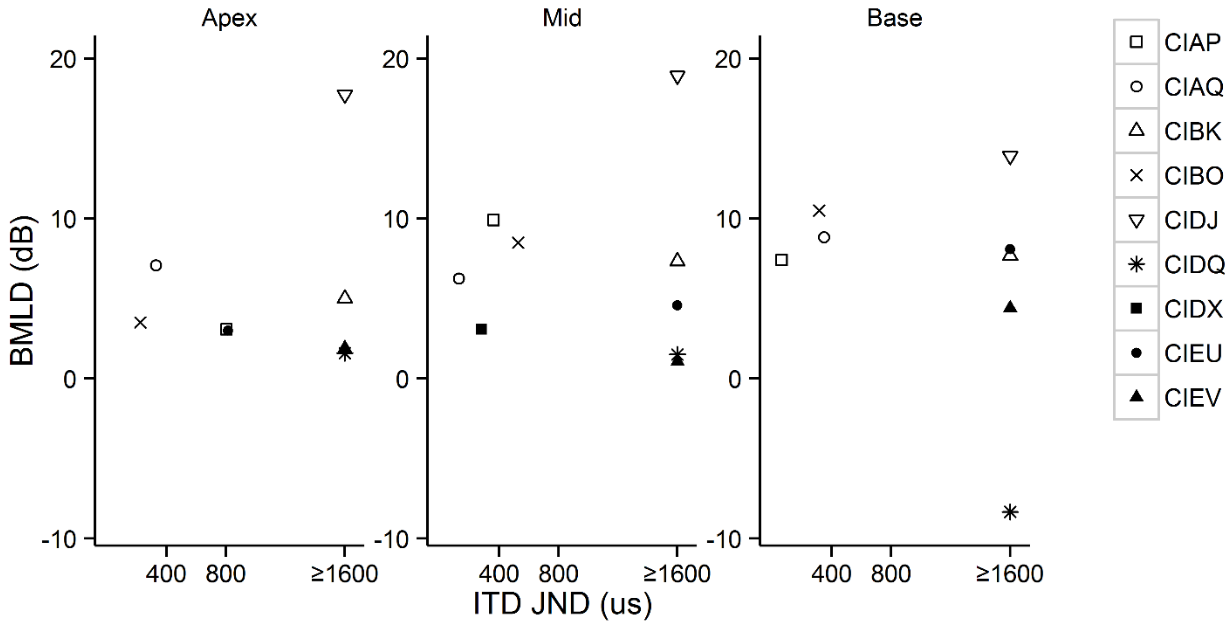

Figure 9. BMLDs (dB) as a function of ITD JNDs (μs) of the children with CIs. Panels from left to right show performance at the apical, middle, and basal sites of stimulation, respectively. Each child is represented with a unique symbol.

Figure 10

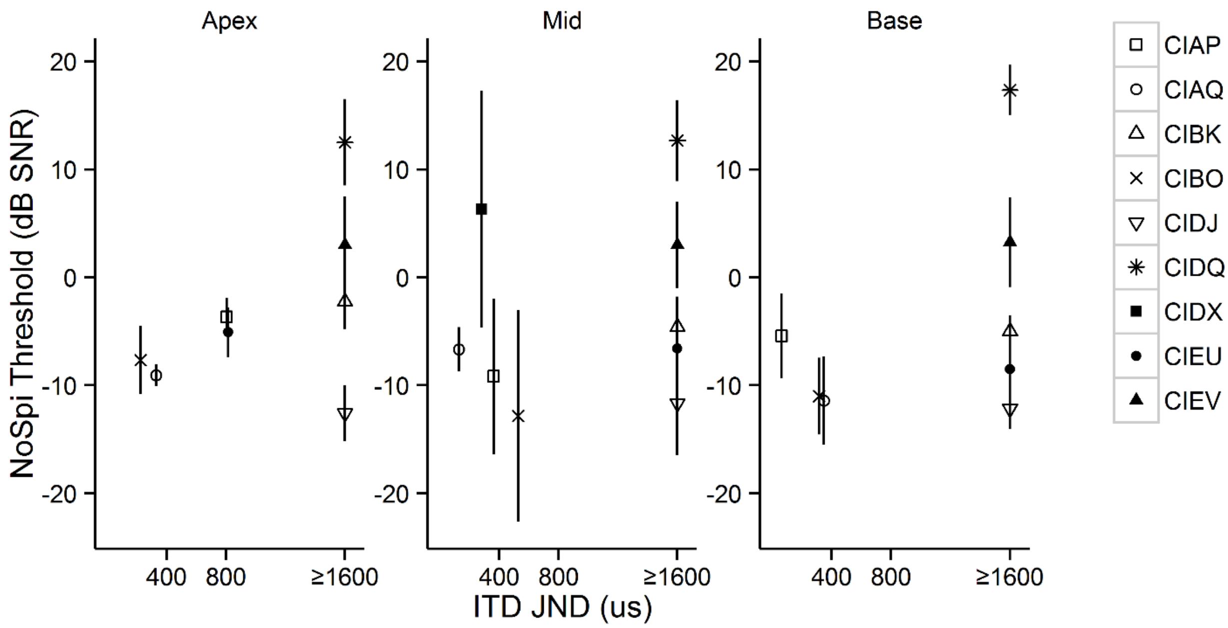

Figure 10. NoSπ (dB SNR) thresholds as a function of ITD JNDs (μs) of the children with CIs. Panels from left to right show performance at the apical, mid, and basal sites of stimulation, respectively. Each child is represented with a unique symbol. Error bars represent standard deviations.

Figure 11

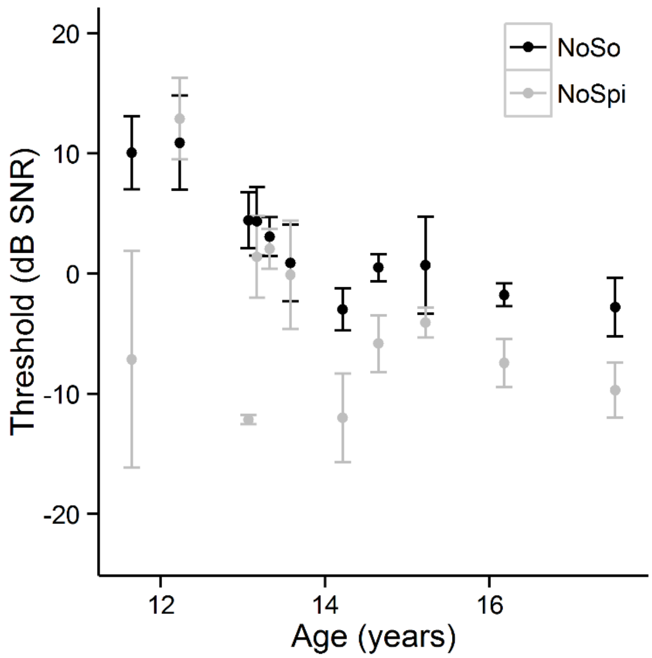

Figure 11. NoSo (black) and NoSπ (gray) thresholds (dB SNR) of the children with CIs as a function of age (years). Error bars show standard deviations.

Chapter 3

The relationship between intensity coding and dichotic signal detection and interaural time difference sensitivity in adults with cochlear implants

ABSTRACT

Many listeners with bilateral cochlear implants show sensitivity to binaural information when stimulation is provided using a pair of synchronized electrodes. However, there is large variability in binaural sensitivity between participants and within participants across stimulation sites in the cochlea. It was hypothesized that within-participant variability in binaural sensitivity is in part affected by peripheral limitations and characteristics, and can be related to monaural hearing performance. Diotic and dichotic signal detection thresholds and interaural time difference discrimination were measured at three stimulation sites for each listener. In addition, dynamic range and amplitude modulation detection were measured at each of the three stimulation sites in each ear separately. Finally, loudness growth was compared between ears. The results showed a relationship between diotic signal detection thresholds and the difference between the dynamic ranges of the two ears. A relationship between binaural sensitivity and dynamic range as well as the difference between comfortable levels of the two ears was also found. The results may be interpreted by assuming that dynamic range and the comfortable levels relate to peripheral neural survival and the width of the excitation pattern which could affect the fidelity with which central binaural nuclei receive bilateral input.

INTRODUCTION

Bilateral cochlear implantation has benefits over unilateral implantation for speech reception in noise and sound localization, but bilateral cochlear implants (CIs) do not provide all of the benefits that listeners with normal hearing derive from having two ears. Localization of sound sources is limited in listeners bilateral CIs (Verschuur *et al.*, 2005), and bilateral benefits for speech reception in noise depend on “head shadow” with little to no benefit from binaural hearing (Schleich *et al.*, 2004; Loizou *et al.*, 2009). Part of the deficit can be attributed to limitations in the information delivered by the devices and the fidelity with which the devices preserve interaural differences. However, there is evidence that even when listeners are provided with processors that encode and deliver information faithfully, binaural hearing for speech reception in noise is still non-existent (van Hoesel *et al.*, 2008).

Psychophysical studies examining binaural hearing of listeners with bilateral CIs have shown that many listeners are sensitive to interaural time and level differences (Litovsky *et al.*, 2010; Litovsky *et al.*, 2012) as well as to changes in interaural correlation (Long *et al.*, 2006; Goupell and Litovsky, 2015). Many of these studies have been done with single pairs of electrodes (one active electrode in each ear) in order to examine binaural sensitivity in the absence of interference from neighboring electrodes. However, even with single-electrode-pair stimulation, there is wide variability in the binaural hearing sensitivity of listeners with CIs. There is also variability in binaural hearing performance across stimulation sites (i.e., electrodes) within listeners, and to date there has been no evidence for a systematic effect of place of stimulation along the cochlear array (van Hoesel *et al.*, 2009; Litovsky *et al.*, 2010).

Some of this variability may be due to central binaural processing deficits caused by auditory deprivation (Litovsky *et al.*, 2010). Additionally, the extent to which there are differences in insertion depths between the left and right electrode arrays has been proposed as a source of variability in binaural hearing sensitivity of listeners with bilateral CIs (van Hoesel and Clark, 1997; Long *et al.*, 2003; Kan *et al.*, 2013). However, it seems likely that even without a mismatch in insertion depths between the two arrays, limitations in binaural hearing sensitivity may arise due to limitations at the auditory periphery.

Peripheral limitations may interfere with binaural sensitivity for two main reasons. First, central binaural processing requires input arising from each of the two individual ears. Therefore, deficits in the peripheral representation of signals at either ear could produce poor binaural hearing performance. Second, research in listeners with normal hearing suggests that binaural sensitivity is best when the inputs arise from frequency-matched peripheral locations (Henning, 1974; Nuetzel and Hafter, 1981). Even with pitch-matched inputs, many bilateral CI users have a significant gap in binaural sensitivity relative to normal hearing listeners. It may be the case that peripheral deficits such as poor neural survival limit the extent to which there can be peripherally matched stimulation.

Monaural psychophysical studies have shown variability across stimulation sites suggesting differences in the characteristics of the electrode-neuron interface across stimulation sites. Stimulation sites differ in the stimulation levels that produce just-audible sensations and maximum acceptable loudness (Pfingst and Xu, 2004). In addition, listeners show variability in performance across stimulation sites on tasks such as phase-duration modulation detection and temporal gap detection (Pfingst *et al.*, 2007; Garadat and Pfingst, 2011; Garadat *et al.*, 2012). These differences may in part reflect the shape and extent of electrical current spread as well as the auditory neural survival and tissue growth at different stimulation sites. Anatomical studies have shown that there is variability in the distance of electrodes from the auditory nerve fibers, the extent of damage and tissue growth within the cochlea, and the auditory nerve survival in listeners with CIs (Nadol, 1997; Finley *et al.*, 2008; Fayad *et al.*, 2009; Bierer, 2010). Furthermore, there is some evidence that characteristics of the electrode-neural interface relate to hearing performance. Kawano *et al.* (1998) found that greater spiral ganglion cell survival is associated with larger dynamic ranges (DRs) within some individuals with CIs.

The goal of the current study was to assess whether there is a relationship between monaural hearing performance, which we take, in part, to reflect the status of peripheral coding of the incoming signal, and binaural hearing sensitivity. A relationship between monaural measures and binaural measures could suggest that limitations at the auditory periphery affect binaural sensitivity (Hall *et al.*, 1984; van Hoesel, 2007; Churchill *et al.*, 2012).

Binaural Unmasking

For listeners with normal hearing, the introduction of an interaural time or phase delay in a speech signal improves speech reception thresholds in diotic noise (Schubert, 1956). Similarly, the introduction of an interaural time or phase delay in a tonal signal improves signal detection in diotic noise (Green, 1966). In the dichotic condition, interaural de-correlation is created in the temporal envelope and temporal fine structure of the combination of noise and signal, which results in information which can be used by the listener to aid in signal detection (Bernstein, 1991a; van de Par and Kohlrausch, 1995). The difference in signal detection ability between the diotic condition in which the signal and noise have the same interaural parameters and a dichotic condition in which the signal and noise have different interaural parameters is known as the binaural masking level difference (BMLD). Listeners with normal hearing can show BMLDs as large as 25 decibels (dB) (van de Par and Kohlrausch, 1997). Listeners with CIs have shown BMLDs when signal detection is measured using a limited number of electrodes in each ear (Long *et al.*, 2006; Lu *et al.*, 2010; 2011). However, there is variability across listeners in the magnitude of binaural unmasking demonstrated. For example, Long *et al.* (2007), found BMLDs in 7 listeners with CIs to range from approximately 3 dB to 24 dB.

For listeners with CIs, information is typically presented as modulations of the temporal envelope of moderate-rate electrical pulse trains (Loizou, 2006). With this type of information, binaural release from masking relies on the ability of the listener to detect interaural de-correlation in temporal envelope. There is evidence that even when temporal fine structure is presented in the electrical temporal envelope, due to calculation of the envelope by half-wave rectification, listeners with CIs rely on the slower modulations in the electrical stimulus representing the temporal envelope of the unprocessed stimulus, for binaural unmasking (Long *et al.*, 2006; Goupell and Litovsky, 2015). The ability to use interaural information in temporal envelope depends on the ability of the peripheral auditory system to encode changes in intensity in both ears. Therefore limitations in dynamic range (DR) and/or resolution for changes in intensity at the periphery in either ear could in and of themselves limit listeners' ability for binaural release with temporal envelope information. However, it may also be that small DRs and poor intensity resolution represent poor neural survival which we would expect to limit the extent to which there would be stimulation to corresponding left and right peripheral neural fibers (i.e., peripheral neural fibers which deliver information to the same central binaural neurons).

In addition to the encoding of intensity within each ear, the symmetry between ears in loudness growth could affect the listener's ability for dichotic signal detection. That is, the ability to detect a decrease in the interaural correlation, which is useful for dichotic signal detection, could be limited by any de-correlation introduced by asymmetries between the ears. Goupell *et al.* (2013) reported that, as the level of a diotic stimulus was varied across percentages of the DR, small deviations from the perception of a centered auditory image occurred, and suggested that this could result in the perception of interaural de-correlation for diotic stimuli. Similarly, we would expect that asymmetries in loudness growth between the two ears could create the perception of interaural de-correlation which would be detrimental to dichotic signal detection.

In this study, DR, amplitude modulation detection, and loudness growth were examined in order to evaluate whether these measures show a relationship with dichotic signal detection. It was expected that limitations in either ear for the DR and modulation detection measures could result in poor binaural sensitivity. For loudness growth, we were specifically interested in the difference between the ears in order to evaluate whether reduced symmetry results in poorer dichotic signal detection. Performance was measured at three stimulation sites per listener in order to examine the relationship between the monaural measures and dichotic signal detection within listeners across stimulation sites which we considered advantageous over examining the relationship across listeners as variability due to differences between listeners in attention and tolerance for loudness was removed.

In addition to the investigation of dichotic signal detection, diotic signal detection was measured, primarily, in order to evaluate whether listeners showed better performance in the dichotic condition which would indicate the use of binaural processing for the dichotic condition. Secondly, the monaural measures were examined in relationship to diotic signal detection. For listeners with CIs, diotic signal detection relies on the ability to detect the effect that the signal has on the temporal envelope of the noise. However, unlike dichotic signal detection, which relies on input from both ears, diotic signal detection only requires input from one ear in order to perform the task. Therefore, we expected that listeners' performance in diotic signal detection would show a relationship with measures made in the better performing ear. Differences between the measures made in each ear were also examined relative to diotic signal detection.

Interaural Time Difference Discrimination

For listeners with normal hearing, low-frequency interaural timing information plays a major role in sound lateralization and spatial release from masking (Bernstein and Trahiotis, 1985; Culling *et al.*, 2004). With single tones, interaural time difference (ITD) just-noticeable differences (JNDs) can be as low as 11 microseconds (µs) (Klumpp and Eady, 1956). For listeners with CIs, ITD discrimination, when measured with low-rate pulse trains to single electrode pairs can be as good as 25-50 µs, but JNDs are generally higher and vary between participants as well as within listeners across stimulation sites (Litovsky *et al.*, 2010; Litovsky *et al.*, 2012).

A relationship between dichotic signal detection and ITD discrimination has been found across listeners with CIs (Goupell and Litovsky, 2015). This relationship may be due to limitations at the auditory periphery which interfere with processing both types of stimuli. Therefore, we sought to determine whether a relationship between ITD discrimination and dichotic signal detection could be found within listeners, across different stimulation sites. Additionally, we examined the relationship between DR and ITD discrimination which allowed us to examine the relationship between DR and binaural sensitivity using stimuli with a constant-amplitude temporal envelope. A relationship between DR and ITD JNDs could suggest that DR represents limitations at the periphery which affect temporal encoding. However, again it may be the case that small DRs represent poor neural survival which we could limit the extent to which there is stimulation to corresponding left and right peripheral neural fibers.

In this study, the primary focus was to investigate whether variability in dichotic signal detection and ITD discrimination results from limitations at the auditory periphery in either ear. We hypothesized that monaural measures (i.e., DR and amplitude modulation detection) would predict dichotic signal detection, because these measures could reflect the integrity with which the temporal envelope is represented and coded. Alternatively, the results may show a relationship between dichotic signal detection, ITD discrimination, and the monaural measures. This finding could suggest that monaural measures represent the integrity with which stimulation is provided to left and right peripheral neural fibers which correspond in place and result in input to the same central binaural units.

METHODS

Participants and Equipment

Participants included 11 adults with bilateral CIs, all with Nucleus device types manufactured by Cochlear Ltd. Table I shows participant characteristics as well as electrode pairs used for the testing in the current study (note that lower numbered electrodes are located in the basal region of the electrode array manufactured by Cochlear Ltd). Stimuli were delivered using the Nucleus Implant Communicator and L34 processors.

Stimuli and Procedure

Electrical stimulation consisted of biphasic pulse trains in MP1+2 (monopolar) stimulation mode. Typically pulse durations were 25 µs per phase with an 8 µs interpulse gap. For participant ICP, a 75 µs phase duration was used in order to achieve levels that were loud enough. Stimuli were presented to individual electrodes (in each ear) or electrode pairs (for the binaural hearing measures) at the base, middle, and apical regions of the electrode array.

Pitch matching

Stimuli for the evaluation of binaural hearing sensitivity were presented to pitch-matched pairs of electrodes in attempts to reduce the impact of interaural place of stimulation mismatch on binaural hearing sensitivity. Stimuli for the pitch-matching task consisted of 300 ms constant amplitude pulse trains of 100 pps presented at comfortable (COM) levels. These stimuli matched the ones used for the ITD discrimination measure. For the pitch matching task, participants rated the pitch of a stimulus presented from an electrode in one ear relative to the pitch of a stimulus presented from an electrode in the other ear. For each stimulus presentation, participants were asked to select one of the following responses: (i) *much higher*, (ii) *higher*, (iii) *same*, (iv) *lower*, or (v) *much lower*. Typically, six electrodes on the right were judged relative to each of 3 electrodes on the left (thus 18 combinations). ICS was the only participant who did not complete the pitch-matching task on the visit in which the other measures were made. For this participant, pitch-matched pairs from a previous visit were used. As can be seen in Table I, pitch-matched pairs often did not deviate greatly from the number-matched pairs. IBQ stands out from the other participants in that she had a large and consistent interaural pitch mismatch.

NoSo and NoSπ detection

Diotic and dichotic signal detection thresholds were measured with 400 ms amplitude modulated pulse train of 1000 pulses per second (pps). Waveform stimuli were created at a sampling rate of 44100 Hz. Thirty-five samples of Gaussian noise were created in the frequency domain with a center frequency of 500 Hz, bandwidth of 50 Hz, and a duration of 400 ms. The target signal was a 300 ms sinusoid of 500 Hz which was temporally centered in the noise when presented. Both the signal and noise were ramped on and off with 50 ms Hann windows. The signal-to-noise ratio (SNR) of the tone and noise varied between 20 and -32 dB SNR. The tone was either presented with no interaural phase delay (So) or a 180 degree phase delay (Sπ). The noise was always diotic i.e., it was presented with no phase delay (No). The Hilbert envelopes of the waveform stimuli were calculated, and normalized such that the average amplitude was .4 which corresponds to 87 percent of the DR after compression. The envelope was resampled at 1000 Hz and was compressed between the listener's threshold (THR) and maximum acceptable loudness (MAL) levels using the compression function used by Long *et al.* (2006): $y = \mathrm{round}[(1 - e^{(-5.09 \cdot x)}) \cdot (M - THR) + THR]$ where x is the acoustic amplitude, y is the electrical amplitude in clinical units[1], THR is the listener's threshold, M is the maximum level at which stimulation is provided. Similar to what was done by Long et al. (2006), values 30 dB below a maximum acoustic amplitude of 1 were dropped in order to provide an input DR of 30 dB. The envelopes were presented with a 400 ms electrical pulse train of 1000 pps.

Loudness mapping was conducted in order to obtain THRs and MAL levels. Stimuli consisted of constant amplitude 400 ms pulse trains of 1000 pps. In order to measure THRs the experimenter manually increased the current level until the participant detected the signal three times consecutively. Two measures of THRs were obtained for each of the 6 electrodes used in the study. If the first two measures were more than 5 clinical units apart a third measure was obtained and the two closest values were averaged. MAL levels were defined for the participant as the loudest level that was still comfortable for short-duration listening or in other words the level just below *too loud*. To assess MAL levels, the experimenter manually increased the current level in small steps until the participant indicated that the level was high enough. Two measures of the MAL level were obtained. If the first two measures were more than 1 dB apart a third measure was obtained. All measures were averaged.

Using the diotic noise of the diotic/dichotic signal detection task, loudness balancing was performed for each of the 6 electrodes by adjusting the M levels used for the compression function. The M levels were initially set to the participant's MAL levels. Two intervals of diotic noise were presented sequentially, one from each of two electrodes, and the participant indicated which of the two stimuli was perceived to be louder, or whether they were the same perceived loudness. The process began with the left ear electrodes, whereby the M level of the basal and apical electrodes were adjusted to match the loudness of the middle electrode. Subsequently, the level of the electrodes in the right ear were adjusted to match the loudness of their counterparts on the left.

Additional M-level adjustments were made in order to ensure that, prior to testing, participants perceived auditory images that were at or near center. A single interval of diotic noise was presented. Participants indicated the intracranial perceived location of the stimulus. If the participant indicated that the stimulus was far off center, the experimenter stimulated each ear individually in order for the participant to hear the range of intracranial positions before re-examining the perceived position of the stimulus. Adjustments were made in order to center the image, for instance, by reducing the level of the ear that dominated the off-center image. However, effort was also made not to compromise the DR in order to obtain a centered image. Table II shows the maximum levels (Ms) that were used for No/NoSo/NoSπ stimulus presentation, as well as MAL levels. It can be seen that typically only minor differences are apparent between the Ms and MALs.

For participants IBR, ICA, and ICP, diotic and dichotic signal detection stimuli were compressed between THRs and approximately COM levels, instead of THRs and approximately MAL levels, which were subject to the same loudness balancing and centering. For participant IBP, diotic and dichotic signal detection stimuli were compressed both ways and signal detection thresholds were first measured with stimuli compressed beneath loudness-balanced and centered COM levels and then with stimuli compressed beneath loudness-balanced and centered MAL levels. The procedure for loudness mapping was changed after these first participants (IBR, ICA, ICP, IBP), because the participants were showing only minimal improvement in the dichotic signal detection task relative to the diotic condition. Table III shows M levels that were used for No/NoSo/NoSπ stimulus presentation for participants IBR, ICA, and ICP, as well as IBP when tested at the lower-level Ms.

Signal detection thresholds for NoSo and NoSπ were measured using a 3-interval 2-alternative forced-choice task in which the signal occurred in either the second or the third interval randomly determined on each trial. Non-target intervals consisted of diotic noise. Each interval contained a different sample of noise, which was randomly selected without replacement for each trial. The SNR was varied using a 2-down 1-up adaptive procedure beginning at 20 dB SNR. Initially the step size was 8 dB and changed to 4 dB after 1 turnaround and 2 dB after 3 turnarounds. Tracks were presented in blocks of NoSo and NoSπ to reduce the number of times the cues for the task switched. Blocks consisted of one adaptive track from each stimulation site (base, mid, or apex) presented in a newly randomized order for each block. At least four tracks were collected per stimulation sites. Initially, participants were familiarized with the stimuli by doing 10 to 20 trials of the signal detection task for NoSo at 12 dB SNR (or 20 dB SNR) and NoSπ signal detection task at 0 dB SNR at each stimulation site. Binaural masking level differences (BMLDs) were computed as the difference in dB between the threshold in the NoSo condition and the threshold in the NoSπ condition ($BMLD = Threshold_{NoSo} - Threshold_{NoS\pi}$).

Amplitude modulation detection

Amplitude modulation detection was measured at each of the 6 electrodes individually. Stimuli consisted of 400 ms pulse trains at 1000 pps. The standard was of constant amplitude. The target was sinusoidally amplitude modulated on a linear milliampere (mA) scale using the formula, $[f(t)][1 + m \cdot \sin(2 \cdot \pi \cdot f_m \cdot t)]$, where $f(t)$ is the average current, f_m is modulation rate of 30 Hz, and m is the modulation depth which was a proportion of 1 and varied during the experimental task. The stimuli were presented at loudness-balanced levels near 40 % of the DR (calculated using loudness balanced MAL levels) in clinical units. The loudness balancing used constant amplitude pulse trains at 1000 pps and followed the procedure used for the diotic and dichotic signal detection stimuli.

The amplitude modulation detection task consisted of a 2-interval 2-alternative forced-choice task. Participants were instructed to choose the interval that was fluctuating in loudness. The modulation depth was varied using a 2-down 1-up adaptive procedure. Initially the step size was 6 dB and changed to 3 dB after 2 turnarounds and 2 dB after 4 turnarounds. One track for each of the 6 electrodes was collected in a randomized order before the next set of tracks were collected in a newly randomized order. Three tracks were collected except if the standard deviation of the thresholds was more than 3 dB for any condition, a 4th track was collected for that condition and the threshold was the average of all of the tracks.

Interaural loudness balancing

Interaural loudness balancing was conducted using 400 ms constant amplitude pulse trains at 1000 pps. The three sites of stimulation (base, mid, apex) were examined separately. Stimuli were presented to the two ears sequentially. The level of the stimulus in one ear was held fixed and the level of the stimulus in the other ear was variable. The fixed stimulus was set to 40-90% of the DR in clinical units in 10% steps. The DR used was that which was used for the diotic and dichotic signal detection task (the higher levels for IBP). The interaural loudness-balancing task consisted of a 2-interval 2-alternative forced choice task in which the participant indicated which interval (1st or 2nd) was louder. The variable stimulus was randomly assigned to one of the two intervals. If the participant indicated that the variable stimulus was louder, the level of the variable stimulus was decreased. If the participant indicated that the fixed stimulus was louder, the level of the variable stimulus was increased. A double staircase adaptive track procedure was used in which for one track, the variable stimulus started 25 clinical units above the % DR of the track, and for the other track, the variable stimulus started 25 clinical units below the same % DR with the restriction that the track did not start above the MAL level or below THR. For most listeners, a 1-down 1-up adaptive procedure was used to track the level at which the participant indicated that the variable stimulus was louder 50 % of the time. For participant IBP, a majority decisions rule (Levitt, 1971; Zeng and Turner, 1991), (i.e., 2 consecutive or 2 out of 3 responses that either the variable or the fixed stimulus was louder resulted in an adjustment to the variable stimulus's level) was used which also tracked the 50% level. The adaptive track changed the current amplitude in 10 clinical-unit steps initially, 5 clinical-unit steps after one turnaround, and 3 clinical-unit steps after 3 turnarounds. For the majority of participants, two double-staircase procedures (4 tracks in total) were collected per % DR, one in which the stimulus in the left ear was fixed and one in which the stimulus in the right ear was fixed. All factor levels of stimulation site, fixed ear, and percent DR were fully randomized, except for ICJ whose mid pair was tested after the apical and base pair due to initial time limitations.

Interaural time difference discrimination

Stimuli used for measuring ITD discrimination consisted of 300 ms pulse trains at 100 pps presented at loudness-balanced and centered COM levels (C levels). Loudness mapping was conducted in order to find COM levels defined for the participants as a level with which they felt they could listen to throughout the day, and that was above the quiet range. THRs and MAL levels were also measured using these stimuli in order to calculate the DR at this pulse rate. The method used for measuring THRs and MAL levels was the same as the method used for the diotic and dichotic signal detection stimuli except that MAL levels were only measured once because the high level of current that the participants can tolerate at lower pulse rates can provide uncomfortable sensations such as facial twitching. Loudness balancing and centering were conducted using the method used for the diotic and dichotic signal detection stimuli in order to adjust the COM levels so that they were equal in loudness and approximately centered intra-cranially. Table IV shows THRs, MAL levels, and levels used for the ITD testing which were the loudness-balanced and centered COM levels.

A method of constant stimuli was used to measure ITD discrimination. A 2-interval 2-alternative forced-choice task was used in which the participants indicated the direction of perceived movement of the sound. Typically, ITD values of 50, 100, 200, and 800 μs were tested. These values were adjusted based on the sensitivity of the listener. Psychometric functions were calculated based on data from at least 4 ITDs and at least 40 trials per ITD (Wichmann and Hill, 2001a; b). JNDs were calculated from the psychometric function as the ITD which produced 70.7 % accuracy.

Correct answer feedback was provided for all tasks in which there was a correct answer, namely, diotic/dichotic signal detection, amplitude modulation detection, and ITD discrimination. All adaptive tracks stopped after 10 turnarounds and the threshold of each track was estimated as the average of the values on the last 6 turnarounds of the track.

RESULTS

Data were fit to linear mixed-effects models with random intercepts for participants, and random intercepts for stimulation sites for diotic and dichotic thresholds (for which multiple tracks were collected). F-tests were used to examine within-participant effects across the three measured stimulation sites unless stated otherwise. When multiple effects were included in the mixed-effects model, F-tests were conducted using Type II sums of squares. Clinical units were converted to mA and then into dB for the DR measure as well as the other measures resulting from loudness mapping. For each pulse rate, DR measures were made in two ways: (1) from THRs to Ms (maximum stimulus level used in the compression function) and (2) from THRs to MALs. The former is the DR with which the stimuli were presented i.e., the stimulus DR. The latter is the DR prior to loudness balancing and centering. The stimulus DR is presented in figures and is always presented before the DR calculated with MALs in the Results. ITDs in μs were log-transformed (base 10) for hypothesis testing.

When examining the relationship between the binaural and monaural measures, for each binaural measure, there were two associated monaural measures (i.e., left-ear and right-ear). Therefore, analyses were conducted using (1) the smaller monaural value, (2) the larger monaural value, (3) the average of the two monaural values, (4) and the difference between the two values. These will be indicated as subscripts e.g., for the DR: DR_{sml}, DR_{lrg}, DR_{avg}, and DR_{diff}. Each of these measures was calculated for each stimulation site (apex, mid, base) independent of the other sites. Differences between the measures from the left and right ears were calculated by subtracting one measure from the other (in dB) and taking the absolute value of the difference. It should be noted that the smaller value of the two measures was assumed to be produce poorer performance for DR but better performance for THRs and modulation detection thresholds.

For the NoSo/NoSπ measures the results focus on the eight participants whose NoSo/NoSπ stimuli were compressed between THRs and approximately MAL levels, but relevant data are shown for the participants whose NoSo/NoSπ stimuli were compressed between THRs and approximately COM levels. IBP's data with the second mapping procedure (using MAL levels) is presented unless otherwise indicated.

NoSo and NoSπ Thresholds

Average NoSo and NoSπ thresholds were .75 dB SNR (S.D. = 3.04) and -9.09 dB SNR (S.D. = 5.05), respectively, [$F_{1,168}$ = 198.61, P < .0001]. The BMLD was 9.8 dB on average (S.D. = 4.85). In Figure 1, it can be seen that all participants showed NoSπ thresholds that were lower on average than NoSo thresholds, and this occurred at each stimulation site. However, thresholds for both NoSo and NoSπ varied across site, as did the BMLD which are shown in Figure 2. The effect of place (apex, mid, base) was not significant [$F_{2,14}$ = 2.45, P = .12]. The phase × place interaction was also not significant [$F_{2,24}$ = 2.10, P = .14].

Figure 3 shows the NoSo and NoSπ thresholds for the participants whose stimuli were compressed between THRs and approximately COM levels. Average NoSo thresholds for this group were 5.24 dB SNR (S.D. = 4.74). Average NoSπ thresholds were 2.23 dB SNR (S.D. = 5.06). The BMLD was 3.00 dB on average (S.D. = 3.33). It should be noted that for IBP and IBR, average NoSπ thresholds were lower than average NoSo thresholds only at one stimulation site. Because it was suspected that the small BMLDs were due to the level of the stimuli, IBP was re-tested with stimuli compressed between THRs and approximately MAL levels. NoSo thresholds were 5.80 dB SNR and 0.72 dB SNR for the first and second data sets, respectively. NoSπ thresholds were 5.18 dB SNR and -3.61 dB SNR for the first and second data sets, respectively. BMLDs were .62 dB and 4.33 dB for the first and second sets of data, respectively.

Effect of dynamic range at 1000 pps

Figures 4 shows THRs (lower points) and Ms (higher points) for each participant for each place of stimulation at 1000 pps. The stimulus DR can be seen by the difference between the higher points and the lower points. THRs, Ms, and stimulus DRs, varied between participants, between different places of stimulation, and between ears.

The effect of stimulus DR on NoSo and NoSπ thresholds was examined. As the sole fixed effect in the model, DR_{lrg} was not significant in predicting NoSo thresholds [$F_{1,14}$ = 1.89, P = .19]. Figure 5 shows NoSo thresholds as a function of DR_{lrg}. The three stimulation sites per participant are grouped by color and a conjoining line. As can be seen, there was little evidence in improvement of NoSo thresholds with larger DR_{lrg}. There was also no significant effect of DR_{sml} [$F_{1,14}$ = .055, P = .82] or DR_{avg} [$F_{1,14}$ = .64, P = .43]. NoSo thresholds did get worse with larger DR_{diff} [$F_{1,14}$ = 6.51, P = .023]. Figure 6 shows NoSo thresholds as a function of DR_{diff}. There was a general increase in NoSo thresholds within participants across the three tested stimulation sites as the difference between the DRs of the left and right ears increased. IBF and ICJ were the only participants whose data notably differed from this pattern. For these two participants the best NoSo thresholds were obtained when the difference between DRs was the largest. Additionally, for a number of participants the effect of DR_{diff} appeared to weaken at larger DR_{diff} values. The effect of DR_{diff} on NoSo thresholds approached but did not reach significance when DR was calculated using MALs at 1000 pps [$F_{1,14}$ = 4.316, P = .057].

Figure 7 shows the relationship between NoSo thresholds and DR_{diff} for the four data sets for which NoSo stimuli were compressed using approximately COM levels in color. The eight data sets for which NoSo stimuli were compressed using approximately MAL levels are shown in black for reference. Standard deviation bars have been removed from the black points for clarity. The data resulting from the stimulus compression with approximately COM levels reasonably follow the same pattern as when stimuli were compressed with approximately MAL levels.

NoSπ thresholds decreased with larger DR_{sml} [$F_{1,14} = 8.12$, $P = .013$]. NoSπ thresholds also decreased with larger DR_{avg} [$F_{1,14} = 5.81$, $P = .030$]. The effect of DR_{lrg} was not significant [$F_{1,14} = 3.35$, $P = .088$]. Figure 8 shows that there was a general improvement in NoSπ thresholds as the smaller DR of the two ears increased. This pattern was less apparent for the two participants with the largest DR_{sml}, IBQ and IBK. DR_{diff} was not significant in predicting NoSπ thresholds as can be seen in Figure 9 [$F_{1,14} = 1.16$, $P = .30$]. The effect of DR_{sml} was also significant on NoSπ thresholds when DR was calculated using MALs at 1000 pps [DR_{sml}: $F_{1,14} = 7.74$, $P = .015$].

Figure 10 shows the relationship between NoSπ thresholds and DR_{sml} for the four data sets for which NoSπ stimuli were compressed using approximately COM levels. NoSπ thresholds did not necessarily decrease with increasing DR_{sml}. Perhaps this was because these listeners were relying less on binaural information when performing the task. The relationship between NoSπ and DR_{diff} was examined for these listeners, because it was thought that for these listeners, the NoSπ thresholds might be affected by DR_{diff} as the NoSo thresholds were. Figure 11 shows the relationship between NoSπ thresholds and DR_{diff} focusing on the four listeners. ICA and ICP's NoSπ thresholds did show some increase with higher DR_{diff}, but this was not the case for IBR or IBP suggesting that there is not a simple relationship between DR and NoSπ thresholds for lower-level stimuli.

Effect of THR and M levels at 1000 pps

NoSo and NoSπ thresholds were analyzed as a function of THRs and Ms at 1000 pps in order to examine whether the effect of DR was an effect of THRs or Ms. For NoSo, there was no effect of THR across stimulation sites (THR_{sml}: [$F_{1,14}$ = 1.40, P = .26]; THR_{lrg}: [$F_{1,14}$ = .21, P = .65]; THR_{avg}: [$F_{1,14}$ = .62, P = .44]; THR_{diff}: [$F_{1,14}$ = 1.37, P = .26]). Similarly, there was no effect of THRs on NoSπ thresholds (THR_{sml}: [$F_{1,14}$ = .23, P = .64]; THR_{lrg}: [$F_{1,14}$ = .53, P = .48]; THR_{avg}: [$F_{1,14}$ = .49, P = .49]; THR_{diff}: [$F_{1,14}$ = .15, P = .70]). There was no effect of M on NoSo thresholds (M_{sml}: [$F_{1,14}$ = .0052, P = .94]; M_{lrg}: [$F_{1,14}$ = .062, P = .81]; M_{avg}: [$F_{1,14}$ = .012, P = .91]; M_{diff}: [$F_{1,14}$ = 2.32, P = .15]), nor was there an effect of M on NoSπ thresholds (M_{sml}: [$F_{1,14}$ = 2.65, P = .13]; M_{lrg}: [$F_{1,14}$ = 1.27, P = .28]; M_{avg}: [$F_{1,14}$ = 2.97, P = .11]; and M_{diff}: [$F_{1,14}$ = 1.85, P = .20]).

Effect of MDTs

NoSo and NoSπ thresholds were examined as a function of modulation detection thresholds (MDTs) in dB. There was no effect of MDTs on NoSo thresholds (MDT_{sml}: [$F_{1,14}$ = .84, P = .37]; MDT_{lrg}: [$F_{1,14}$ = 1.00, P = .33]; MDT_{avg}: [$F_{1,14}$ = .97, P = .34]; MDT_{diff}: [$F_{1,14}$ = .015, P = .90]). There was also no effect of MDTs on NoSπ thresholds (MDT_{sml}: [$F_{1,14}$ = 2.85, P = .11]; MDT_{lrg}: [$F_{1,14}$ = 1.15, P = .30]; MDT_{avg}: [$F_{1,14}$ = 2.79, P = .12]; MDT_{diff}: [$F_{1,14}$ = .066, P = .80]). Figure 12 shows modulation detection thresholds by place of stimulation for each participant.

Interaural loudness balancing

For each stimulus that was presented at a fixed percentage of the DR in clinical units of a specific ear, the percentage of the DR of the stimulus in the other ear that provided a matched-loudness judgment was calculated. Clinical units were used because the envelopes were mapped to the electric DR in clinical units for the diotic and dichotic signal detection task. Each panel of Figure 13a and 13b shows the interaural loudness balancing data from a listener at a particular place of stimulation (apex, mid, base). Each data point of each panel shows percentages of the DRs of left and right ears that were matched in loudness through the adaptive procedure. Thresholds calculated from the upper and lower track of the double-staircase procedure have been averaged. Black symbols indicate that the stimulus in the left ear was fixed and the stimulus in the right ear was variable, and vice versa for the white symbols. The diagonal line shows the line of equality. Points falling below the line of equality can be interpreted as indicating that the right ear was louder and points falling above as indicating that the left ear was louder. A triangle was plotted when a value could not be estimated due to the necessity for the variable stimulus to exceed MAL levels. This occurred on one pair for IBK and two pairs for ICI.

Three values were calculated from the interaural loudness balancing task. The first measure was the root-mean-square (RMS) error of each data point (in % DR) from the line of equality (intercept = 0, slope = 1). Additionally, multiple RMS measures were made by calculating the RMS of lines of varying intercepts (in 1% steps; slope = 1). The minimum RMS value from this process was the second measure. The third measure was the intercept that provided the minimum RMS value. There was no relationship between the RMS error from the line of equality and NoSo or NoSπ thresholds [NoSo: $F_{1,12}$ = .00, P = .98; NoSπ: $F_{1,12}$ = .81, P = .39]. There was also no relationship between the minimum RMS error and NoSo or NoSπ thresholds [NoSo: $F_{1,12}$ = .14, P = .71; NoSπ: $F_{1,12}$ = .024, P = .88], nor was there a relationship between the absolute value of the intercept at which the minimum RMS error occurred and NoSo or NoSπ thresholds [NoSo: $F_{1,12}$ = .029, P = .87; NoSπ: $F_{1,12}$ = 1.19, P = .30]. Table V shows the RMS error from the line of equality, the minimum RMS error, and the intercept at which the minimum RMS error occurred for the participants at the basal, mid, and apical stimulation sites.

For 6 of the 7 participants who were tested on this measure, data were collected with both the left ear fixed and the right ear fixed. For IBF and IBQ (Figure 13a), estimates from the left-fixed and right-fixed conditions fit together to form the same line or curve. For IBK, ICJ, and ICT (Figure 13a and b), estimates from the left-fixed condition and the right-fixed condition gave the same general impression of interaural differences in loudness growth but were not necessarily well lined-up. However, for ICJ (Figure 13b) there were some consistent discrepancies between the left-fixed and right-fixed estimates in that in the right-fixed condition, there was a tendency to over-estimate the loudness on the right more so than in the left-fixed condition. ICI (Figure 13b) showed erratic deviations from the line of equality, but for each stimulation site, both left-fixed estimates and right-fixed estimates fairly consistently indicated one ear being louder than the other.

At times there were consistent deviations from a linear relationship between left- and right-ear loudness. IBF (Figure 13a) showed loudness growth that was well-matched between left and right ears, but as can be seen on the apical pair at the lowest percentages of the DR, there were discrepancies in loudness (deviations from the line of equality). This pattern of discrepancy in loudness at the lowest measured percentages of the DR was also apparent for IBP (base) and IBQ (mid) (Figure 13a). There was also evidence of a discrepancy in loudness growth between left and right ears in the middle percentages of the DR such that the lowest and highest percentages of the DR deviated less from the line of equality than the middle percentages. This was apparent for IBK (mid), IBQ (apex), and ICJ (mid and base) (Figure 13a and b).

No relationship was found between DR_{diff} (which showed a relationship with diotic signal detection thresholds) and the RMS error from the line of equality [$F_{1,12}$ = .043, P = .84], the minimum RMS [$F_{1,12}$ = .095, P = .76], or the absolute value of the intercept at which the minimum RMS error occurred [$F_{1,12}$ = .48, P = .50].

Effect of ITD JNDs

NoSπ thresholds were examined as a function of ITD JNDs. The relationship between ITD JNDs and NoSπ thresholds was not significant [$F_{1,14}$ = 4.26, P = .058]. As can be seen in Figure 14, the within-participant effect was inconsistent across participants. When IBP was removed from the dataset, the relationship between ITD JNDs and NoSπ thresholds was no longer near significance [$F_{1,12}$ = 1.00, P = .34]. Figure 15 shows the between-participant relationship between ITD JNDs and NoSπ thresholds. Each data point represents the average NoSπ threshold for a participant as a function of the participant-average ITD JND. From this small number of participants it seems as though a relationship may exist between NoSπ thresholds and ITD JNDs for lower-valued ITD JNDs.

ITD JNDs

Figure 16 shows ITD JNDs as a function of place of stimulation for each participant individually. There was no effect of place on ITD JNDs [$F_{2,20}$ = 1.65, P = .22].

Effect of dynamic range at 100 pps

Figures 17 shows THRs and C levels used for presenting the ITD stimuli at 100 pps for each place of stimulation. The effect of stimulus DR on ITD JNDs was examined. Stimulation sites with larger DR_{sml}, DR_{lrg}, and DR_{avg} had lower ITD JNDs [DR_{sml}: $F_{1,20}$ = 10.14, P = .0047; DR_{lrg}: $F_{1,20}$ = 8.89, P = .0074; DR_{avg}: $F_{1,20}$ = 10.85, P = .0036]. Both DR_{sml} and DR_{lrg} were only significant when the other was not included in the model. The relationship between DR_{left} and DR_{right} was significant [$F_{1,20}$ = 20.06, P = .00023]. The effect of DR_{diff} on ITD JNDs [$F_{1,20}$ = .13, P = .72] was not significant. Figure 18 and 19 show the relationship between ITD JNDs and DR_{sml} and ITD JNDs and DR_{lrg} and ITD JNDs, respectively. The relationship between DR and ITD JNDs did not appear as consistent as the relationship between DR and NoSπ thresholds. That is, there were a number of participants who did not show better ITD JNDs with larger DRs. The effect of DR_{avg} remained significant when IBP and ICA were removed from the analysis [DR_{avg}: $F_{1,16}$ = 6.39, P = .022]. The effect of DR_{sml} on ITD JNDs was significant when DR was calculated using MALs at 100 pps [DR_{sml}: $F_{1,18}$ = 4.42, P = .0498]. The effect of DR_{lrg} was no longer significant when DR was calculated using MALs at 100 pps [DR_{lrg}: $F_{1,18}$ = 2.49, P = .13], and the effect of DR_{avg} was not quite significant [DR_{avg}: $F_{1,18}$ = 4.15, P = .056].

The relationship between the stimulus DR for diotic and dichotic signal detection (1000 pps) and the stimulus DR for ITD discrimination (100 pps) is shown in Figure 20 (left ear) and 21 (right ear). There was not a consistent relationship between the two stimulus DRs across stimulation sites.

Effect of THR and C levels at 100 pps

There was no effect of THRs on ITD JNDs (THR_{sml}: [$F_{1,20}$ = 2.95, P = .10]; THR_{lrg}: [$F_{1,20}$ = 2.70, P = .12]; THR_{avg}: [$F_{1,20}$ = 3.28, P = .084]; THR_{diff}: [$F_{1,20}$ = .52, P = .48]). The effect of stimulation levels (C levels) on ITD JNDs was examined. There was a significant relationship between ITD JNDs and C_{diff} [$F_{1,20}$ = 9.82, P = .0052]. In Figure 22 it can be seen that ITD JNDs were poorer with larger C_{diff} in some of the participants. The relationship between ITD JNDs and C_{diff} remained significant when ICA was removed from the analysis [$F_{1,18}$ = 5.56, P = .030]. C_{sml}, C_{lrg}, and C_{avg} were not significant effects (C_{sml}: [$F_{1,20}$ = .24, P = .63]; C_{lrg}: [$F_{1,20}$ = 1.64, P = .21]; C_{avg}: [$F_{1,20}$ = .81, P = .38]). When both DR_{avg} and C_{diff} were included as effects in the model predicting ITD JNDs, both effects were significant [DR_{avg}: $F_{1,19}$ = 8.92, P = .0076; C_{diff}: $F_{1,19}$ = 7.93, P = .011].

DISCUSSION

In this study, we examined whether monaural measurements made in each of the ears of listeners with bilateral CIs could explain some of the variance in binaural sensitivity across different stimulation sites. A relationship between the monaural measures and the binaural measures could suggest that limitations at the auditory periphery affect binaural hearing in listeners with CIs. Loudness balancing and centering were conducted under the assumption that differences in loudness and the perception of centered images could affect binaural sensitivity across stimulation sites. In addition to DR and amplitude modulation detection, symmetry in loudness growth between the left and the right ears was examined. Binaural measures included dichotic signal detection (NoSπ) and ITD discrimination. The influence of monaural measures on diotic signal detection (NoSo) was also examined.

Diotic signal detection

It was expected that monaural encoding of intensity could affect diotic signal detection because the diotic signal detection task required the listener to detect characteristics of the temporal envelope of the stimuli to detect the presence of the tone. The hypothesis regarding performance in the diotic condition was that diotic signal detection would be affected by the better performing ear (left or right) more so than the poorer performing ear, because the information necessary to do the task should be more available in the better performing ear. No relationship was found between diotic signal detection and modulation detection thresholds. Also, no relationship was found between diotic signal detection and the larger DR or smaller DR of the two ears. It was assumed that the ear with the larger DR would be the better performing ear; however this did not necessarily have to be the case. It may be that for listeners who have larger DRs, information is mapped to lower levels if the DR is larger because the listener is able to hear very low sounds. However, diotic signal detection thresholds were not predicted by either the larger or smaller DR or modulation detection threshold.

The results did show that diotic signal detection thresholds were worse when there was a greater difference between the DRs of the two ears. This indicates that variability across stimulation sites was not related to the size of the DR but rather the difference between the DRs of the two ears. This suggests that when there were discrepancies between the DRs, this resulted in either masking between the two ears or a lack of summation of information from the two ears.

For speech reception in quiet, listeners with CIs have shown summation of information from left and right ears i.e., better performance in a bilateral condition than with either left or right CI alone (Litovsky *et al.*, 2006a). However, there is reason to believe that summation was unlikely given the stimuli used in the current study. Studies with normally hearing listeners have found that there is no summation with NoSo signal detection (i.e., monaural and diotic signal detection thresholds are the same) when the level of the noise is high enough and the sample of noise changes from trial to trial (Langhans and Kohlrausch, 1992a). Given that the stimuli were compressed between thresholds and approximately MAL levels and that numerous samples of noise were used in the current study, it is not expected that diotic signal detection thresholds would have been better than the best monaural performance.

We could expect that if the internal representation of the stimuli differs in each ear, this could result in contralateral masking between the ears. While small differences were sometimes found in the loudness growth between the left and right ears of the listeners, no relationship was found between the discrepancies in loudness growth and the difference between the DRs of the left and right ears or diotic signal detection thresholds. However, the loudness growth does not tell us how well each ear was able to follow the modulations of the stimuli. The neural encoding of the stimulus envelope may have been distorted to varying degrees due to neural adaptation and refractoriness (Jeng *et al.*, 2009). Therefore, the neural representations of the envelopes in each ear may have been quite different from each other, but it is not obvious why this would be associated with a difference between the DRs of the two ears.

In this study we did not test left and right monaural signal detection. A comparison between monaural signal detection in each ear and diotic signal detection would provide evidence as to whether there was interference or a lack of summation when the DRs differed between the ears. Lu *et al.* (2010) examined monaural and diotic signal detection in five listeners with CIs. Three of the listeners performed much worse in at least one of the monaural conditions relative to the diotic condition suggesting that they may have been relying on the better-performing ear in the diotic condition. Two of the listeners performed similarly between the monaural and diotic conditions suggesting no benefit from summation in these listeners. However, the very poor performance of some of the listeners suggests the need to further assess the relationship between monaural and diotic signal detection in listeners with CIs. A monaural signal detection measure also would allow us to assess the effect of DR on signal detection for listeners with CIs in the absence of the effect of the difference between the DRs of the two ears, since potentially, an effect of DR (as opposed to the difference between DRs) could be found on monaural signal detection.

Dichotic signal detection

All eight listeners at all places showed lower dichotic thresholds than diotic thresholds suggesting that they were using binaural processing to detect the signal in the dichotic condition. The average BMLD of the eight listeners was 9.8 dB which is comparable to the average BMLD (with 50 Hz bandwidth noise) found by Long *et al.* (2007) (BMLD = 12 dB) and that found by Goupell and Litovsky (2015) of (BMLD = 8.5 dB), but it is larger than that found by Lu *et al.* (2010) (BMLD = 4.9 dB).

It was expected that dichotic signal detection would be affected by the performance of the poorer performing ear more so than the better performing ear, because the ability to use interaural differences for signal detection relies on the listener making use of information from both ears. Similar to diotic signal detection, no relationship was found between dichotic signal detection and amplitude modulation detection. Amplitude modulation detection was examined at a low level of the DR while the diotic and dichotic signal detection stimuli varied across the DR which may have contributed to the lack a relationship between modulation detection thresholds and signal detection thresholds since somewhat different neural populations were involved with the two sets of stimuli.

Dichotic signal detection was worse when the smaller DR of the two ears was smaller. It has been suggested that smaller DRs are related to poor neural survival (Kawano *et al.*, 1998; Bierer and Nye, 2014). For CI users, there can be degeneration of peripheral processes of neurons as well as loss of spiral ganglion cells (Fayad and Linthicum, 2006). It has been estimated through histological evaluations of human temporal bones with CIs that spiral ganglion neural survival is on average 25 percent of what it is in normal human temporal bones, with wide variability between individuals (Pfingst *et al.*, 2011). Kawano *et al.* (1998) found spiral ganglion cell survival in human temporal bones to positively correlate with DR within some individuals. It would be expected that poor neural survival would make it such that a larger spread of current is needed to achieve sufficient loudness (Cohen *et al.*, 2006). With poorer survival, the likelihood of stimulating peripheral neural fibers on the left and right which provide input to the same central binaural processing units would be reduced. This would likely be the case regardless of whether there is poor neural survival on one or both sides. Therefore, this could be an explanation for the finding that dichotic signal detection thresholds were predicted by the smaller DR of the two sides. A similar explanation of the relationship between DR and dichotic signal detection thresholds could be provided by the existence of fibrous tissue and bone growth in the cochlea which has been found in implanted cochleae and which may impede the electrical current from the neural elements (Kawano *et al.*, 1998; Fayad *et al.*, 2009).

It is reasonable to assume that smaller DRs do not reflect quieter stimuli since the stimuli were adjusted to reduce any loudness differences and adjustments to the stimuli made for centering were minor. One caveat to this statement is that listeners were asked to judge loudness of sounds on electrodes that differed by place, and therefore the percepts elicited by the electrodes were likely different in pitch and/or quality. Tolerance for loudness at one place of stimulation might be different from tolerance for loudness at a different place of stimulation due to differences in pitch or quality which could have played a role in the loudness balancing.

We would expect quieter stimuli to reduce the number of neural fibers which were stimulated during the dichotic signal detection task.

IBQ and IBK were two subjects whose dichotic signal detection thresholds did not show a consistent pattern with DR. This was likely because these two participants had large DRs at all places of stimulation and therefore performance did not depend on DR. There was some indication that IBQ's data which showed large variability across stimulation sites may have been affected by the difference between the DRs of the two ears as can be seen in Figure 9. However, there were not enough participants who had especially large DRs to know if this pattern is at all reliable.

The participants who were tested at lower levels of stimulation (stimuli compressed using approximately COM levels) showed smaller BMLDs compared to the listeners whose stimuli were presented using higher levels. Compared with single-electrode-pair stimulation, multi-electrode-pair stimulation requires lower levels per electrode to maintain comfortable loudness. This suggests that part of the reason that listeners with CIs fail to show binaural release from masking with multi-electrode stimulation may be that interaural information is less salient for each individual electrode. It would be informative to determine whether listeners with CIs can demonstrate BMLDs for single-electrode-pair stimulation using the levels of stimulation which are needed for their clinical maps, which are intended for multi-electrode stimulation.

No relationship was found between symmetry in loudness growth between the left and right ears and dichotic signal detection thresholds despite some participants showing small but reliable deviations in loudness growth between the ears. Similar to the current study, Fu (2005) compared loudness growth between two different places of stimulation, however, he did so monaurally between an apical stimulation site and a basal stimulation site. Inconsistencies in loudness growth between the two sites were found for some listeners, but this only occurred at a stimulation rate of 100 pps not 1000 pps. Cohen *et al.* (2006) furthermore found that there can be variability between listeners in the slopes of loudness growth functions at 250 pps. Possibly, greater deviations in loudness growth may have been found between ears in the current study had lower pulse rates been of interest (1000 pps was used in the current study). It may be that differences in loudness growth between the ears affects binaural hearing but we were unable to show it by comparing the performance between the loudness growth and dichotic signal detection measures. It may be revealing to examine the effect of deviations in loudness growth between the ears on binaural hearing by manipulating the compression function between the ears independently. The finding of a relationship between diotic signal detection thresholds and differences between the DRs of the two ears suggest asymmetries between the ears that one would expect could affect dichotic signal detection as well.

ITD JNDs

A relationship was found between ITD JNDs and DR within participants similar to what was found for dichotic detection thresholds. Similar to dichotic detection thresholds, if we assume that DR reflects neural survival, the relationship between ITD JNDs and DR can be explained on the basis of neural survival. When there is a high level of neural survival, there would likely be stimulation to a greater number of corresponding left and right neural fibers.

However, the finding that both ITD JNDs and dichotic signal detection thresholds are related to DR does not necessarily imply the same explanation for each relationship. It is possible that DR is associated with each of two different mechanisms that limit ITD discrimination and dichotic signal detection separately. It should be noted that the relationship between ITD JNDs and DR appeared less consistent than the relationship between dichotic signal detection thresholds and DR in that a number of listeners did not show a relationship between ITD JNDs and DR.

There was no relationship between ITD JNDs and dichotic signal detection thresholds across stimulation sites despite the previous finding of a relationship between the two measures between participants (Goupell and Litovsky, 2015), and the appearance of a relationship between the two measures at lower-valued ITD JNDs between participants in this study. The lack of a relationship between the two binaural measures across stimulation sites may be related to the fact that different pulse rates (100 pps vs. 1000 pps) and therefore different stimulation levels were used for the two types of stimuli. Furthermore, there was an inconsistent relationship between the DRs in each ear at the two different pulse rates. That there was an inconsistent relationship between DRs at low and moderate rate pulse rates could possibly be explained in part by the finding that thresholds decrease with higher rates at some stimulation sites more so than others (Zhou *et al.*, 2012). The lack of a consistent relationship between ITDs and dichotic signal detection thresholds furthermore suggests that performance of stimulation sites is dependent on pulse rate and/or current level.

In addition to a relationship between ITD JNDs and DR, a relationship was also found between the differences in stimulation levels (C levels) between ears and ITD JNDs. This suggests that either low or high C levels can result in relatively good ITD sensitivity, but a problem occurs when the C levels differ. One explanation for this result is that differences in C levels between the ears reflect a difference in the shape and/or width of current spread between the ears. Cohen *et al.* (2006) found that higher comfortable levels were associated with larger widths of excitation. Therefore, differences between the ears in C levels could have resulted in differences in current spread between the ears which could indicate reduced stimulation to corresponding left and right neural fibers.

There is evidence that comfortable levels are affected by the distance of the electrodes from the modiolus and thus from the neural elements. Comparison of comfortable levels between listeners with straight electrode arrays and perimodiolar arrays have shown that comfortable levels (and thresholds) are lower with perimodiolar arrays which is likely due to the closer proximity of the electrodes to the neural fibers with the perimodiolar array (Parkinson *et al.*, 2002; Saunders *et al.*, 2002). Saunders *et al.* (2002) found that it is not atypical for comfortable levels (and thresholds) within individual listeners to be lower for electrodes that are closer to the modiolus. Furthermore, greater distance of electrodes from the modiolus has also been found to relate to larger widths of excitation (Cohen *et al.*, 2003). Therefore, differences in C between the ears may possibly reflect differences in the distance of the electrodes from the neural fibers.

The reasons for limitations in binaural sensitivity at individual stimulation sites is likely multi-faceted including peripheral limitations such neural survival as well as asymmetries between ears in the shape and spread of excitation. The findings of a relationship between binaural sensitivity and DR as well as between binaural sensitivity and the difference in C levels between the ears in this study provides some support for this idea. Psychophysical measure made in either ear are limited in their interpretation because the relationships between psychophysical measures and the characteristics at the auditory periphery of listeners with CIs is likely complex. However, knowledge of the relationship between monaural measures and binaural hearing sensitivity should provide some insight into why limitations exist in binaural sensitivity at individual stimulation sites. Ideally, these problems can be addressed to provide listeners with CIs better access to the advantages of binaural hearing.

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ENDNOTES

[1] One clinical unit is equal to .1759 dB for the cic3 internal devices (CI24M, CI24R, CI24R(CS)). One clinical unit is equal to .1569 for the cic4 internal devices (CI24RE, CI24RE(CA), CI512, CI513).

Table I. Participant characteristics, device types, and tested electrodes.

Participant	Age (yrs)	Duration unilateral (yrs)	Duration bilateral (yrs)	Internal device L	Internal device R	Apex pair (L,R)	Mid pair (L,R)	Base pair (L,R)
IBF	63	8.6	7.1	CI24RE(CA)	CI24RE(CA)	4,5	12,12	21,21
IBK	74	11.5	5.6	CI24R(CS)	CI24RE(CA)	4,5	12,12	18,18
IBP	64	10.3	9.7	CI24M	CI24M	4,8	12,14	20,17
IBQ	83	12.1	9.2	CI24RE(CA)	CI24R(CS)	8,2	12,5	20,15
IBR	59	9.9	6.3	CI512	CI24R(CS)	4,6	12,12	18,16
ICA	54	12.0	5.2	CI24RE(CA)	CI24R(CS)	3,3	12,13	20,20
ICI	56	5.7	5.0	CI24RE(CA)	CI24RE(CA)	4,8	12,14	20,20
ICJ	65	0	4.8	CI512	CI512	4,6	12,10	20,16
ICP	51	5.2	2.2	CI24RE(CA)	CI24RE(CA)	4,8	12,14	20,20
ICS	87	12.0	4.0	CI513	CI24R(CS)	4,5	12,12	18,19
ICT	21	2.6	2.6	CI512	CI512	4,3	12,12	20,19

Table II. Thresholds (THRs), maximum levels (Ms) used for the diotic/dichotic signal detection task, and maximum acceptable loudness levels (MALs) in clinical units at 1000 pps for each electrode on the left (L) and right (R) for each of the participants whose M levels were near MAL levels.

Participant	Elect. (L)	THR (L)	MAL (L)	M (L)	Elect. (R)	THR (R)	MAL (R)	M (R)
IBF	4	121	175	173	5	114	169	169
	12	114	197	197	12	121	193	191
	21	128	185	185	21	130	194	194
IBK	4	143	236	234	5	143	238*	238
	12	156	244	244	12	145	238*	236
	18	161	244	242	18	147	235*	233
IBP	4	159	221	220	8	142	204	202
	12	143	210	210	14	129	201	201
	20	147	211	211	17	130	201	201
IBQ	8	130	225	225	2	122	216	216
	12	119	225	225	5	135	215	214
	20	130	222	222	15	128	215	215
ICI	4	133	173	172	8	124	168	168
	12	133	188	188	14	133	173	173
	20	124	159	159	20	119	165	163
ICJ	4	124	179	174	6	139	168	162
	12	107	173	173	10	134	173	170
	20	93	163	159	16	92	154	148
ICS	4	158	196	198	5	158	195	195
	12	123	193	193	12	157	210	210
	18	114	191	193	19	141	201	201
ICT	4	83	148	146	3	114	154	156
	12	94	154	154	12	103	162	156
	20	91	150	152	19	95	161	158

*Limited by twitching/physical sensation as opposed to loudness

Table III. Thresholds (THRs), maximum levels (Ms) used for the diotic/dichotic signal detection task, and maximum acceptable loudness levels (MALs) in clinical units at 1000 pps for each electrode on the left (L) and right (R) for each of the participants in which M levels were near COM levels.

Participant	Elect. (L)	THR (L)	MAL (L)	M (L)	Elect. (R)	THR (R)	MAL (R)	M (R)
IBR	4	115	173	162	6	119	188	167
	12	116	184	176	12	138	209	200
	18	114	186	172	16	130	203	192
ICP	4	105	189	179	8	145	215*	206
	12	90	161	146	14	117	192	177
	20	79	146	127	20	75	162	145
ICA	3	107	194	186	3	155	210	197
	12	163	233	218	13	175	219	206
	20	148	232	201	20	168	221	208
IBP	4	159	221	209	8	142	204	194
	12	143	210	198	14	129	201	187
	20	147	211	203	17	130	201	187

*Limited by twitching/physical sensation as opposed to loudness

Table IV. Thresholds (THRs), maximum acceptable loudness levels (MALs), and levels (Cs) used for ITD discrimination task in clinical units at 100 pps for each electrode on the left (L) and right (R)

Participant	Elect (L)	THR (L)	MAL (L)	C (L)	Elect (R)	THR (R)	MAL (R)	C (R)
IBF	4	130	215	192	5	123	208	193
	12	140	225	204	12	139	222	213
	21	140	195	183	21	136	213	198
IBK**	4	155	NM	240	5	154	NM	243*
	12	189	NM	250	12	170	NM	243
	18	191	NM	246	18	172	NM	240
IBP	4	180	231	212	8	167	212	210
	12	172	227	212	14	160	216	204
	20	176	219	189	17	161	214	194
IBQ	8	151	245	242	2	174	239	239
	12	164	246	243	5	171	239	239
	20	170	231	228	15	170	230	227
IBR	4	158	197	189	6	160	195	187
	12	155	197	195	12	160	211	206
	18	153	193	190	16	147	210	205
ICA	3	117	206	199	3	164	227	216
	12	171	242	233	13	178	224	221
	20	186	247	239	20	189	223	219
ICI	4	163	192	183	8	157	181	175
	12	173	199	196	14	161	184	180
	20	153	174	171	20	132	177	174
ICJ	4	151	196	193	6	146	199	190
	12	155	195	186	10	148	198	190
	20	133	197	192	16	108	190	172
ICP	4	121	224*	218	8	161	220*	217
	12	132	197	180	14	144	216*	187
	20	119	170	148	20	113	219*	155
ICS	4	180	219	208	5	167	225	210
	12	160	244	205	12	186	232	215
	18	163	222	209	19	173	239	215
ICT	4	100	157	148	3	145	164	164
	12	126	162	159	12	138	170	158
	20	99	148	132	19	134	160	148

* limited by facial twitching/physical sensation as opposed to loudness

**MALs not measured because of twitching near comfortable levels

Table V. Calculations made from the interaural loudness balancing data as percentage of the DR in clinical units for each left-right electrode pair tested: the RMS error from the line of equality (RMS), the minimum RMS error (min. RMS) to a linear fit with slope = 1, and the intercept of the linear fit for which the minimum RMS deviations occurred.

Participant	Elect. (L, R)	RMS (% DR)	Min. RMS (% DR)	Intercept of min. RMS (% DR)
IBF	4, 5	2.31	2.28	-1
	12, 12	4.03	1.90	-4
	21, 21	4.41	4.02	-2
IBK	4, 5	10.97	8.36	-7
	12, 12	10.17	8.61	-5
	18, 18	6.65	5.28	-4
IBP	4, 8	8.70	5.52	7
	12, 14	4.29	4.12	-1
	20, 17	5.65	2.34	-5
IBQ	8, 2	9.29	5.32	8
	12, 5	5.23	3.49	-4
	20, 15	5.46	4.31	3
ICI	4, 8	13.30	12.66	4
	12, 14	11.87	9.00	-8
	20, 20	28.07	10.23	-25
ICJ	4, 6	11.08	7.84	-8
	12, 10	6.96	3.66	-6
	20, 16	8.92	6.42	-6
ICT	4, 3	8.11	5.08	6
	12, 12	6.16	5.17	-3
	20, 19	8.40	4.12	7

Figure 1

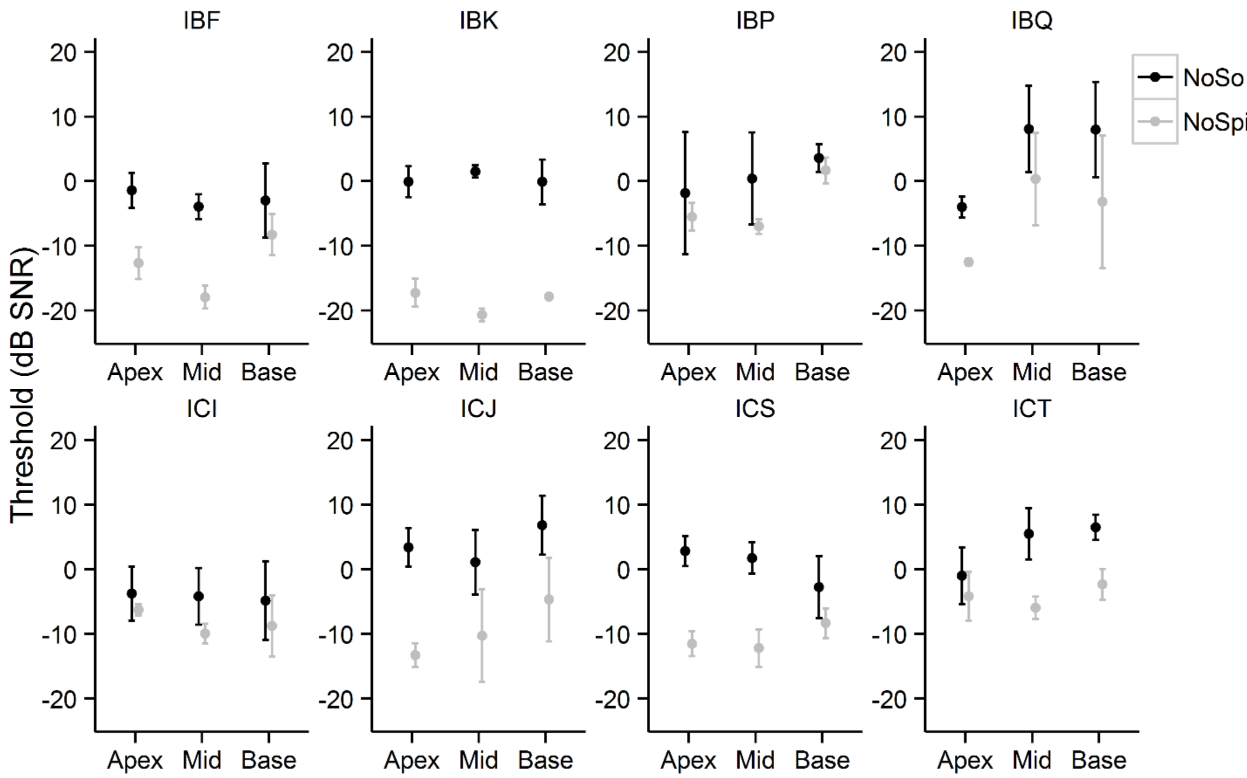

Figure 1. Diotic (NoSo; black) and dichotic (NoSπ; gray) signal detection thresholds (dB SNR) as a function of place of stimulation. Each panel shows data from an individual participant. Error bars show standard deviations.

Figure 2

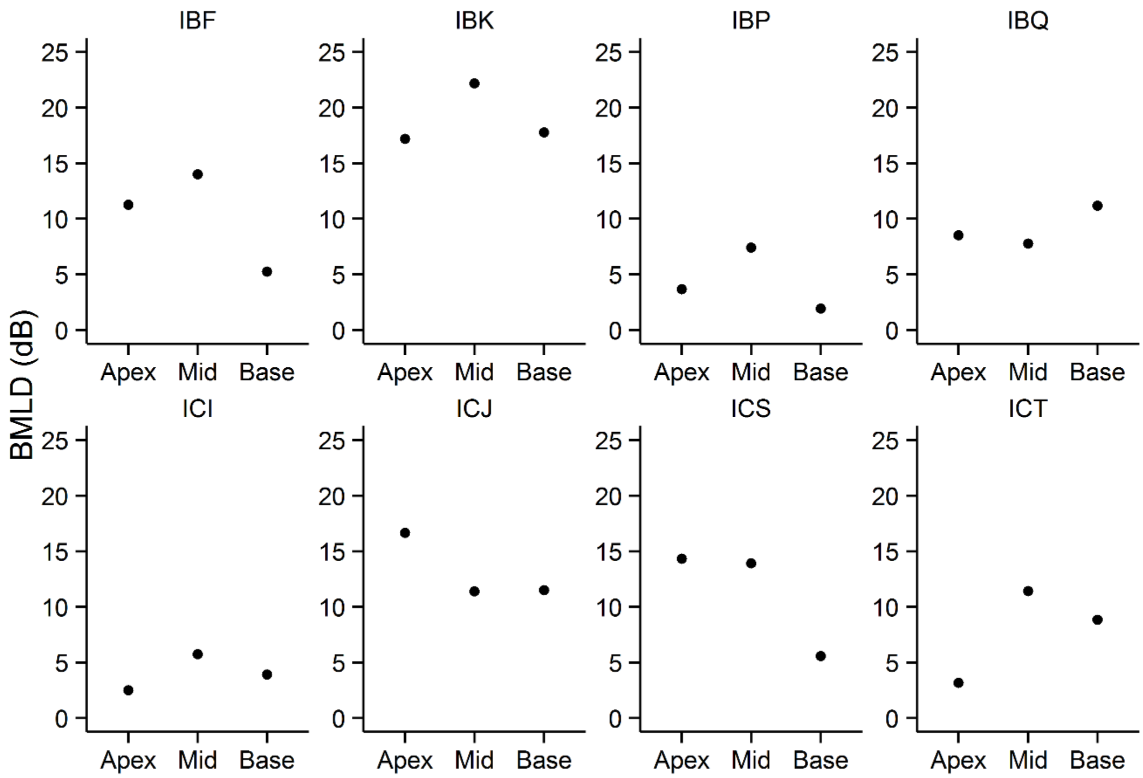

Figure 2. BMLDs (dB) as a function of place of stimulation. Each panel shows data from an individual participant.

Figure 3

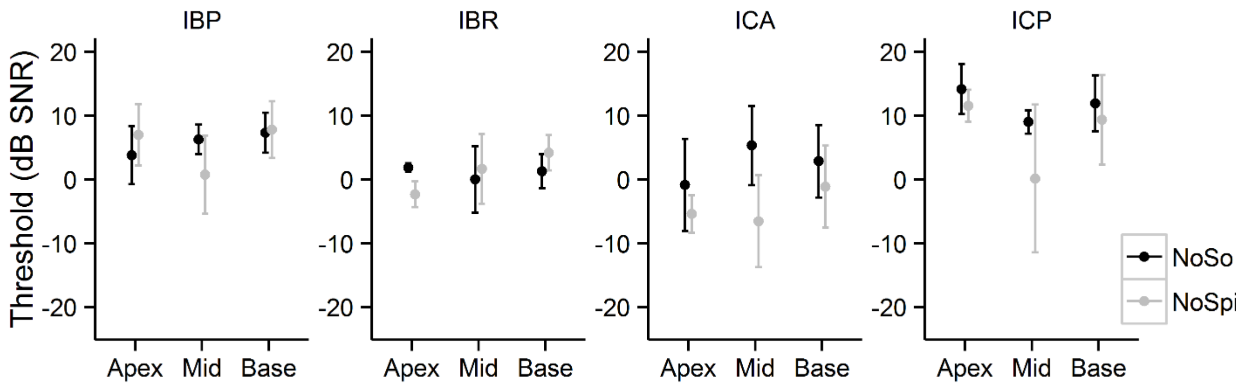

Figure 3. Diotic (NoSo; black) and dichotic (NoSπ; gray) signal detection thresholds (dB SNR) as a function of place of stimulation. Each panel shows data from an individual participant whose stimuli were compressed using approximately COM levels. Error bars show standard deviations.

Figure 4

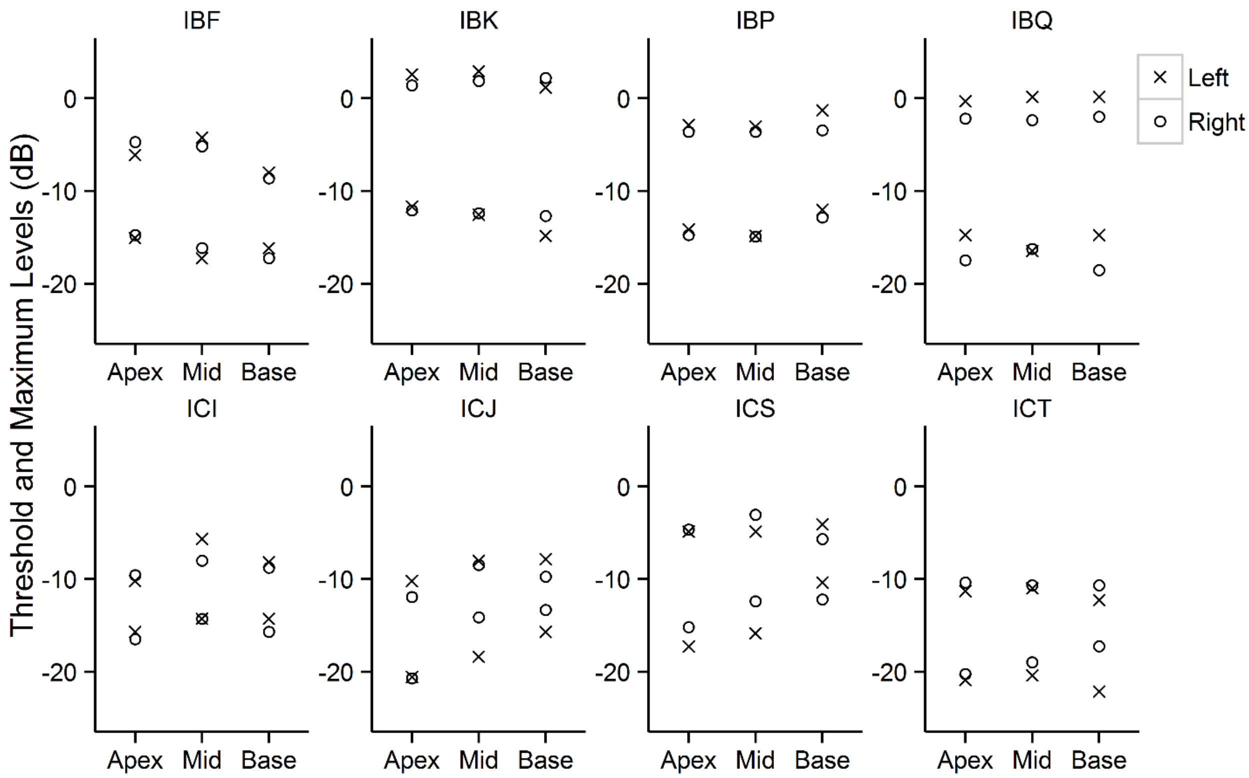

Figure 4. Thresholds (lower) and maximum levels (higher) used for the signal detection stimuli (in dB re 1 mA) for left and right ears (indicated by symbols) as a function of place of stimulation. Each panel shows data from an individual participant.

Figure 5

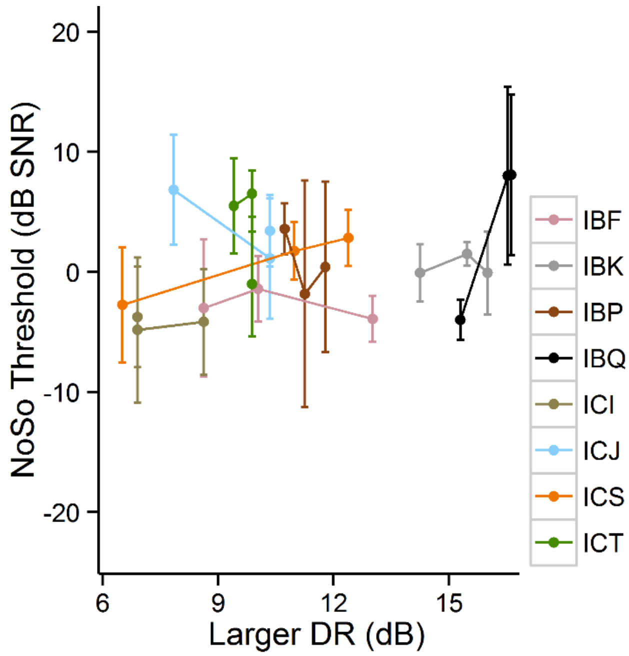

Figure 5. Diotic signal detection thresholds (dB SNR) as a function of the larger DR (dB) of the two ears. Thresholds of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant. Error bars show standard deviations.

Figure 6

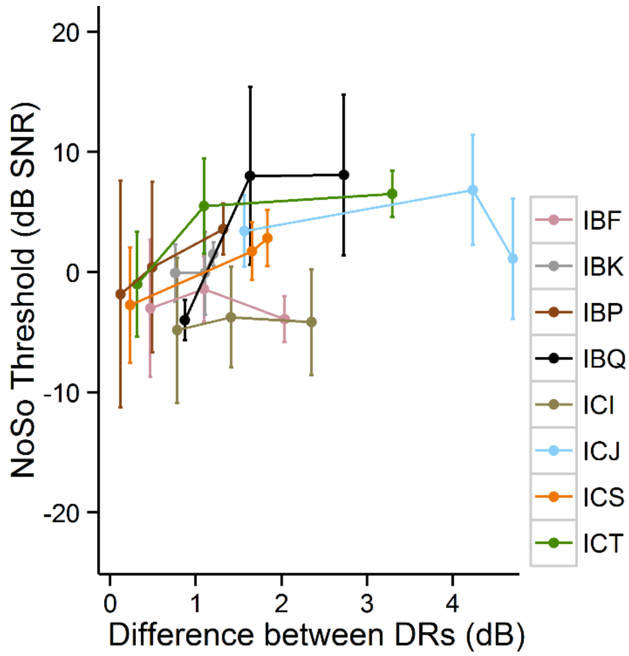

Figure 6. Diotic signal detection thresholds (dB SNR) as a function of the absolute value of the difference between the DRs (dB) of the two ears. Thresholds of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant. Error bars show standard deviations.

Figure 7

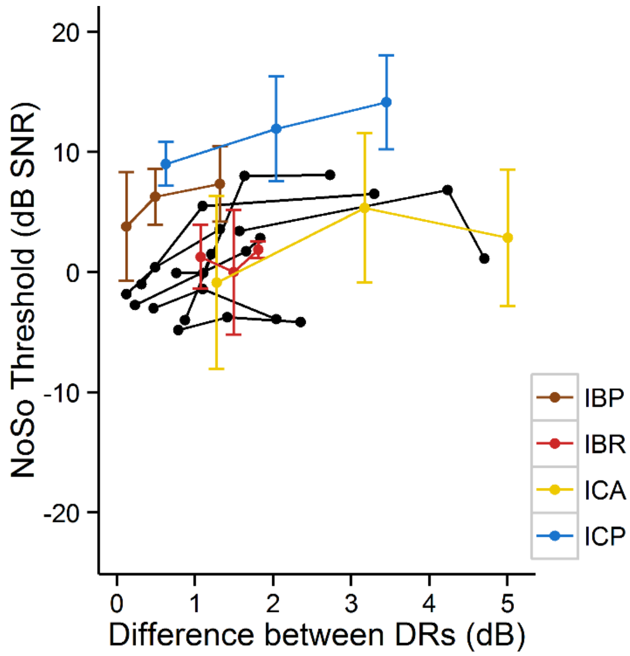

Figure 7. Diotic signal detection thresholds (dB SNR) as a function of the absolute value of the difference between the DRs (dB) of the two ears. Three places of stimulation are shown per participant. Performance of 4 participants whose stimuli were compressed using approximately COM levels is shown in color. Error bars show standard deviations. Performance of 8 participants whose stimuli were compressed using approximately MAL levels is shown in black. Error bars have been removed for clarity. Data from participant IBP appear in both groups.

Figure 8

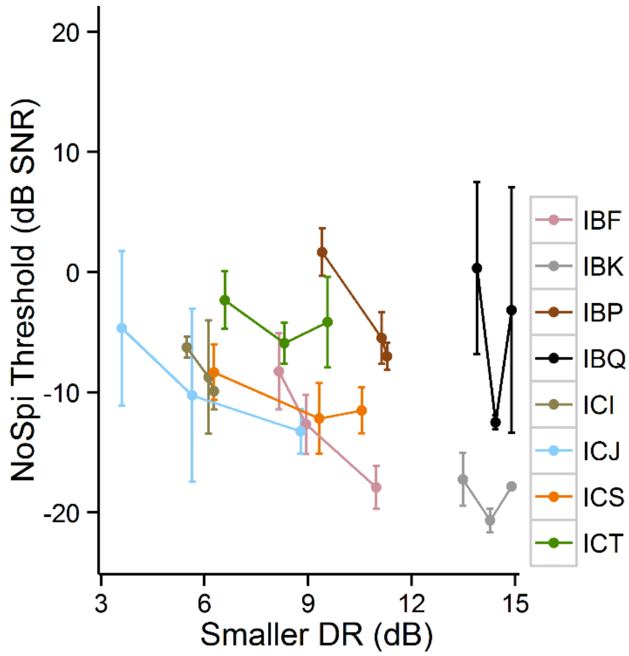

Figure 8. Dichotic signal detection thresholds (dB SNR) as a function of the smaller DR (dB) of the two ears. Thresholds of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant. Error bars show standard deviations.

Figure 9

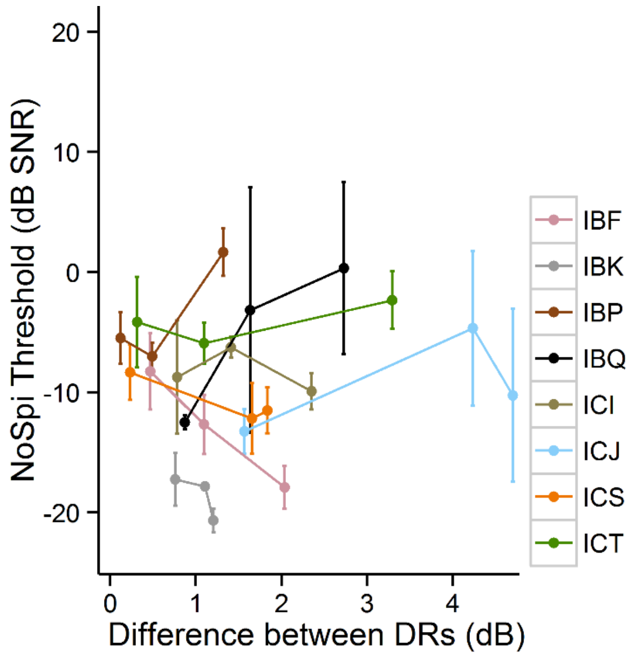

Figure 9. Dichotic signal detection thresholds (dB SNR) as a function of the absolute value of the difference between the DRs (dB) of the two ears. Thresholds of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant. Error bars show standard deviations.

Figure 10

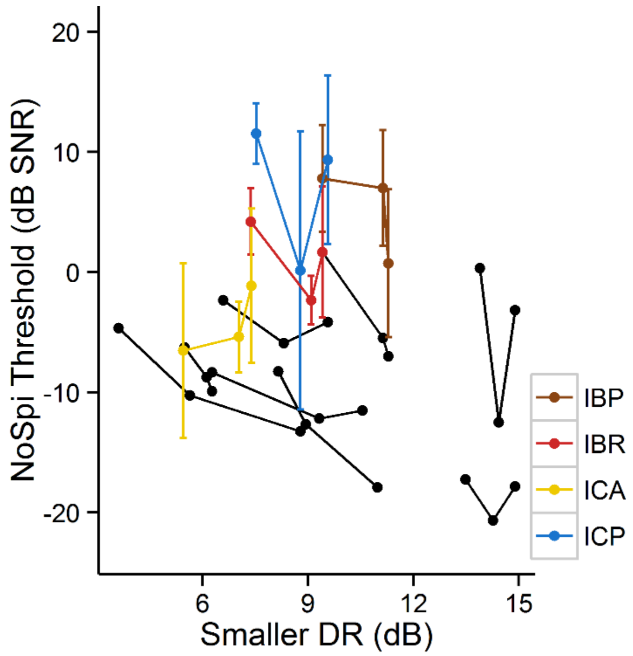

Figure 10. Dichotic signal detection thresholds (dB SNR) as a function of the smaller DR (dB) of the two ears. Three places of stimulation are shown per participant. Performance of 4 participants whose stimuli were compressed using approximately COM levels is shown in color. Error bars show standard deviations. Performance of 8 participants whose stimuli were compressed using approximately MAL levels is shown in black. Error bars have been removed for clarity. Data from participant IBP appears in both groups.

Figure 11

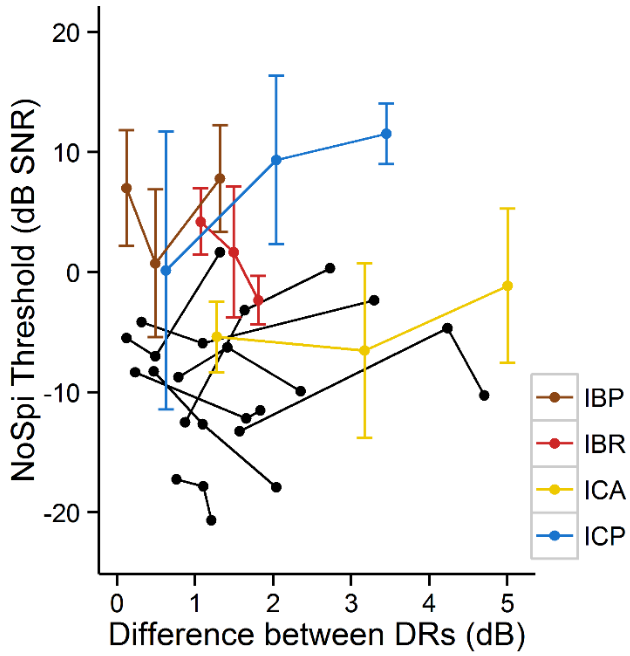

Figure 11. Dichotic signal detection thresholds (dB SNR) as a function of the absolute value of the difference between the DRs (dB) of the two ears. Thresholds of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant. Error bars show standard deviations.

Figure 12

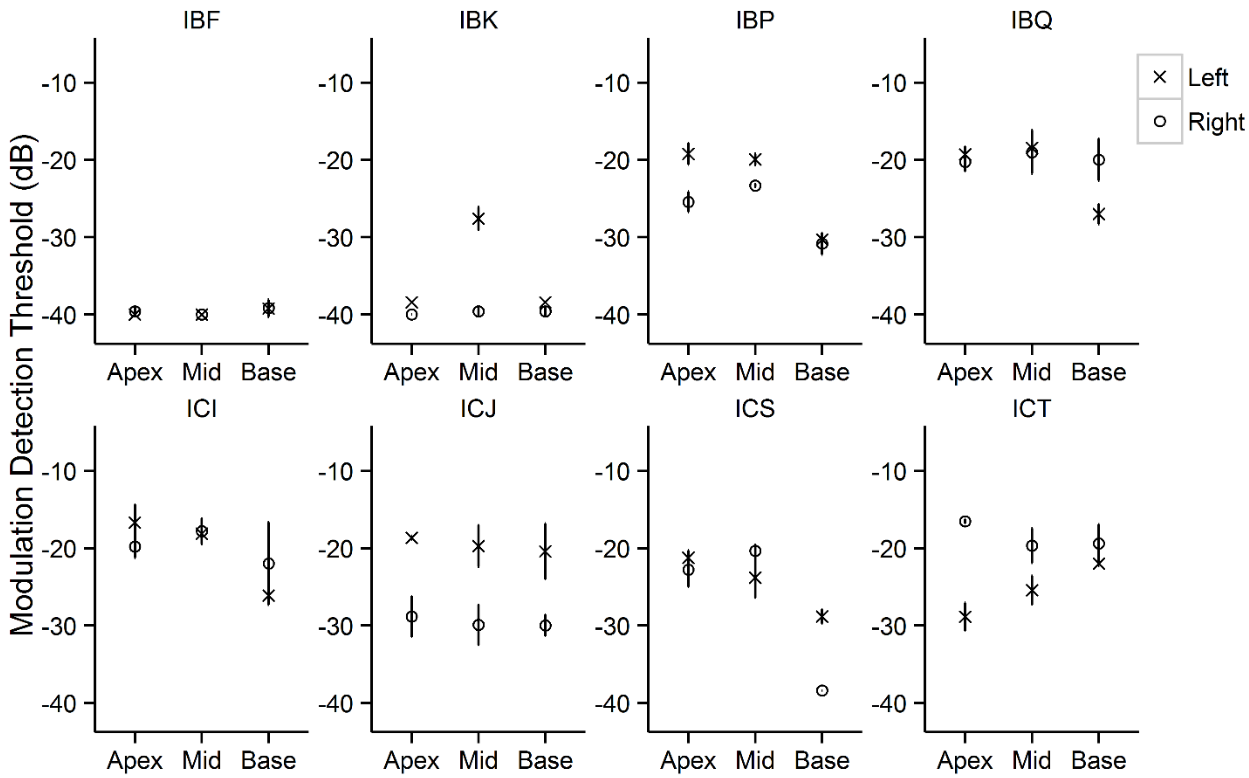

Figure 12. Amplitude modulation detection thresholds (dB) as a function of place of stimulation. Each panel shows data from an individual participant. Thresholds for left and right ears are indicated by symbol. Error bars show standard deviations.

Figure 13a

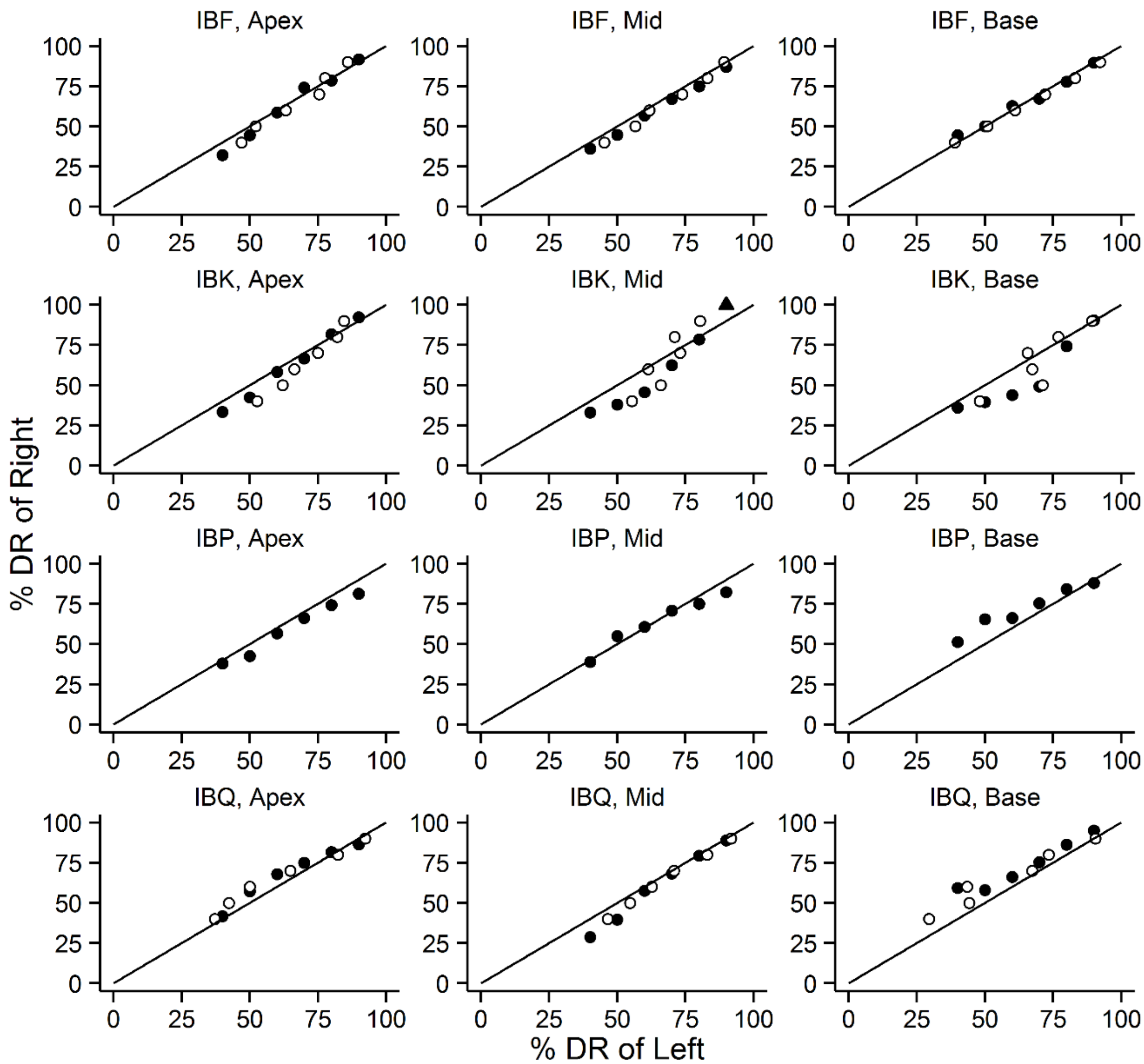

Figure 13a. Interaural loudness balancing data from participants IBF, IBK, IBP, and IBQ. Percentage of the DR on the right (ordinate) matching in loudness to a percentage of the DR on the left (abscissa) is shown. The diagonal line shows the line of equality. Each row shows data from an individual participant. Each column shows data from a specific place of stimulation. White points represent the data for which the right-ear stimulus was fixed and the left-ear

variable. The black points represent the data for which the left-ear stimulus was fixed and the right-ear was variable.

Figure 13b

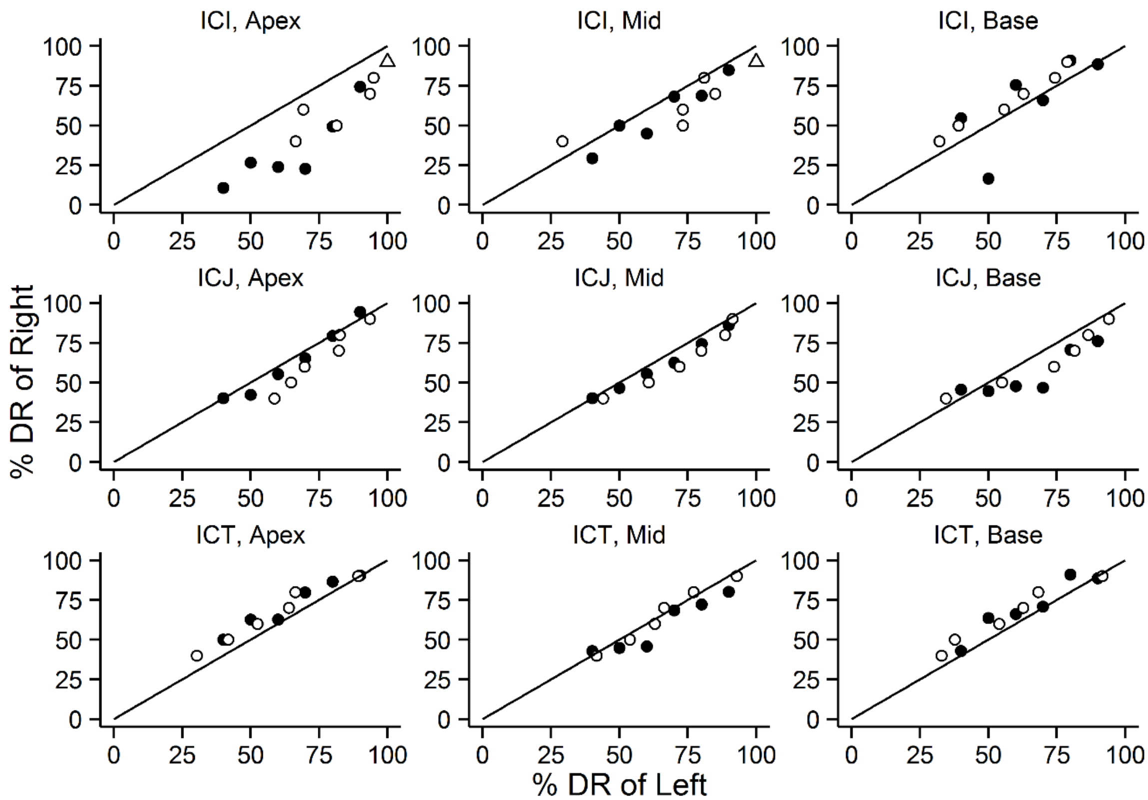

Figure 13b. Interaural loudness balancing data from participants ICI, ICJ, and ICT. Percentage of the DR on the right (ordinate) matching in loudness to a percentage of the DR on the left (abscissa) is shown. The diagonal line shows the line of equality. Each row shows data from an individual participant. Each column shows data from a specific place of stimulation. White points represent the data for which the right-ear stimulus was fixed and the left-ear variable. The black points represent the data for which the left-ear stimulus was fixed and the right-ear was variable.

Figure 14

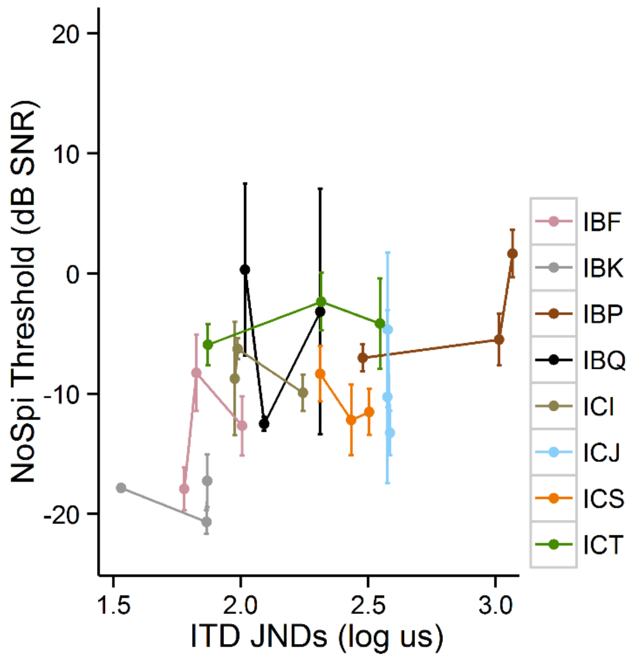

Figure 14. Dichotic signal detection thresholds (dB SNR) as a function of ITD JNDs (in $\log_{10}$ µs). Thresholds of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant. Error bars show standard deviations.

Figure 15

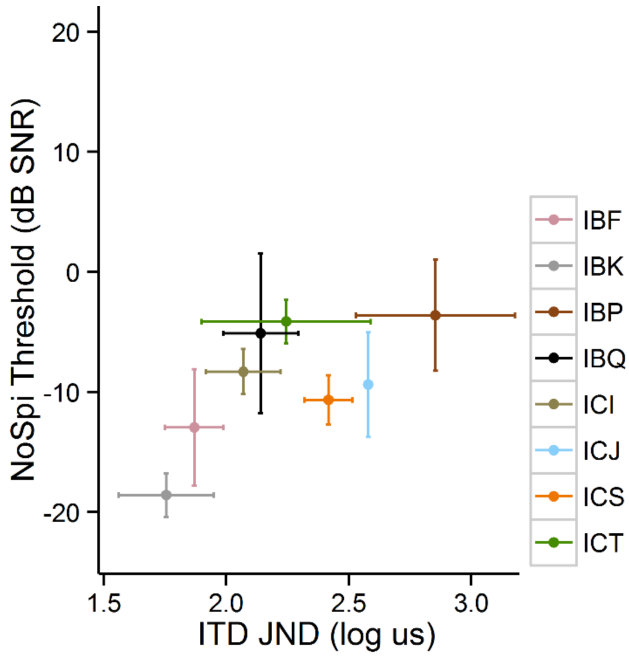

Figure 15. Dichotic signal detection thresholds as a function of ITD JNDs (in $\log_{10}$ µs). There is one point per participant. Error bars show standard deviations.

Figure 16

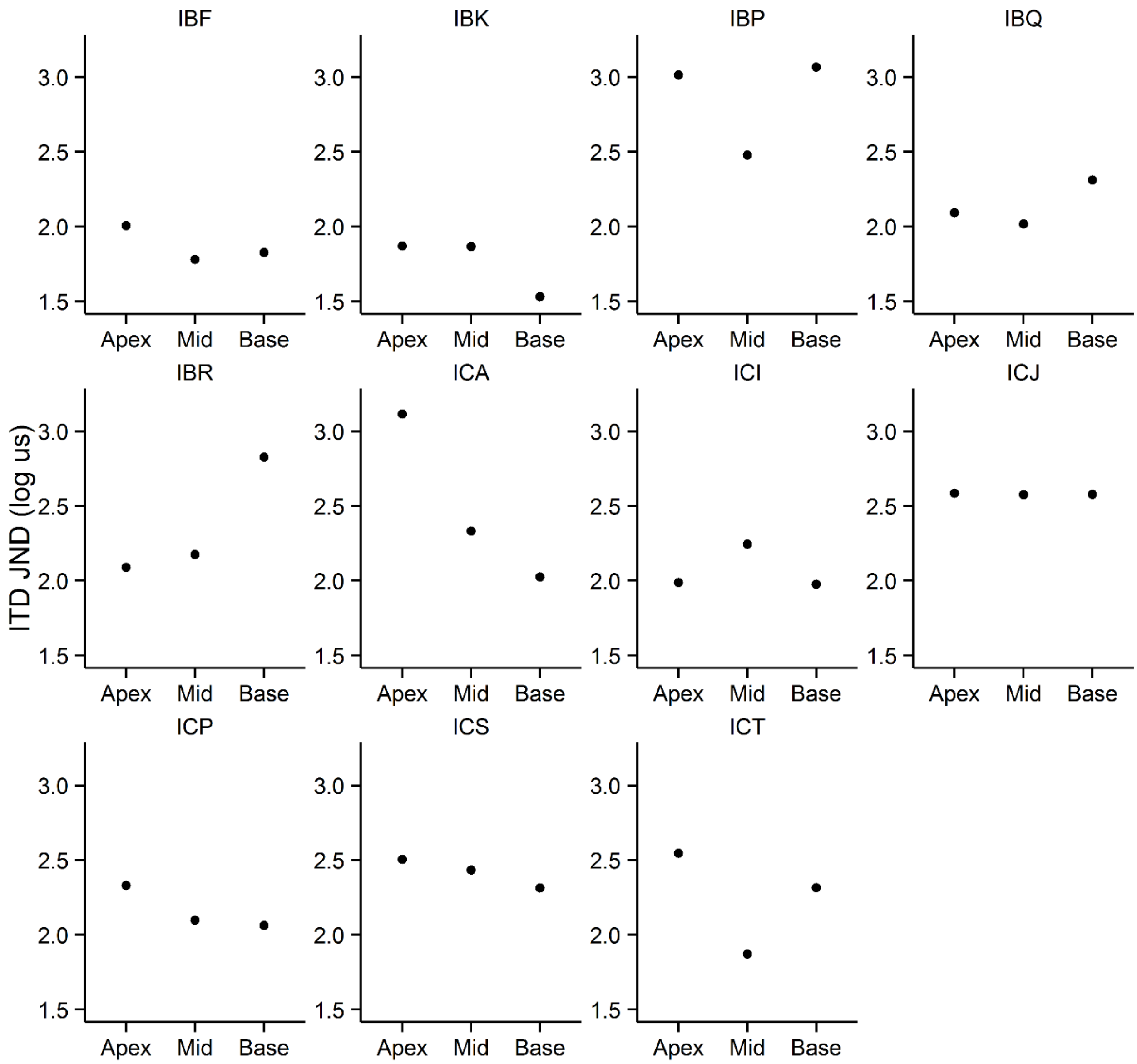

Figure 16. ITD JNDs (in $\log_{10}$ µs) as a function of place of stimulation. Each panel shows data from an individual participant.

Figure 17

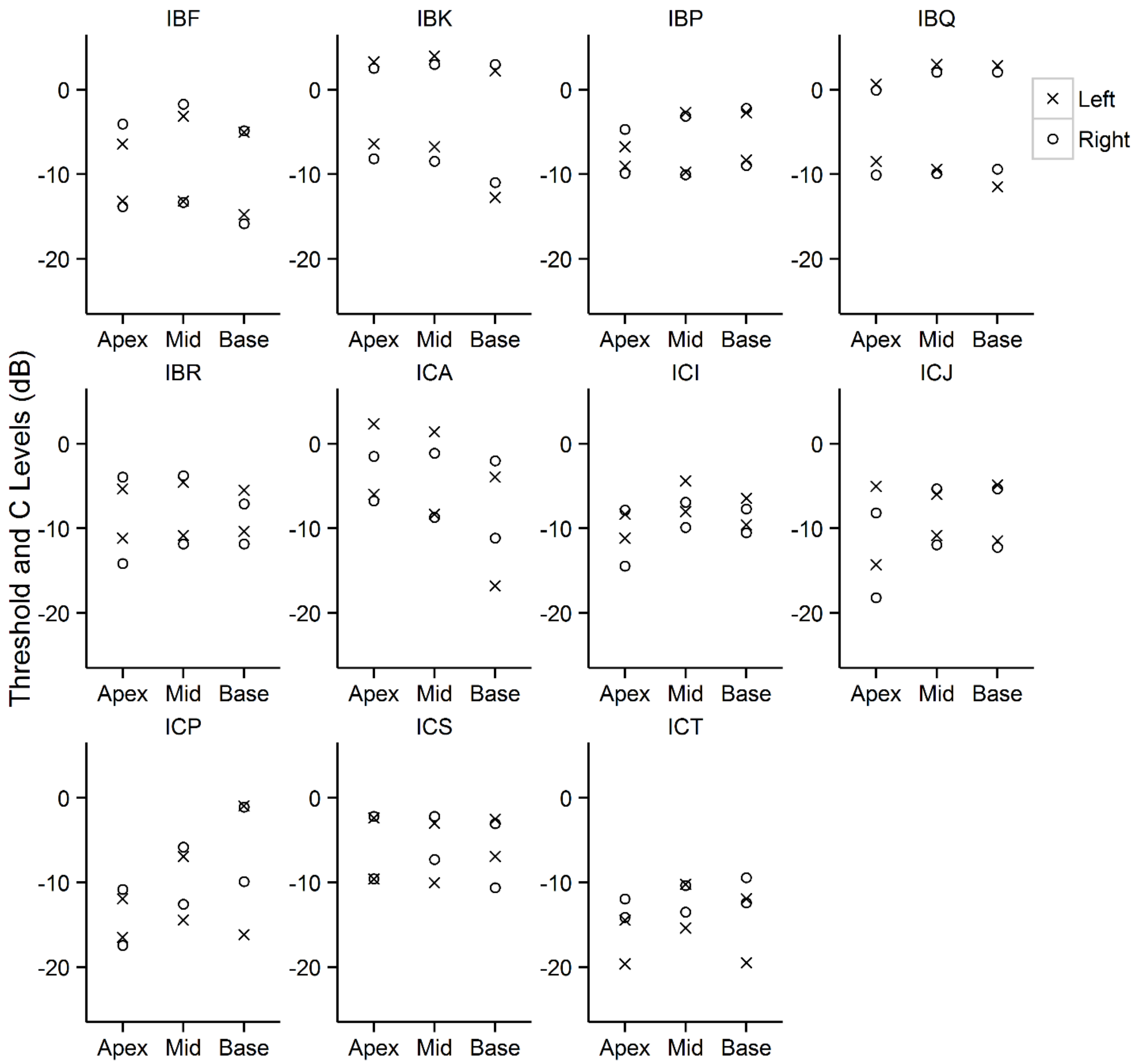

Figure 17. Thresholds (lower) and C levels (higher; used for the ITD discrimination task) (in dB re 1 mA) for left and right ears (indicated by symbols) as a function of place of stimulation. Each panel shows data from an individual participant.

Figure 18

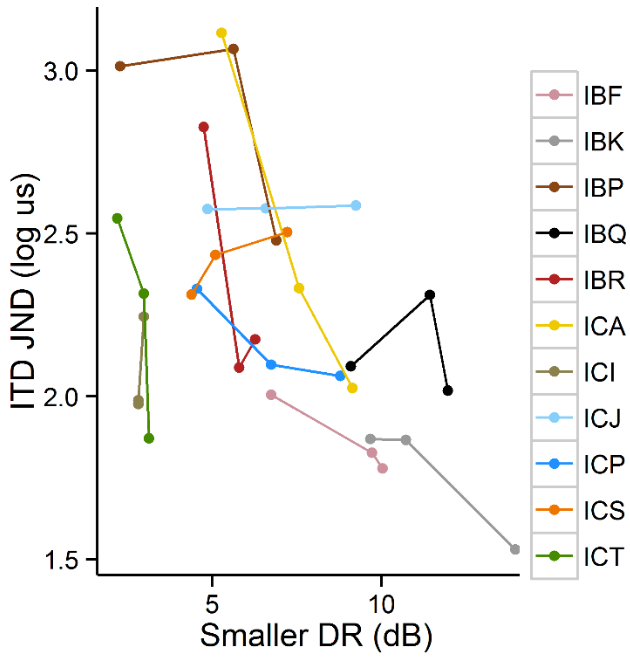

Figure 18. ITD JNDs (in $\log_{10}$ µs) as a function of the smaller DRs (dB) of the two ears. JNDs of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant.

Figure 19

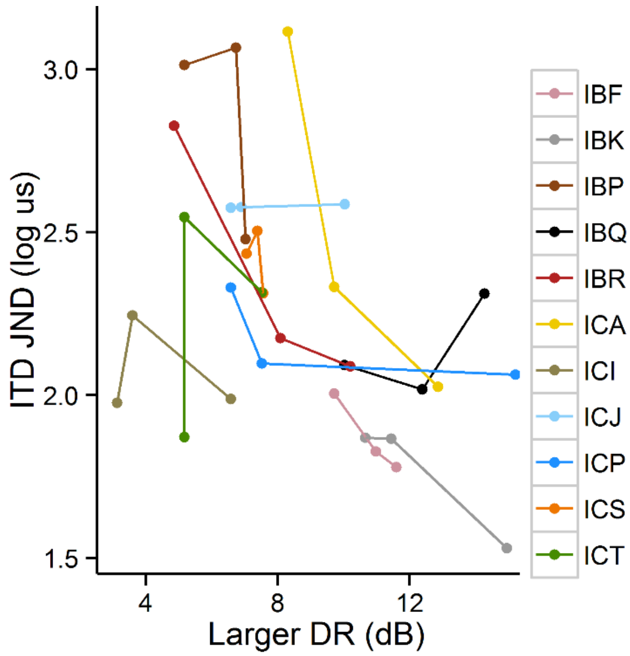

Figure 19. ITD JNDs (in $\log_{10}$ µs) as a function of the larger DRs (dB) of the two ears. JNDs of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant.

Figure 20

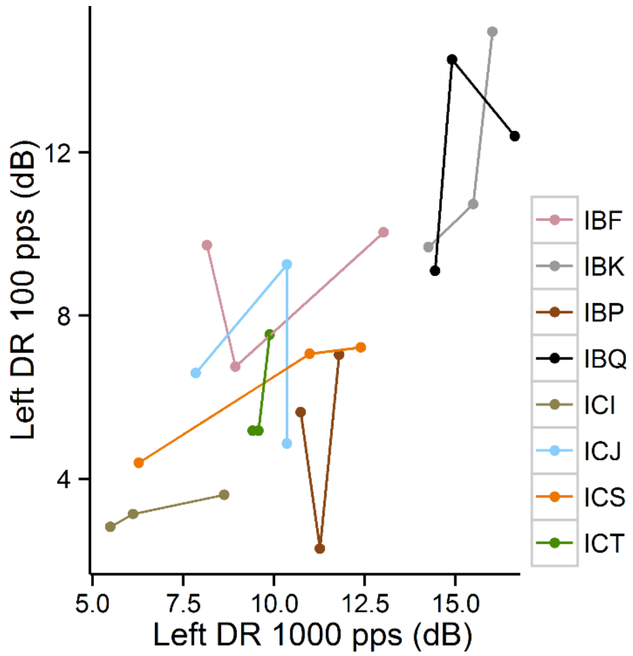

Figure 20. The stimulus DR (dB) at 100 pps as a function of the stimulus DR (dB) at 1000 pps for the left ear. DRs of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant.

Figure 21

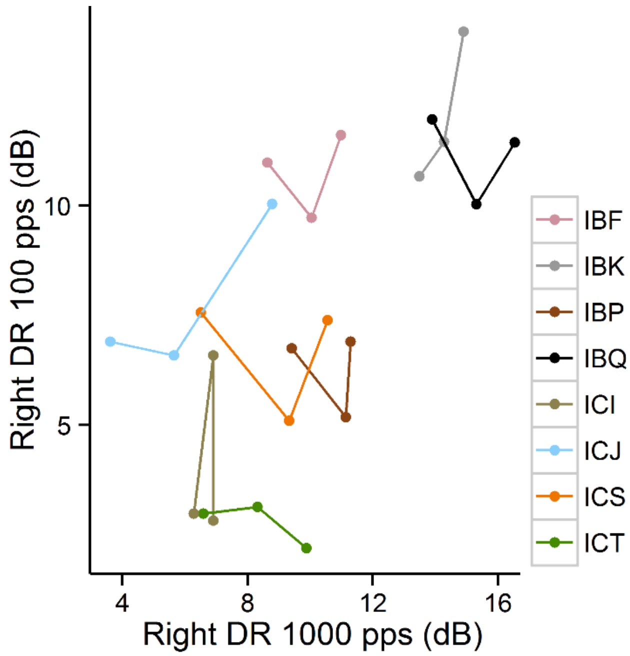

Figure 21. The stimulus DR (dB) at 100 pps as a function of the stimulus DR (dB) at 1000 pps for the right ear. DRs of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant.

Figure 22

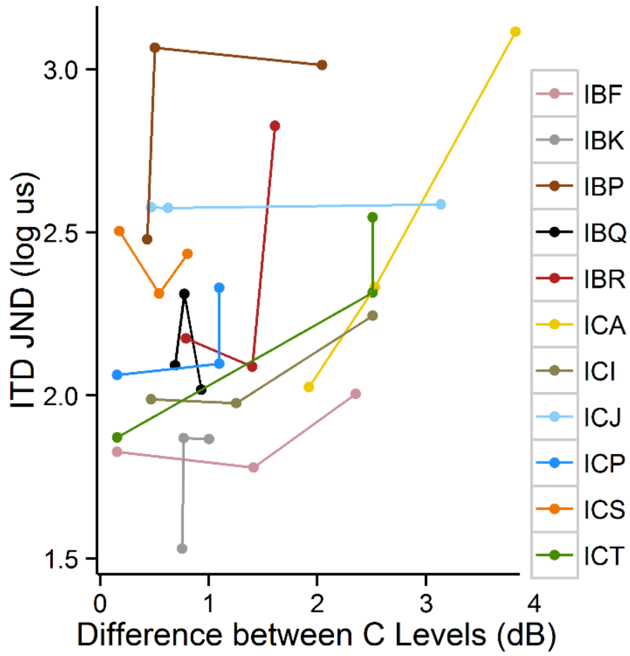

Figure 22. ITD JNDs (in $\log_{10}$ µs) as a function of the difference between C levels (dB) between the left and the right sides. JNDs of 3 places of stimulation per participant are shown and are grouped by color and a conjoining line by participant.

Chapter 4

Summary and Conclusions

The focus of this dissertation was on binaural unmasking for signal detection with temporal envelope information in children and adults with bilateral cochlear implants (CIs). A number of studies have demonstrated that children and adults with CIs show binaural sensitivity with stimulation provided by single-electrode-pairs (Long *et al.*, 2006; Van Deun *et al.*, 2009a; Lu *et al.*, 2010; 2011; Van Deun *et al.*, 2011). However, there is wide variability in performance. Of interest was binaural unmasking with multi-electrode stimulation relative to binaural unmasking with single-electrode stimulation (Chapter 2). This was motivated by the fact that there is variability in performance across stimulation sites and real-world stimuli are conveyed via multi-electrode stimulation. In addition, of interest was the relationship between binaural unmasking and monaural measures of hearing across stimulation sites (Chapter 3).

In Chapter 2 children with CIs showed better dichotic signal detection thresholds than diotic signal detection thresholds (i.e., binaural masking level differences (BMLDs)), similar to the findings of Van Deun *et al.* (2009a). This suggests that children with CIs have binaural processing capabilities, despite the fact that some children do not demonstrate sensitivity to interaural time differences (ITDs) possibly due to early-life deprivation of binaural information. It was suggested that the children with CIs were likely using central processing mechanisms for dichotic signal detection that are also used for processing interaural level differences. Since the children without ITD sensitivity showed BMLDs of comparable size to those of the children who showed ITD sensitivity, it may be that ITD processing is of little benefit for dichotic signal detection with temporal envelope information. That children with CIs show binaural unmasking suggests that they have the potential, like adults with CIs, to show binaural release from masking with speech stimuli given a stimulation strategy that is able to preserve information at the level of the auditory nerve.

In Chapter 2, results showed that children with CIs had diotic and dichotic signal detection thresholds with three-electrode-pair stimulation that were similar to single-electrode-pair thresholds at the site that produced the best thresholds. This finding is promising in that it suggests that the information from better performing sites remains salient in the presence of stimulation at poorer performing sites. Performance with three-electrode-pair stimulation was, however, not found to be better than performance with single-electrode-pair stimulation at the best performing site suggesting little benefit from the presentation of redundant information at separate stimulation sites. It should be noted that for the multi-electrode stimulation the same sample of noise was presented to each of the individual electrodes. Had different samples of noise been used, performance may have been poorer due to energetic masking between the channels in each ear (Lu *et al.*, 2011).

In Chapter 3, dichotic signal detection as well as ITD sensitivity was examined in adults with CIs at each of 3 stimulation sites in relationship to monaural measures made at those stimulation sites in each of the two ears. A relationship was found between dynamic range and dichotic signal detection thresholds as well as between dynamic range and ITDs sensitivity. Specifically, smaller dynamic ranges (at least in the ear that had the smaller dynamic range of the two ears) were associated with poorer dichotic signal detection and poorer ITD sensitivity. Furthermore, greater differences in stimulation levels between the two ears were associated with poorer ITD discrimination. These results may indicate that peripherally the extent of neural survival, tissue growth, and the spread of the excitation pattern affect binaural sensitivity. It was suggested that smaller dynamic ranges represent poorer neural survival or increased tissue growth which would necessitate broader excitation patterns which would likely result in stimulation of fewer corresponding left and right peripheral fibers which deliver input the same central binaural neurons. It was furthermore suggested that differences in stimulation levels reflect a difference in the width or shape of current spread between the ears which could similarly result in reduced stimulation to corresponding left and right peripheral fibers compared to when the spread of current is symmetrical between the ears.

These explanations for the findings in Chapter 3 are reminiscent of the motivation for stimulating at interaurally pitch-matched electrodes i.e., the desire to present information at corresponding locations on the left and right auditory nerves such that information is delivered to the same central binaural units (van Hoesel and Clark, 1997; Long *et al.*, 2003; Kan *et al.*, 2013). However, limitations such as neural survival and different excitation-pattern shapes would not be properly address by changing the electrode that provides stimulation in one of the ears as is done with pitch-matching. Research in the area of neural preservation may be beneficial for increasing the number of neural fibers available for binaural processing (Rejali *et al.*, 2007). Furthermore, current shaping may allow for improvement in the extent to which stimulation is provided to corresponding left and right peripheral fibers (Bonham and Litvak, 2008).

At the level of clinical treatment, if one is to introduce CI mapping procedures and speech processing strategies that attempt to provide binaural information with fidelity, it may be informative to consider participant characteristics such as dynamic range and differences between the arrays of the two ears relative to performance with new interventions. Perhaps certain listeners e.g., those with larger dynamic ranges, could benefit more from bilaterally coordinated input.

In Chapter 3, the relationship between symmetry in loudness growth and dichotic signal detection was also investigated. The relationship between the symmetry of loudness growth between ears and dichotic signal detection was of interest because it was hypothesized that asymmetries could introduce the perception of interaural de-correlatinon in diotic stimuli which would limit signal detection based on interaural de-correlation (the standard in the dichotic signal detection task was diotic noise). Listeners in some cases showed slight inconsistencies between ears in loudness growth. However, no relationship was found been the symmetry of loudness growth and dichotic signal detection. The effect of asymmetries between the ears may be better examined by artificially creating loudness growth asymmetries to measure their effect on dichotic signal detection as well as on lateralization with stimuli which vary in intensity.

In Chapter 3, a relationship was found between the difference in dynamic range between the ears and diotic signal detection. That is, greater differences between the dynamic ranges of the two ears were associated with poorer diotic signal detection thresholds. This finding was not expected and requires further investigation. A comparison between diotic signal detection and left and right monotic signal detection would be helpful to determine whether this finding was due to a type of contralateral masking between the ears or a lack of summation when there are difference between the ears. It must be noted that it is unclear what the relevant differences were between the representations of the signals in each ear, since differences in dynamic range do not necessarily imply differences other than perhaps the number of discriminable intensity steps available to the listener. Instructing the participant to listen to one ear or the other could clarify whether the finding arose because of uncertainty by the listeners of to what information to listen.

In considering the results of Chapters 2 and 3 together two ideas stand out. First, in Chapter 3 it was hypothesized that poor performance in dichotic signal detection and ITD discrimination result from limitations in stimulation of corresponding left and right peripheral fibers. However, the lack of an improvement in dichotic signal detection with multi-electrode stimulation compared to single-electrode stimulation (Chapter 2) suggests that increasing the number of central binaural units which receive bilateral input by increasing the number of stimulation sites, will not necessarily result in better dichotic signal detection. It may be that for listeners with CIs, increasing the number of peripheral fibers on the left and right that result in information *to any one* central binaural processing unit (which would not result from increasing the number of stimulation sites) is more important for improving binaural sensitivity than increasing the number of central binaural units involved in processing. In addition, binaural interference resulting from unilateral stimulation of central binaural units (which would not be reduced from increasing the number of stimulation sites) may play a role in limiting binaural sensitivity (Bernstein, 1991b).

Second, there are likely both central and peripheral factors which limit binaural hearing sensitivity. The explanation that many children with CIs fail to show ITD sensitivity due to central deficits is appealing based on animal models of auditory deprivation and research in other areas suggesting sensitive periods for development (Mayberry and Eichen, 1991; Hancock *et al.*, 2013). The finding in Chapter 2 that children who are not sensitive to ITDs can show large BMLDs provides no evidence of poorer peripheral processing by the children who lack ITD sensitivity. Nevertheless, examination of monaural hearing performance between children with and without ITD sensitivity would help to determine whether any differences between the groups in binaural hearing sensitivity could be explained by peripheral factors.

Ideally, improved understanding of binaural unmasking in listeners with CIs will contribute to the development of interventions which will allow listeners with CIs to reap more of the benefits that having two ears can provide.

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