Perception of amplitude modulation with single or multiple channels in cochlear implant users

John J. Galvin III

Perception of amplitude modulation with single or multiple channels in cochlear implant users

John J. Galvin III

ISBN: 978-90-367-9009-3

Cover art by Lisa Barry

Perception of amplitude modulation with single or multiple channels in cochlear implant users

PhD thesis

to obtain the degree of PhD at the
University of Groningen
on the authority of the
Rector Magnificus Prof. E. Sterken
and in accordance with
the decision by the College of Deans.

This thesis will be defended in public on

Monday 24 October 2016 at 16.15 hours

by

John Galvin

born on 14 January 1963
in Rhode Island, USA

Supervisors
Prof. D. Baskent
Prof. Q.-J. Fu

Assessment Committee
Prof. A. van Wieringen
Prof. C. Lorenzi
Prof. G.J. Verkerke

Table of contents

Chapter 1

General introduction

How a cochlear implant works

Cochlear implants (CIs) have restored hearing sensation to more than 300,000 deaf people worldwide. The CI hardware consists of several components (Fig. 1.1): acoustic sound is picked up by a microphone that is part of a speech processor that analyzes and digitizes the sound and then wirelessly transmits the signal to a subcutaneous receiver that decodes the signal and delivers pulse trains to the electrodes implanted in the scala tympani. Figure 1.1 illustrates hardware components used in most CI systems.

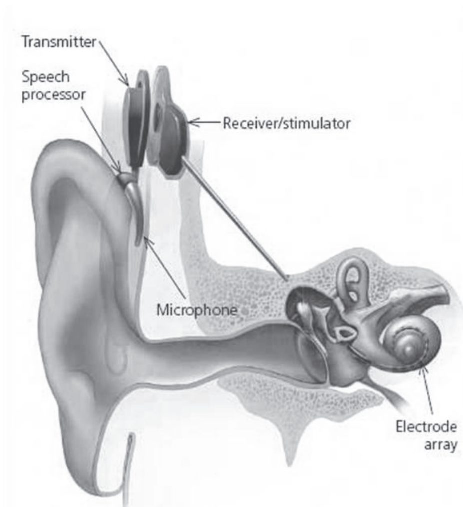

Figure 1.1. Hardware components of a cochlear implant system.

CIs typically use 12-22 electrodes to stimulate the remaining auditory neurons. Despite differences among CI manufacturers' implant designs, signal processing, and the number of electrodes implanted, there are no clear advantages among the different devices. Despite differences in the number of spectral channels provided (12 or more), CI users can typically access only about 8 channels, due to the interference among the implanted electrodes (Friesen et al., 2001). Such "channel interaction" is caused by current field spread and/or by the spread of excitation from stimulated electrodes and greatly limits spatial selectivity along the electrode array. Also, CI users differ greatly in terms of the distribution and health of neural populations. As such, CI users are ultimately limited by the "electrode-neural interface" (i.e., the proximity of electrodes to healthy neurons), rather than the number of electrodes implanted. These two factors – channel interaction and the electrode-neural interface – ultimately limit CI users' functional "spectral resolution." Under ideal listening conditions (e.g., clear speech in quiet), listeners need only 4 spectral channels for good performance (Shannon et al., 1995). However, as the difficulty and complexity of the listening task increases, many more spectral channels are needed, but are unavailable to CI users (Shannon et al., 2004). As such, CI users have difficulty segregating speech from noise, one talker from another, and music perception.

Importance of temporal envelope cues for speech perception in electric hearing

Because of the limited spectral resolution, CI users depend strongly on temporal envelope cues (amplitude changes over time) provided on each channel. Figure 1.2 illustrates basic CI signal processing. In typical CI signal processing, the temporal envelope is extracted from each frequency analysis band and used to modulate pulse trains delivered to each electrode. The extracted temporal envelope is typically low-pass filtered.

Temporal envelope cues can be divided into 3 categories: envelope information (< 50 Hz), which is important for speech segments, periodicity information (50-500 Hz), which is important for voice pitch, speech prosody, etc., and fine structure (500-10,000 Hz), which is important for harmonic pitch (Rosen, 1992).

Most CI signal processing transmits envelope and periodicity information, but not fine-structure information. Envelope information is well represented and perceived. Perception of periodicity information has been shown to interact with theavailable spectral resolution, with temporal cues contributing more as the spectral resolution is reduced (e.g., voice gender recognition in Fu et al., 2004, 2005; vocal emotion recognition in Luo et al., 2007). Perception of periodicity information is also limited by temporal processing, which declines rapidly above 300 Hz (e.g., Shannon, 1992; Fraser and McKay, 2012).

Psychophysical measures of temporal envelope perception

Many previous studies have used amplitude modulation (AM) detection to characterize CI users' temporal processing (e.g., Shannon, 1992; Donaldson and Viemeister, 2000; Fu, 2002; Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et.al., 2007; Fraser and McKay, 2012; Green et al., 201). In an AM detection task, listeners must detect amplitude fluctuations in a stimulus, relative to steady-state stimuli, typically presented at the same reference amplitude. Compared to other measures of temporal processing (e.g., gap detection, pulse rate discrimination, etc.), AM detection has been correlated with speech performance in CI users (Cazals et al., 1994; Fu, 2002) and users of auditory brainstem implants (Colletti and Shannon; 2005). AM detection has been shown to worsen as a function of current level and AM frequency, and carrier stimulation rate (Galvin and Fu, 2005, 2009; Pfingst et al., 2007). AM detection has been correlated with electrode discrimination (Chatterjee and Yu, 2010) and been shown to vary across stimulation site (Pfingst et al., 2007; Zhou and Pfingst, 2012), possibly reflecting neural health across the cochlea.

AM frequency discrimination has also been used to characterize CI users' temporal processing. Different from AM detection, in an AM frequency discrimination task, listeners must discriminate between a reference and probe AM frequency; the AM depth used for AM frequency discrimination is typically well above the discrimination threshold. AM frequency discrimination is typically measured in the periodicity range (i.e., voice pitch). Like AM detection, AM frequency discrimination has been shown to worsen as a function of current level and reference AM frequency. Like AM detection, AM frequency discrimination has been correlated with CI users' speech perception (prosody perception in Chatterjee and Peng, 2008 and Deroche et al., 2012, 2014; tonal language perception in Luo et al., 2008).

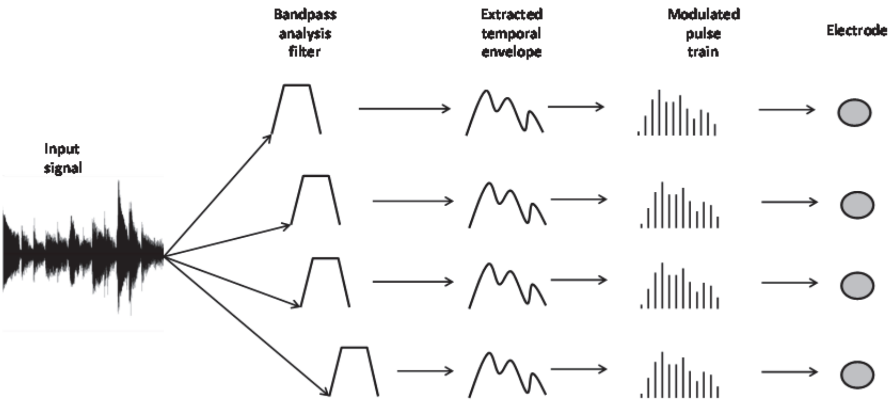

Figure 1.2. Illustration of 4 channels of CI signal processing

These previous studies measured AM detection and frequency discrimination for single electrodes. But in everyday device, CI users receive multi-channel stimulation. One could measure multi-channel perception of speech envelopes directly, but top-down processes related to speech pattern perception may obscure the limits of temporal processing. It is important to know the limits of temporal envelope processing for both single- and multi-channel stimulation in order to improve and/or optimize CI signal processing. Up to now, there have been relatively few studies of CI users' temporal processing. Geurts and Wouters (2001) measured single- and multi-channel AM frequency discrimination, finding better performance with multiple channels than with any of the single component channels. Won et al. (2011) found a correlation between AM detection and speech performance in CI users tested while listening with their clinical processors. Several modulation detection interference (MDI) studies have shown that AM presented on one electrode can interfere with AM detection on another electrode, even when the electrodes are spatially remote (Chatterjee, 2003; Chatterjee and Oba, 2004). Similarly, AM presented on one electrode can interfere with AM frequency discrimination presented on another electrode (Chatterjee and Ozerbut, 2009; Kreft et al., 2013).

While the above studies provide some insight, there are many factors that must be properly controlled to better understand multi-channel temporal envelope processing. One major factor is the effect of multi-channel loudness summation. During clinical fitting of CI speech processors, electrode dynamic ranges (DRs) are typically measured between threshold and comfortable loudness one electrode at a time. When all the electrodes are activated, the thresholds and comfort levels must often be reduced to fit within the CI user's comfortable operating range. Work by Mckay et al. (2001; 2003) has shown significant multi-channel loudness summation that was independent of relative electrode locations. Because single-channel AM detection and frequency discrimination have been shown to depend on current level (Morris and Pfingst, 2000; Donaldson and Viemeister, 2000; Galvin and Fu, 2005, 2009; Luo et al., 2008; Chatterjee and Ozerbut, 2011; Green et al., 2012), the current level reductions needed to accommodate multi-channel loudness summation might adversely affect multi-channel temporal envelope perception. In Geurts and Wouters (2001), there was no explicit control for multi-channel loudness summation; multi-channel stimuli were most likely louder than single-channel stimuli. As such, it is unclear whether the multi-channel advantage in AM frequency discrimination was due to multiple envelope representations or to increased loudness.

To understand the limits of CI users' temporal envelope perception, it is important to have carefully controlled stimuli and experimental design. For many psychophysical measures, it is also important to bypass CI users' clinical processors, which have been optimized for multi-channel speech perception and may or may not reflect CI users' true psychophysical capabilities. Clinical processor components (e.g., microphone, microphone sensitivity, automatic gain control, volume setting, frequency allocation, sharpness of analysis filters, the number of electrodes stimulated within each frame, acoustic-to-electric amplitude mapping, etc.) can greatly distort electrical stimulation pattern relative to the acoustic input. As such, using acoustic stimuli delivered to a CI user's clinical processor may not be the best approach for single- and multi-channel psychophysics. Thus, many previous studies have used research interfaces to directly stimulate electrodes, bypassing CI users' clinical processors. CI research interfaces allow for precise control of all stimulation parameters and selective stimulation of electrodes to be tested. Even with a research interface, it is important to have good stimulus and experimental control. Depending on the research question, reference and probe stimuli might need to be loudness-balanced, level roving might need to be applied to protect against unwanted loudness cues, adaptive or non-adaptive procedures may be preferable for some tasks, etc. With multi-channel stimuli, the need for stimulus control is even greater. Component electrodes should be equally loud, current levels for multi-channel stimuli may need to be reduced to be equally loud as single-channel stimuli, component channels must be optimally interleaved in time, temporal envelopes must be applied to multi-channel stimuli to avoid artifacts that might provide alternative cues, etc. In the research presented here, we used a custom research interface (HEINRI; Wygonski and Robert, 2001) and custom software to deliver all single- and multi-channel stimuli. As such, we believe that the measurements are a good estimate of CI users' temporal envelope perception.

Research questions

Previous studies have shown that single-channel AM detection and frequency discrimination depends strongly on current level, which is a physical dimension (e.g., Donaldson and Viemester, 2000; Chatterjee and Robert, 2001; Galvin and Fu, 2005, 2009; Pfingst et al., 2007; McKay and Henshall, 2010; Chatterjee and Ozerbut, 2011; Green et al., 2012). However, with clinical CI processors, current level must be contextualized according the perceptual dimension of loudness, which relates to physical dimensions of current level as well asthe number of channels and the stimulation rate per channel. As the stimulation rate and/or the number of channels increase, the perceived loudness for a fixed current level will also increase. Because overall loudness must be considered when clinically fitting a CI processor, it is important to consider how temporal envelope perception might be affected by loudness, and not just current level. When current levels must be reduced on component channels to accommodate multi-channel loudness summation and/or multi-pulse integration, it is unclear how individual channels, which may vary considerably in terms of temporal envelope perception, contribute to the multi-channel percept, especially when current levels are reduced. Alternatively, envelope perception may be driven by loudness, whether associated with current level, stimulation rate, and/or the number of channels. Given these many factors to consider, we aimed to answer the following research questions:

1. Does controlling for potential loudness cues associated with the peak level of an AM stimulus affect AM detection? How does such a control interact with the overall level presentation and stimulation rate (each of which contribute to loudness perception)? Figure 1.3 illustrates some of the stimuli used for the single-channel experiments described in Chapter 2 (“A method to dynamically control unwanted loudness cues when measuring amplitude modulation detection in cochlear implant users”). At large AM depths there is greater potential for

potential loudness cues associated with the peak of the AM stimulus. At smaller AM depths, the potential for these loudness cues is lessened. At high overall presentation levels, the AM depth at threshold is typically small, necessitating less compensation for AM peak loudness. At low overall presentation levels, the AM depth at threshold is often high, necessitating greater compensation for AM peak loudness.

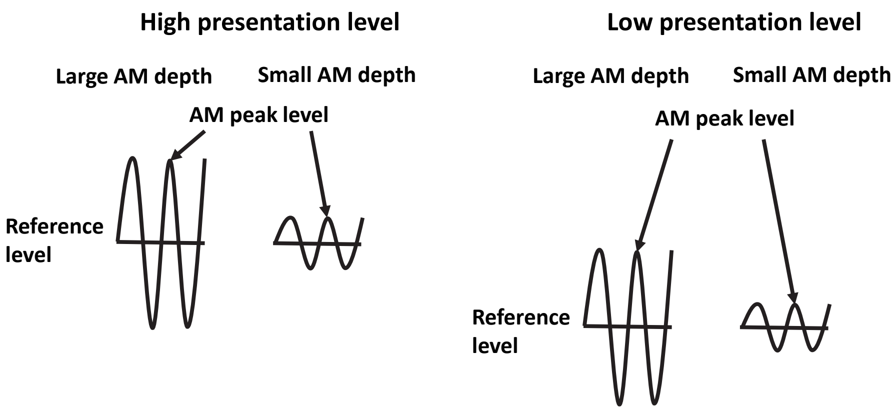

Figure 1.3. Stimuli used to measure AM detection thresholds while controlling for unwanted loudness cues associated with the peak amplitude of the AM stimuli (Chapter 2). The left and right sides show stimuli for a high and low overall presentation levels, respectively. The horizonatal line shows the reference current level for the AM stimuli and the current level for steady non-AM stimuli. During AM detection, the current level of the non-AM stimuli was adjusted to match the loudness of the AM stimuli.

2. How does multi-channel loudness summation affect AM detection? How do individual channels contribute to the multi-channel percept? How does loudness a ffect single- and multi-channel AM detection? Figure 1.4 illustrates stimuli used to measure single- and multi-channel AM detection in Chapter 3 ("Single- and multi-channel modulation detection in cochlear implant users"). All single channels were loudness balanced, and multi-channel stimuli were loudness-balanced to the single-channel stimuli by reducing the current levels on each component channel by the same ratio. In the left side of Figure

1.4, AM detection thresholds vary across single channels. In the right side of Figure 1.4, AM depth was adjusted for all channels by the same amount.

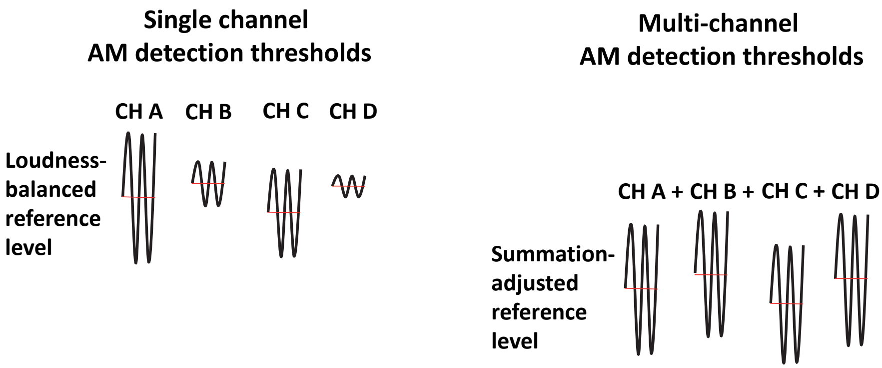

Figure 1.4. Stimuli used to measure single- and multi-channel AM detection thresholds (Chapter 3). The left side shows AM detection thresholds for equally loud single-channel stimuli (note that the absolute current levels differ among single-channel stimuli); thresholds differ across channels. The right side shows AM detection thresholds for a multi-channel stimulus that is equally loud to the single-channel stimuli shown in the left panel. Note that to accommodate multi-channel loudness summation, current levels were reduced by the same ratio on each channel, thus preserving relative loudness across channels. The AM depth was adjusted by the same amount for all channels during testing.

3. Similarly, how does multi-channel loudness summation affect AM frequency discrimination? How does overall loudness affect single- and multi-channel AM frequency discrimination? How does channel spacing affect the multi-channel percept? Figure 1.5 illustrates stimuli used to measure single- and multi-channel AM frequency discrimination in Chapter 4 ("Modulation frequency discrimination with single and multiple channels in cochlear implant users"). The left and middle groups of ovals were of similar loudness, but with different current levels, while the middle and right groups of ovals were of different loudness (the multi-channel stimulus was louder), but

with the same current levels used on each component channel. Note also that the range of modulation (in dB) is the same for each component channel is the same for single- and multi-channel stimuli for all conditions. This manipulation allows overall loudness effects to be compared between single-and multi-channel stimuli and current/loudness effects to be compared within single-channel stimuli.

4. How do individual channels contribute to the multi-channel AM frequency discrimination? How do different envelopes presented to multiple channels interact, and how does channel spacing affect these interactions? Figure 1.5 illustrates example stimuli used to measure multi-channel envelope perception in Chapter 5 ("Envelope interactions in multi-channel amplitude modulation frequency discrimination by cochlear implant users"). In this example, summation-adjusted current levels were used for single- and multi-channel measures; as such, the single-channel were much softer than the multi-channel stimuli. The electrode spacing was varied to compare the effects of channel interaction due to the spread of excitation from each channel (i.e., peripheral contributions). If the spread of excitation was not a factor in performance (i.e., no difference between the wide and narrow spacing conditions), then performance would be due to envelope interaction at a more central auditory processing level. Multi-channel AM frequency discrimination was compared between conditions where the target AM was delivered to one of three or to all three channels to explore how envelope information was combined, and whether some channels contributed more to the percept than others.

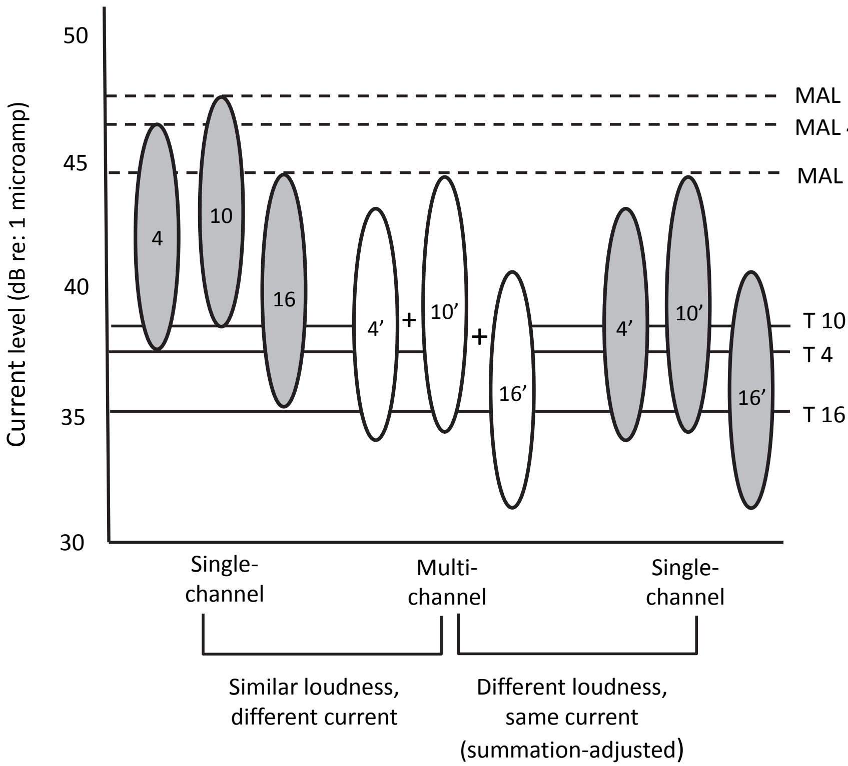

Figure 1.4. Illustration of stimuli used for AM frequency discrimination. The ovals on the left side of the figure show the range of modulation for electrodes 4, 10, and 16 (original single-channel AM stimuli); the solid lines show the original thresholds (T) and the dashed lines show the original maximum acceptable loudness (MAL). Thus, the AM depth was maximal, between T and MAL. These single-channel AM stimuli were similarly loud. The middle group of white ovals shows current levels of the multi-channel AM stimuli after adjusting for multi-channel loudness summation. The right group of ovals shows the same summation-adjusted current levels for single-channel AM stimuli as used for the multi-channel AM stimuli.

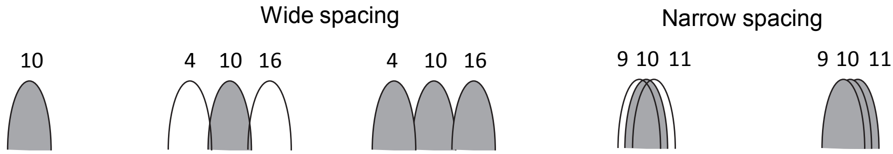

Figure 1.5. Illustration of stimuli used for AM envelope interaction. The numbers indicate which channels were stimulated. The filled and open shapes indicate which channels received the target and reference AM for the probe stimuli; for the reference stimuli, all channels received the reference AM rate (100 Hz). The target AM was delivered to either a single channel, one of three channels, or to all three channels. For the single-channel and one-of-three channel conditions, the target channel was varied to be either the apical, middle (shown above), or basal channel.

Outline of the thesis

In this chapter (Chapter 1), we present a general background of CIs and the perceptual limits of single-channel measures of temporal envelope processing reported in previous studies.

Chapter 2 presents the study “A method to dynamically control unwanted loudness cues when measuring amplitude modulation detection in cochlear implant users.” When measuring AM detection, listeners are typically asked to discriminate among stimuli in which AM is applied to one stimulus and the remaining stimuli are steady-state. The AM depth is varied relative to a reference amplitude, and this same reference amplitude is typically used for the steady-state stimuli. However, the peak amplitude of the AM stimulus will always be higher than that of the steady-state stimuli, allowing for a potential loudness cue that may drive AM detection. To ensure that AM detection reflects the sensitivity to fluctuations in amplitude (rather than to peak amplitude), it is important to control for peak AM loudness when measuring AM detection. Here, we designed and evaluated a method to dynamically control (from trial to trial) for such AM loudness cues when adaptively measuring AM detection thresholds.

Chapter 3 presents the study "Single- and multi-channel modulation detection in cochlear implant users." Single-channel AM detection has shown to worsen as the current level is reduced. In multi-channel stimulation, current levels must often be reduced to accommodate multi-channel loudness summation. It is unclear how these current level reductions might affect multi-channel AM detection. Single-channel AM detection has also been shown to vary across electrodes. It is unclear how differences in single-channel temporal processing might contribute to the multi-channel percept. Here, we compare single-channel AM detection to multi-channel AM detection, with and without the current level adjustments to accommodate multi-channel loudness summation.

Chapter 4 presents the study "Modulation frequency discrimination with single and multiple channels in cochlear implant users." AM frequency discrimination is another important measure of temporal envelope perception, and has been correlated with various speech measures in CI users. Different from AM detection, listeners must discriminate between AM frequencies for temporal envelopes that are well above detection thresholds. Similar to AM detection, single-channel AM frequency discrimination worsens as the current level is reduced. Because current levels must be reduced to accommodate multi-channel loudness summation, it is unclear how these current level reductions might affect AM frequency discrimination. Also, it is unclear whether multi-channel AM frequency discrimination is affected by the distribution of temporal envelope information across the cochlea. Here, we compare AM frequency discrimination with single and multiple channels, with and without the current level adjustments to accommodate multi-channel loudness summation. Coherent AM was applied to all channels in the multi-channel stimuli. We also compare multi-channel AM frequency discrimination for widely and narrowly spaced electrodes.

Chapter 5 presents the study "Envelope interactions in multi-channel amplitude modulation frequency discrimination by cochlear implant users." While AM frequency discrimination may be enhanced when coherent AM is delivered to multiple channels, CI users regularly must process different temporal envelopes delivered to different electrodes in multi-channel stimulation. In both cases, temporal envelope information must be somehow combined across channels. Previous studies have shown that temporal envelope information presented on one electrode can interfere with AM detection or frequency discrimination measured on another electrode. However, these studies did not control for multi-channel loudness summation, which might affect how temporal envelope information is combined. Also, across-site differences in temporal processing might contribute to the interference produced by one electrode onto another. Channel interaction may also affect how temporal envelope information is combined across channels. Here, we measured multi-channel AM frequency discrimination for stimuli in which the target AM was delivered to 1 of 3 channels and the reference AM was delivered to the other 2 channels. The target AM channel was varied across conditions, as was the spacing of electrodes. Data from this study were compared to that of the previous study (Chapter 4) to examine how temporal envelope information is combined across channels when channels contain the same or different envelope information.

Chapter 6 presents a general discussion of Chapters 1-5. In particular, we discuss the implications of loudness summation on measures of multi-channel temporal envelope processing, as well as the effects of channel interaction and across-site variability. We also discuss the importance of strong experimental controls and methods for the research presented here. Finally, we discuss implications for CI signal processing.

Chapter 2

A method to dynamically control unwanted loudness cues when measuring amplitude modulation detection in cochlear implant users

John J. Galvin III
Qian-Jie Fu
Sandy Oba
Deniz Başkent

This chapter is a modified version of a paper published in:
The Journal of Neuroscience Methods (2014); 222:207-212

Abstract

Amplitude modulation (AM) detection is a measure of temporal processing that has been correlated with cochlear implant (CI) users' speech understanding. For CI users, AM stimuli have been shown to be louder than steady-state (non-AM) stimuli presented at the same reference current level, suggesting that unwanted loudness cues might contribute to CI users' AM sensitivity as measured in a modulation detection task. In this paper, a new method is introduced to dynamically control unwanted AM loudness cues when adaptively measuring modulation detection thresholds (MDTs) in CI users.

MDTs were adaptively measured in 9 CI subjects using a three-alternative, forced-choice procedure, with and without dynamic control of unwanted AM loudness cues. To control for AM loudness cues during the MDT task, the level of the steady-state (non-AM) stimuli was increased to match the loudness of the AM stimulus using a non-linear amplitude scaling function, which was obtained by first loudness-balancing non-AM stimuli to AM stimuli at various modulation depths. To further protect against unwanted loudness cues, ±0.75 dB of level roving was also applied to all stimuli during the MDT task.

Absolute MDTs were generally poorer when unwanted AM loudness cues were controlled. However, the effects of modulation frequency and presentation level on modulation sensitivity were fundamentally unchanged by the availability of AM loudness cues. Conclusions: The data suggest that the present method controlling for unwanted AM loudness cues might better represent CI users' MDTs, without changing fundamental effects of modulation frequency and presentation level on CI users' modulation sensitivity.

Introduction

Amplitude modulation (AM) detection is one of the few psychophysical measures shown to predict speech understanding by cochlear implant (CI) users (Cazals et al., 1994; Fu, 2002; Won et al., 2011). For studies with direct stimulation via research interfaces, various stimulation parameters have been shown to affect modulation detection thresholds (MDTs), including stimulation level, modulation frequency, and stimulation rate (Shannon, 1992; Donaldson and Viemeister, 2000: Fu, 2002; Chatterjee and Oba, 2005; Colletti and Shannon, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007; Luo et al., 2008; Garadat et al., 2012).

One potential issue with some of these studies is that loudness cues associated with dynamic stimuli were not adequately or consistently controlled. As such, it is difficult to know whether MDTs measured in previous studies were influenced by sensitivity to AM loudness cues or to sensitivity to the temporal envelope (i.e., changes in amplitude over time). Given a fixed reference amplitude, the peak amplitude of an AM stimulus will be higher (and possibly louder) than the peak of a steady-state (non-AM) stimulus. McKay and Henshall (2010) found that CI users perceived AM stimuli to be louder than non-AM stimuli with the same average current level. At equal loudness, mean current levels (across subjects) for non-AM stimuli were found to be between the peak and average current levels of the AM stimuli. Accordingly, the authors argued that it might be necessary to control AM loudness cues when measuring CI users' modulation detection. If AM loudness cues are not adequately controlled, MDTs may reflect listeners' sensitivity to the peak amplitude of the AM signal (similar to an increment detection task), rather than the changes amplitude over time. Recent studies by Chatterjee and Ozerbut (2011), Green et al. (2012), and Fraser and McKay (2012) have attempted to control for these potential loudness cues in various ways, with somewhat inconsistent results.

Chatterjee and Ozerbut (2011) found markedly smaller current level differences between equally-loud AM and non-AM stimuli for modulation depths <16%, compared with McKay and Henshall (2010). The authors also measured MDTs with and without some control of loudness cues. Increasing amounts of level roving applied to all stimuli significantly worsened mean MDTs, but did not change the slope of the temporal modulation transfer function (TMTF). Although a few subjects exhibited sensitivity to loudness cues in AM, most did not. The authors argued that such level roving seemed only to add "noise" to the modulation detection task, but did not fundamentally change the effects of stimulation level and modulation frequency on MDTs.

Fraser and McKay (2012) combined level roving (±0.75 dB, i.e., ±4 clinical units) with level compensation for AM loudness cues; the level roving was added to address potential loudness imbalances (Dai and Micheyl, 2010). Non-AM and AM stimuli (at various modulation depths) were first loudness-balanced at different stimulation rates and levels. Loudness balancing results were similar to those of McKay and Henshall (2010) and Chatterjee and Ozerbut (2011), in that the amount of non-AM level compensation increased with AM modulation depth. Different from McKay and Henshall (2010), Fraser and McKay (2012) found that at equal loudness, non-AM current levels were closer to AM peak levels than to average current levels. The loudness-balanced AM and non-AM stimuli were used for modulation detection using a (non-adaptive) method of constant stimuli. With the level compensation and roving, the effects of modulation frequency and presentation level were similar to those from previous studies that did not control for AM loudness cues (Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007): MDTs worsened with increasing modulation frequency and decreasing presentation level. In a few conditions and subjects, MDTs also were collected without the level compensation and roving. For these few cases reported, MDTs were better without the level compensation and roving, suggesting that CI users were indeed sensitive to AM loudness cues when detecting AM.

AM loudness cues were not controlled in many previous modulation detection studies (Shannon, 1992; Donaldson and Viemeister, 2000: Fu, 2002; Chatterjee and Oba, 2005; Colletti and Shannon, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007; Luo et al., 2008; Garadat et al., 2012). Other studies seem to offer inconsistent and/or incomplete pictures regarding the effect of AM loudness cues on modulation detection by CI users. Chatterjee and Ozerbut (2011) compared MDTs with and without level roving only. Green et al. (2012) measured MDTs with level roving, but not without. Fraser and McKay (2012) combined level roving and AM loudness compensation, but only compared MDTs without the roving/compensation in a few conditions; also Fraser and McKay used a method of constant stimuli. None had implemented control for AM loudness cues within an adaptive modulation detection procedure, a common method used to measure MDTs in CI listeners. Given that MDTs have been significantly correlated with CI and ABI speech performance (Cazals et al., 1994; Fu, 2002; Colletti and Shannon, 2005), it is important to know how these AM loudness cues might affect CI users' modulation detection.

In this study, MDTs were adaptively measured with and without a novel method to dynamically control AM loudness cues. During the adaptive MDT task, the level of non-AM stimuli was dynamically adjusted to match the loudness of AM stimuli, followed by global level-roving of all stimuli. Thus, the new adaptive method was different from the method of constant stimuli used by Fraser and McKay (2012), and different from Chatterjee and Ozerbut (2011) and Green et al. (2012) in that AM loudness compensation and level roving were combined within the adaptive modulation detection task. By adjusting the level of the non-AM stimulus to match the loudness of the modulation depth during the adaptive procedure, listeners must primarily attend to the temporal envelope of the AM stimulus.

Methods

Participants

Nine adult, post-lingually deafened CI users participated in this experiment. All had more than 2 years of experience with their implant device. Relevant subject details are shown in Table 2.1; subjects S1, S2 and S5 participated in the Galvin and Fu (2009) study. All subjects provided informed consent in accordance with the guidelines of the local Institutional Review Board, and all were financially compensated for their participation.

Subject	Gender	Age at Testing (yrs)	CI experience (yrs)	Duration of deafness (yrs)	Device	Electrode (Mode)
S1	F	77	10	12	N-24	17 (MP1+2)
S2	F	67	7	20	N-24	14 (MP1+2)
S3	M	81	15	1	N-22	14 (BP+1)
S4	F	78	23	14	Freedom	15 (MP1+2)
S5	M	70	21	4	N-22	14 (BP+1)
S6	F	58	17	20	N-22	15 (BP+1)
S7	F	28	5	5	Freedom	14 (MP1+2)
S8	F	66	7	24	Freedom	14 (MP1+2)
S9	M	74	3	2	Freedom	14 (MP1+2)

Table 2.1. CI subject demographic information.

Stimuli

All stimuli were 300-ms biphasic pulse trains. The pulse phase duration was 100 μs; the inter-phase gap was 20 μs; note that these values are larger than typically used in the ACE strategy, but were necessary to obtain adequate loudness for subjects who used BP+1 stimulation mode. The test electrode was generally located in the middle-apical region of the cochlea, similar to Fu (2004). Table 2.1 lists the test electrodes and stimulation mode for each subject. The stimulation rate was 500 or 2000 pulses per second (pps), spanning the range of rates typically used in clinical processors. The stimulation levels were referenced to 25% or 50% of the dynamic range (DR) of the 500 pps stimulus. The relatively low and high presentation levels were selected because MDTs have been shown to be level-dependent in many previous studies (Donaldson and Viemeister, 2000; Fu, 2002; Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007). The modulation frequency was 10 Hz or 100 Hz, as MDTs generally worsen with increasing modulation frequency, up to ~300 Hz (Shannon, 1992; Fraser and McKay, 2012; Green et al., 2012).

Sinusoidal AM was applied as a percentage of the carrier pulse train amplitude according to:

$$[f(t)][1 + m\sin(2 * \pi * fm * t)]$$

where *f(t)* is a steady-state pulse train, *m* is the modulation index, and *fm* is the modulation frequency. All stimuli were presented via research interface (Wygonski and Robert, 2001), bypassing CI subjects' clinical speech processors and settings.

Loudness balancing across stimulation rates

DRs were estimated for the 500 pps and 2000 pps stimuli, presented without modulation (non-AM). Absolute detection thresholds were estimated according to the "counting" method commonly used for clinical fitting. In the counting method, a number of 300-ms pulse trains were presented to the subject. If the subject correctly identified the number of beeps, the current level was reduced. If the subject incorrectly identified the number of beeps, the current level was increased. The initial step size for adjustments was 5 clinical units (CUs) and the final step size was 2 CUs. The current level after six reversals was taken to be the detection threshold. Maximum acceptable loudness (MAL) levels, defined as the "loudest sound that could be tolerated for a short time," were estimated by slowly increasing the current level until reaching MAL. Threshold and MAL levels were averaged across of a minimum of two runs, and the DR was calculated as the difference in current (in microamps) between MAL and threshold.

Stimuli (non-AM) were loudness balanced using an adaptive two-alternative, forced-choice (2AFC), double-staircase procedure (Jesteadt 1980). Reference stimuli were 500 pps, presented at 25% or 50% DR. The current amplitude of the 2000 pps stimulus was adjusted according to subject response (2-down/1-up or 1-down/2-up, depending on the track). During each trial, the subject would hear two intervals, one which contained the 500 pps reference and the other which contained the 2000 pps probe. The subject was asked to pick which interval was louder, ignoring all other sound qualities (e.g., pitch). For each run, the final 8 of 12 reversals in current amplitude were averaged, and the mean of 2-4 runs was considered to be the loudness-balanced level. In almost all cases, 2 runs were averaged to determine the loudness-balanced level. In cases where the loudness-balanced level differed by 1 dB or more (S2: 25% DR; S5: 25% DR, 50% DR; S8: 25% DR, 50% DR), 2 more runs were performed. In this paper, the low and high presentation levels are referred to as the 25 loudness-balanced level (LL) and 50 LL, respectively. Thus, MDTs were measured at equally loud levels across stimulation rates and modulation frequencies.

Modulation detection

MDTs were measured using an adaptive, 3AFC procedure. The modulation depth was adjusted according to subject response (3-down/1-up), converging on MDT that corresponded to 79.4% correct (Levitt, 1971). One interval (randomly assigned) contained the AM stimulus and the other two intervals contained non-AM stimuli. Subjects were asked to indicate which interval was different (ignoring the difference in loudness). For each run, the final 8 of 12 reversals in AM depth were averaged to obtain the MDT; 3-6 test runs were conducted for each experimental condition.

Method for dynamically controlling unwanted AM loudness cues

For each stimulation rate, modulation frequency, and presentation level condition, MDTs were measured with and without control for unwanted AM loudness cues. To control for loudness cues within each trial, two current level adjustments were made across stimuli: 1) Upward adjustment to the level of non-AM stimuli to compensate for the loudness of AM stimuli, and 2) Level roving across all stimuli (to address potential inaccuracies in loudness balancing and to further reduce loudness cues).

To determine how much non-AM level compensation was required for AM loudness, non-AM stimuli were first loudness-balanced to AM stimuli using an adaptive, 2AFC, double-staircase procedure (Jesteadt, 1980), similar to methods used by Chatterjee and Ozerbut (2011) and Fraser and McKay (2012). During loudness-balancing, the AM stimulus served as the reference. To cover the range of stimulation rates, modulation frequencies, and presentation levels to be tested during modulation detection, four AM reference conditions were tested: 1) 500 pps, 10 Hz, 25 LL, 2) 500 pps, 100 Hz, 50 LL, 3) 2000 pps, 100 Hz. 25 LL, and 4) 2000 pps, 10 Hz, 50 LL. Within these four AM reference conditions, AM depths were 5%, 10%, 20%, or 30%. The current amplitude of non-AM stimulus was adjusted according to subject response (2-down/1-up or 1-down/2-up, depending on the track). For each run, the final 8 of 12 reversals in current amplitude were averaged, and the mean of 2-4 runs was considered to be the current level needed to equate the loudness of the non-AM stimulus to that of the AM stimulus. In almost all cases, 2 runs were averaged to determine the loudness-balanced level. In cases where the loudness-balanced level differed by 1 dB or more (S4: 25 LL/10 Hz; S8: 25% DR/10 Hz, 50% DR/100 Hz), 2 more runs were performed.

Exponential fits were applied to the loudness balance data (averaged across conditions). For individual subjects, the amount of level compensation y (in dB) was dynamically adjusted during the MDT task according to:

$$y = 20\ x\ log_{10}\ \frac{(1 + m)}{(1 + \alpha m)}$$

where m is the modulation index of the modulated stimulus and α is the exponent (ranging from 0 to 1) of the exponential function fit to each subject's AM vs. non-AM loudness-balance data. Thus, during each trial of the modulation detection task, the level of the non-AM stimulus was upwardly adjusted by y dB to match the loudness of the AM stimulus at the target modulation depth according to each subject's loudness-balancing data. After applying this level compensation to the non-AM stimuli, the current level of each stimulus in each trial was independently roved by a random value between - 0.75 and 0.75 dB (±4 clinical units) as in Fraser and McKay (2012). Level roving was applied to all stimuli to further reduce any residual loudness differences between AM and non-AM stimuli that may not have been addressed by the loudness balancing. MDTs were also measured without controlling for loudness cues, as in many previous studies (e.g., Shannon, 1992, Donaldson and Viemeister, 2000; Galvin and Fu, 2005, 2009; Pfingst et al. 2007).

Results

Loudness balancing

At equal loudness, the mean current level difference between 500 pps and 2000 pps non-AM stimuli was 3.29 and 2.73 dB for 25 LL and 50 LL, respectively. Current level differences at equal loudness across rates were quite variable across subjects, ranging from 0.48 dB (S5, 50 LL) to 4.95 dB (S7, 25 LL). A one-way repeated measures analysis of variance (RM ANOVA) showed no significant effect of presentation level (25 LL or 50 LL) on current level differences between equally loud 500 pps and 2000 pps non-AM stimuli [$F(1,8)=2.398$, $p=0.160$].

Figure 2.1 shows exponential fits to the non-AM vs. AM loudness balance data for individual subjects. These functions were eventually used to dynamically adjust the level of the non-AM stimuli to match the loudness of the AM stimulus during the modulation detection task. For each subject, the slope of the fits was averaged across the 4 AM reference conditions. The slope (a) of the fits (listed in the legend of Fig. 1) was variable across subjects, reflecting differences in sensitivity to AM loudness. Slopes for some subjects (S5 and S9) were close to the peak level of AM, and for others were midway between the reference and peak level of AM (S4 and S9). The data were well fit by the functions, as reflected by the high r2 values.

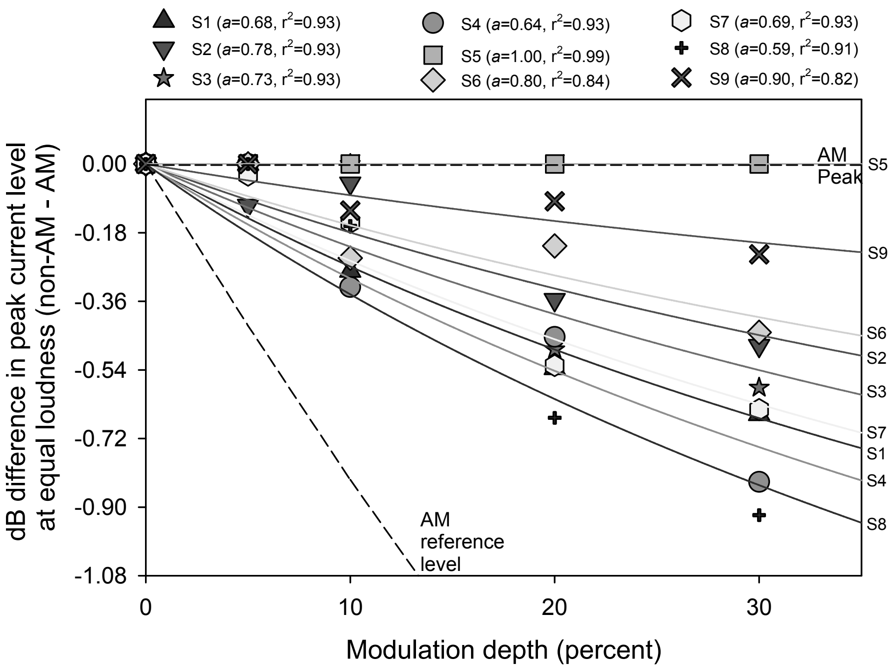

Figure 2.1. Non-linear fits to loudness-balance data between AM and non-AM stimuli, as a function of modulation depth. Data were fit according to Eq. 1 (see Methods). The slope (*a*) and goodness of fit (r^2) for the functions are listed next to individual subject symbols in the legend. The top dashed line shows the difference between AM and non-AM loudness in terms of average current level and the bottom dashed line shows the difference in terms of peak level. Each y-axis tic is equivalent to 1 clinical unit in the Nucleus CI device.

Modulation detection

Figure 2.2 shows mean MDTs (across subjects) with and without control for AM loudness cues. With the 500 pps stimulation rate, MDTs were consistently poorer when AM loudness cues were controlled. With the 2000 pps stimulation rate, controlling for AM loudness cues had a much smaller effect. A multi-way RM ANOVA showed significant main effects for presentation level [F(1,8)=13.053, p=0.007], modulation frequency [F(1,8)=23.777, p=0.001], and controlling for AM loudness cues[F(1,8)=10.704, p=0.011], but not for stimulation rate [F(1,8)=4.537, p=0.066]. There were significant interactions between modulation frequency and controlling for AM loudness cues [F(1,8)=8.960, p=0.017] and among modulation frequency, stimulation rate, and controlling for AM loudness cues [F(1,8)=10.413, p=0.012].

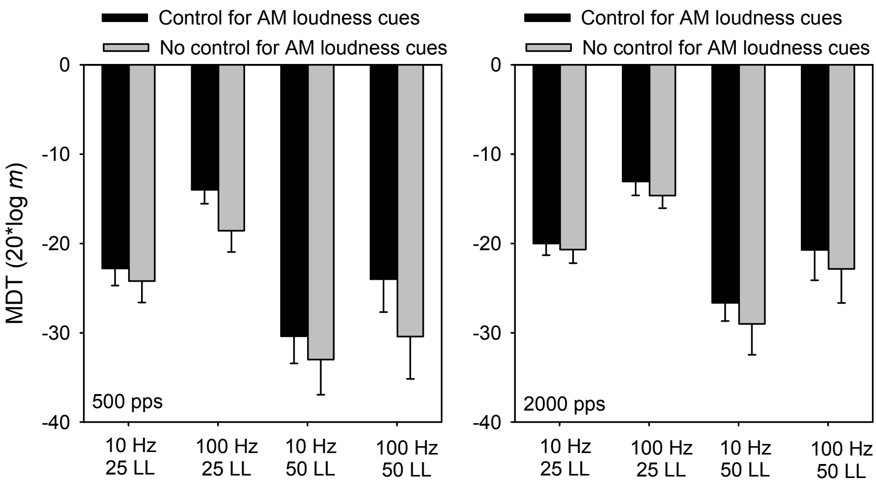

Figure 2.2. Mean MDTs (across subjects) as a function of modulation frequency and stimulation level conditions. The black and gray bars show data with and without control for unwanted AM loudness cues, respectively. The asterisks show significant differences (paired t-tests, p<0.05). The error bars show the standard error. The left and right panels show data for the 500 and 2000 pps carrier rates, respectively.

Discussion

The present method appears to be appropriate for controlling unwanted AM loudness cues when measuring modulation detection by CI users. Different from the simple level roving used by Chatterjee and Ozerbut (2011) and Green et al. (2012) when adaptively measuring MDTs, the present method incorporated an AM loudness adjustment. Different from the method of constant stimuli used by Fraser and McKay, the present method incorporated level roving and AM loudness adjustment within an adaptive procedure, which is most commonly used when measuring MDTs. Controlling for AM loudness cues generally increased absolute MDTs, but did not fundamentally change the effects of modulation frequency and presentation level on modulation sensitivity.

With or without controlling for AM loudness cues, MDTs improved as the presentation level increased and as the modulation frequency was reduced, consistent with previous studies (Pfingst et al., 2007; Galvin and Fu, 2009). Controlling for AM loudness cues significantly interacted with the effect of stimulation rate on MDTs, possibly due to small and/or inconsistent differences in MDTs across stimulation rates. This suggests that previous findings (Galvin and Fu, 2005, 2009; Pfingst et al., 2007) regarding the effect of stimulation rate on MDTs might have been influenced by AM loudness cues.

AM stimuli were consistently louder than non-AM stimuli with the same reference amplitude, consistent with previous studies (McKay and Henshall 2010; Chatterjee and Ozerbut 2011; Fraser and McKay 2012). For the present loudness balance data, adjustments to non-AM current levels were closer to the AM peak amplitude than to the AM reference amplitude, consistent with Fraser and McKay (2012), but different from McKay and Henshall (2010), who found non-AM current levels closer to average than to peak current levels of equally loud AM stimuli. This difference might be due to the lower presentation levels and lower modulation frequencies used in the present study than in McKay and Henshall (2010).

There was a wide variability in subjects' perception of AM loudness, as reflected by the different AM loudness fits in Figure 2.1. Peak level differences between equally loud non-AM and AM stimuli were as large as -1.57 dB (i.e., nearly 16 clinical units less than the peak AM level), but mostly were close to the peak AM level. Differences across subjects' AM loudness judgments might reflect individual differences in loudness integration. As such, loudness balancing might be necessary for tasks in which loudness cues could influence perception, such as modulation detection and pulse rate discrimination. In such cases, simple level roving (as is sometimes done) might not be adequate because given a fixed reference level and any amount of level roving, AM stimuli would remain louder than non-AM stimuli, on average. Too much level roving might simply make the task too difficult, as suggested by Chatterjee and Ozerbut (2011). By first compensating for the loudness of the AM stimuli, and then roving by a relatively small amount, MDTs may be measured without consistent loudness cues that could influence modulation detection. Whether elevated MDTs were due to controlling loudness cues or due to introducing greater uncertainty in level roving is not possible to know given the present study. Further studies may wish control for loudness cues or rove the level independently to isolate their effects on MDTs. It is likely that the present elevated MDTs at small modulation depths may have been more due to the level roving, as the AM loudness cues at those depths would have been quite small. It may also be preferable in future studies to rove only the level of the non-AM intervals, as MDTs have been shown to be very level dependent (Donaldson and Viemeister, 2000: Chatterjee and Oba, 2005; Colletti Galvin and Fu, 2005, 2009; Pfingst et al., 2007). In the present study, the level of the AM signal was roved from trial to trial, which may have resulted in unwanted changes in modulation sensitivity during the test run.

In summary, this study presented a novel method to dynamically adjust the level of non-AM stimuli to compensate for unwanted AM loudness cues during an adaptive modulation detection task. On average, controlling for AM loudness cues significantly worsened absolute modulation sensitivity, but did not fundamentally change the effects of modulation frequency and presentation level on MDTs. Thus, findings from many previous CI modulation studies (Shannon, 1992; Donaldson and Viemeister, 2000: Fu, 2002; Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007) would remain fundamentally true, albeit with possibly elevated absolute MDTs. Different from previous studies (Galvin and Fu, 2005, 2009; Pfingst et al., 2007), there was no significant difference in MDTs between the 500 pps and 2000 pps stimulation rates when AM loudness cues were controlled. The present data suggest that controlling for AM loudness cues might better represent CI users' limits to temporal processing, as measured with an adaptive modulation detection task.

Acknowledgments

We thank all implant subjects for their participation, Joseph Crew for help with data collection, and Monita Chatterjee, David Landsberger, David Bakhos, Bob Shannon, and Justin Aronoff for helpful comments. Work supported by NIH grant DC004993. Dr. Baskent was supported by VIDI grant 016.096.397 from the Netherlands Organization for Scientific Research (NWO) and the Netherlands Organization for Health Research and Development (ZonMw), and a Rosalind Franklin Fellowship from University of Groningen, University Medical Center Groningen.

Chapter 3

Single- and multi-channel modulation detection in cochlear implant users

John J. Galvin III
Sandy Oba
Qian-Jie Fu
Deniz Başkent

This chapter is a modified version of a paper published in:
PLOS ONE (2014); 9(6):e99338

Abstract

Single-channel modulation detection thresholds (MDTs) have been shown to predict cochlear implant (CI) users' speech performance. However, little is known about multi-channel modulation sensitivity. Two factors likely contribute to multi-channel modulation sensitivity: multi-channel loudness summation and the across-site variance in single-channel MDTs.

In this study, single- and multi-channel MDTs were measured in 9 CI users at relatively low and high presentation levels and modulation frequencies. Single-channel MDTs were measured at widely spaced electrode locations, and these same channels were used for the multi-channel stimuli. Multi-channel MDTs were measured twice, with and without adjustment for multi-channel loudness summation (i.e., at the same loudness as for the single-channel MDTs or louder).

Results showed that the effect of presentation level and modulation frequency were similar for single- and multi-channel MDTs. Multi-channel MDTs were significantly poorer than single-channel MDTs when the current levels of the multi-channel stimuli were reduced to match the loudness of the single-channel stimuli. This suggests that, at equal loudness, single-channel measures may over-estimate CI users' multi-channel modulation sensitivity. At equal loudness, there was no significant correlation between the amount of multi-channel loudness summation and the deficit in multi-channel MDTs, relative to the average single-channel MDT. With no loudness compensation, multi-channel MDTs were significantly better than the best single-channel MDT. The across-site variance in single-channel MDTs varied substantially across subjects. However, the across-site variance was not correlated with the multi-channel advantage over the best single channel. This suggests that CI listeners combined envelope information across channels instead of attending to the best channel.

Introduction

Temporal amplitude modulation (AM) detection is one of the few psychophysical measures that have been shown to predict speech perception by users of cochlear implants (CIs) (Cazals et al., 1994; Fu, 2004) or auditory brainstem implants (Colletti and Shannon, 2005). Various stimulation parameters have been shown to affect modulation detection thresholds (MDTs) measured on a single electrode, including current level, modulation frequency, and stimulation rate (Shannon, 1992; Busby et al., 1993; Donaldson and Viemester, 2000; Chatterjee and Robert, 2001; Fu, 2004; Galvin and Fu, 2005, 2009; Pfingst et al., 2007; Arora et al., 2011; Chatterjee and Oberzut, 2011; Green et al., 2012; Fraser and McKay, 2012). In these single-channel modulation detection studies, MDTs generally improve as the current level is increased and as the modulation frequency is reduced. However, given that nearly all CIs are multi-channel, it is crucial to characterize multi-channel MDTs and their relation to the single-channel MDTs.

One factor that may affect multi-channel temporal processing is loudness summation. Clinical CI speech processors are generally fitted with regard to loudness (i.e., between barely audible and the most comfortable levels), and adjustments are often necessary to accommodate multi-channel loudness summation. As such, current levels on individual channels may be lower when presented in a multi-channel context compared to those when measured in isolation. Because MDTs are level-dependent (Shannon, 1992; Donaldson and Viemester, 2000; Fu, 2004; Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007), modulation sensitivity on individual channels may be poorer after adjusting for multi-channel loudness summation. Another factor that may affect multi-channel temporal processing is across-site variability in single-channel modulation sensitivity. Garadat et al. (2012) showed significant variability in single-channel MDTs across stimulation sites within and across CI subjects. It is unclear how single-channel across-site variability may contribute to multi-channel modulation sensitivity. These two factors – loudness summation and across-site variability – may combine in some way such that CI users may attend to the channels with the best modulation sensitivity, but at lower current levels after adjusting for summation. Alternatively, CI users may combine temporal information from all channels when detecting modulation with multiple channels.

While single-channel temporal processing has been extensively studied, there are relatively few studies regarding multi-channel temporal processing. Geurts and Wouters (2001) measured single- and multi-channel AM frequency detection in CI users. They found that AM frequency detection was improved with multi-channel stimulation, relative to single-channel performance. However, no adjustment was made for multi-channel loudness summation. Chatterjee (2003) and Chatterjee and Oba (2005) measured modulation detection interference (MDI) by fluctuating maskers in CI subjects. They found significant MDI, even when the maskers were spatially remote from the target, suggesting that CI users combined temporal information across distant neural populations (i.e., more central processing of temporal envelope information). Although their results supported the notion that central processes mediate envelope interactions, they did not find evidence for modulation tuning of the sort observed in normal-hearing (NH) listeners (Dau et al., 1997ab). Kreft et al. (2013) measured AM frequency discrimination in NH and CI listeners in the presence of steady-state and modulated maskers that were spatially proximate or remote to the target; the maskers were presented with or without a temporal offset relative to the target. Similar to the MDI findings by Chatterjee and colleagues, Kreft et al. (2013) found significant interference by modulated maskers, but with some effect of masker location; temporal offset between the masker and target did not significantly reduce interference. These previous studies present some evidence that central mechanisms result in combinations of and interactions between envelopes on remote spatial channels.

In this study, single- and multi-channel MDTs were measured in 9 CI subjects. MDTs were measured at relatively low and high presentation levels, and at low and high modulation frequencies. Single-channel MDTs were measured at 4 maximally spaced stimulation sites to target spatially remote neural populations, which would presumably result in greater across-site variability than with 4 closely spaced electrodes. Multi-channel MDTs were measured using the same electrodes used to measure single-channel MDTs. To explore the effects of loudness summation on multi-channel modulation sensitivity, multi-channel MDTs were measured with and without adjustment for multi-channel loudness summation.

Methods

Participants

Nine adult, post-lingually deafened CI users participated in this experiment. All were users of Cochlear Corp. devices and all had more than 2 years of experience with their implant device. Relevant subject details are shown in Table 3.1. All subjects previously participated in a related study (Galvin et al., 2013). All subjects provided written informed consent prior to participating in the study, in accordance with the guidelines of the St. Vincent Medical Center Institutional Review Board (Los Angeles, CA), which specifically approved this study. All subjects were financially compensated for their participation.

Subject	Gender	Age at testing (yrs)	CI exp (yrs)	Dur deafness (yrs)	Device	Stim mode	Experimental electrodes [1]			
							A	*B*	*C*	*D*
S1	F	77	10	12	N-24	MP1+2	8	12	17	22
S2	F	67	7	20	N-24	MP1+2	2	8	14	20
S3	M	81	15	1	N-22	BP+1	2	8	14	20
S4	F	78	23	14	Freedom	MP1+2	3	9	15	21
S5	M	70	21	4	N-22	BP+1	2	8	14	20
S6	F	58	17	20	N-22	BP+1	5	10	15	20
S7	F	28	5	5	Freedom	MP1+2	2	8	14	20
S8	F	66	7	24	Freedom	MP1+2	2	8	14	20
S9	M	74	3	2	Freedom	MP1+2	2	8	14	20

Table 3.1. CI subject demographic information. The experimental electrode used as the reference for loudness-balancing in shown in column C. CI exp = experience with cochlear implant device; Dur deafness = duration of diagnosed severe-to-profound deafness; Stim mode = stimulation mode; MP1+2 = intracochlear monopolar stimulation with two extracochlear grounds; BP+1 = intracochlear bipolar stimulation with active and return electrode separated by one electrode.

Single-channel Modulation Detection Thresholds (MDTs)

Stimuli

All stimuli were 300-ms biphasic pulse trains. The pulse phase duration was 100 μs; the inter-phase gap was 20 μs; note that these values are larger than typically used in the ACE strategy, but were necessary to obtain adequate loudness for subjects who used BP+1 stimulation mode. Four test electrodes were selected and assigned to channel locations that spanned the electrode array from the base (A) to the basal-middle (B) to the middle-apical (C) to the apex (D). Electrodes were selected to maintain the maximum distance between active electrodes within each subject's device; because all electrodes were not active for some subjects, the specific electrodes for each channel were different for some subjects (e.g., S1, S4, and S6). Table 3.1 lists the test electrode, channel assignment and stimulation mode for each subject. The stimulation rate was 500 pulses per second (pps). The presentation level was referenced to 25% or 50% of the dynamic range (DR) of a 500 pps stimulus. The modulation frequency was 10 Hz or 100 Hz.

Sinusoidal AM was applied as a percentage of the carrier pulse train amplitude according to:

$$[f(t)][1 + m\sin(2 * \pi * fm * t)]$$

where *f(t)* is a steady-state pulse train, *m* is the modulation index, and *fm* is the modulation frequency. All stimuli were presented via research interface (Wygonski and Robert, 2002), bypassing CI subjects' clinical speech processors and settings.

Dynamic range (DR) estimation

DRs were estimated for all single-channel stimuli, presented without modulation (non-AM). Absolute detection thresholds were estimated according to the "counting" method commonly used for clinical fitting. Maximum acceptable loudness (MAL) levels, defined as the "loudest sound that could be tolerated for a short time," were estimated by slowly increasing the current level until reaching MAL. Threshold and MAL levels were averaged across a minimum of two runs, and the DR was calculated as the difference in current (in microamps) between MAL and threshold.

Loudness balancing

The four test electrodes were loudness-balanced to a common reference using an adaptive two-alternative, forced-choice (2AFC), double-staircase procedure (Jestead, 1980; Zeng and Turner, 1991). Stimuli were loudness-balanced without modulation. For each subject, the reference was the C channel (see Table 3.1) presented at 25% or 50% of its DR. The current amplitude of the probe was adjusted according to subject response (2-down/1-up or 1-down/2-up, depending on the track). The initial step size was 1.2 dB and the final step size was 0.4 dB. For each run, the final 8 of 12 reversals in current amplitude were averaged, and the mean of 2-6 runs was considered to be the loudness-balanced level. The low and high presentation levels were referenced to 25% DR or 50% DR of the reference electrode, and are referred to as the 25 loudness level (LL) and 50 LL, respectively. Thus, test electrodes A, B, C, and D were equally loud at the 25 LL and at the 50 LL presentation levels.

To protect against potential loudness cues in AM detection (McKay and Henshall, 2010; Fraser and McKay, 2012), an adaptive AM loudness compensation procedure was used during the adaptive MDT task, as in Galvin et al. (2013). The AM loudness compensation functions were the same as in Galvin et al. (2013), as the subjects, reference stimuli, and loudness-balance conditions were the same. Briefly, non-AM stimuli were loudness-balanced to AM stimuli using an adaptive, 2AFC double-staircase procedure (Jestead, 1980; Zeng and Turner, 1991). The reference was the AM stimulus (AM depths = 5%, 10%, 20%, or 30%) presented to electrode

C at either 25% or 50% DR. The probe was the non-AM stimulus, also presented to electrode C. The current amplitude of the probe was adjusted according to subject response (2-down/1-up or 1-down/2-up, depending on the track). For each run, the final 8 of 12 reversals in current amplitude were averaged, and the mean of 2-6 runs was considered to be the current level needed to loudness-balance the non-AM stimulus to the AM stimulus. For each loudness balance condition, an exponential function was fit across the non-AM loudness-balanced levels at each modulation depth. The mean exponent across the exponential fits was used to customize an AM loudness compensation function for each subject. For more details, please refer to Galvin et al. (2013).

Modulation detection

MDTs were measured using an adaptive, 3AFC procedure. The modulation depth was adjusted according to subject response (3-down/1-up), converging on the threshold that corresponded to 79.4% correct [27]. One interval (randomly assigned) contained the AM stimulus and the other two intervals contained non-AM stimuli. Subjects were asked to indicate which interval was different. For each run, the final 8 of 12 reversals in AM depth were averaged to obtain the MDT; 3-6 test runs were conducted for each experimental condition.

MDTs were measured while controlling for potential AM loudness cues, as in Galvin et al. (2013). For each subject, the amount of level compensation y (in dB) was dynamically adjusted throughout the test run according to:

$$y = 20\ x\ log_{10}\ \frac{(1+m)}{(1+\alpha m)}$$

where m is the modulation index of the modulated stimulus and α is the exponent (ranging from 0 to 1) of the exponential function fit to each subject's AM vs. non-AM loudness-balance data. After applying this level compensation to the non-AM stimuli, the current level of all stimuli in each trial was independently roved by a random value between -0.75 and +0.75 dB (±4 clinical units) as in Fraser and McKay (2012).

Multi-channel MDTs

Stimuli

All stimuli were 300-ms biphasic pulse trains. The pulse phase duration was 100 μs; the inter-phase gap was 20 μs. The stimulation rate was 500 pps/electrode (ppse), resulting in a cumulative stimulation rate of 2000 pps. The modulation frequency was 10 Hz or 100 Hz. The component electrodes for the 4-channel stimuli were the same as used for single-channel modulation detection. The loudness-balanced current levels for each component electrodes were used for the 4-channel stimulus. The four channels were interleaved in time with an inter-pulse interval of 500 μs. Because of multi-channel loudness summation, the 4-channel stimulus was louder than the single-channel stimuli (McKay et al., 2001, 2003) To see the effects of loudness summation on modulation sensitivity, multi-channel MDTs were also measured after loudness-balancing the 4-channel stimulus to the same single-channel references used for the single-channel loudness balancing. Thus, 4-channel MDTs were measured with and without adjustment for loudness summation.

Coherent sinusoidal AM was applied as a percentage of the carrier pulse train amplitude according to:

$$[f(t)][1 + m \sin(2 * \pi * fm * t)]$$

where *f(t)* is a steady-state pulse train, *m* is the modulation index, and *fm* is the modulation frequency. All stimuli were presented via research interface (Wygonski and Robert, 2002).

Loudness balancing

The loudness-balanced current levels for the component electrodes were used as the initial stimulation levels for the 4-channel stimulus. The four-channel stimulus was loudness-balanced to the same single-channel reference stimuli used for single-channel loudness balancing (channel C, 500 pps, 25% or 50% DR) using the same adaptive procedure as for the single-channel loudness balancing. The current amplitude of the 4-channel probe was globally adjusted (in dB) according to the subject's response, thereby adjusting the amplitude for each electrode by the same ratio. Thus, the 4-channel stimulus was equally loud to the single-channel stimuli at the 25 LL and at the 50 LL presentation levels.

Modulation detection

Multi-channel MDTs were measured using the same adaptive, 3AFC procedure as used for single-channel modulation detection. The modulation depth applied to all 4 electrodes was adjusted according to subject response. Potential AM loudness cues were controlled using the same AM loudness compensation and level roving methods used for single-channel modulation detection. Additionally, the reference current levels within the 4-channel stimulus were independently jittered by ±0.75 dB to reduce any loudness differences across the component electrodes.

Results

Figure 3.1 shows individual and mean single-channel MDTs for the different listening conditions. Overall MDTs were highly variable across subjects, with subjects exhibiting relatively good (S1, S2, S5, S9) or poor modulation sensitivity (S3, S4, S8). Across modulation frequencies, mean MDTs were 7.57 dB better (lower) at the higher presentation level than at the lower level. Across presentation levels, mean MDTs were 7.05 dB better (lower) with the 10 Hz modulation frequency than with the 100 Hz modulation frequency. MDTs were variable across channel locations. Mean MDTs (across subjects) differed by as much as 5.74 dB across channels. For individual subjects, MDTs differed across channels by as little as 1.77 dB (S6, 25 LL, 100 Hz) to as much as 15.55 dB (S6, 50 LL, 10 Hz). A three-way repeated-measures analysis of variance (RM ANOVA) was performed on the data, with presentation level (25 LL, 50 LL), modulation frequency (10 Hz, 100 Hz), and stimulation site (A, B, C, or D) as factors. Results showed significant effects of presentation level [$F(1,8)=46.488$, $p<0.001$], modulation frequency [$F(1,8)=39.665$, $p<0.001$], and stimulation site [$F(3,24)=4.545$, $p=0.012$]. There was a significant interaction only between presentation level and modulation frequency [$F(1,8)=7.043$, $p=0.029$], most likely due to ceiling effects with the higher presentation level, especially for the 10 Hz modulation frequency. At very small modulation depths, the amplitude resolution may limit modulation sensitivity as the current level difference between the peak and valley of the modulation may be the same as or even less than each current level unit, which is approximately 0.2 dB.

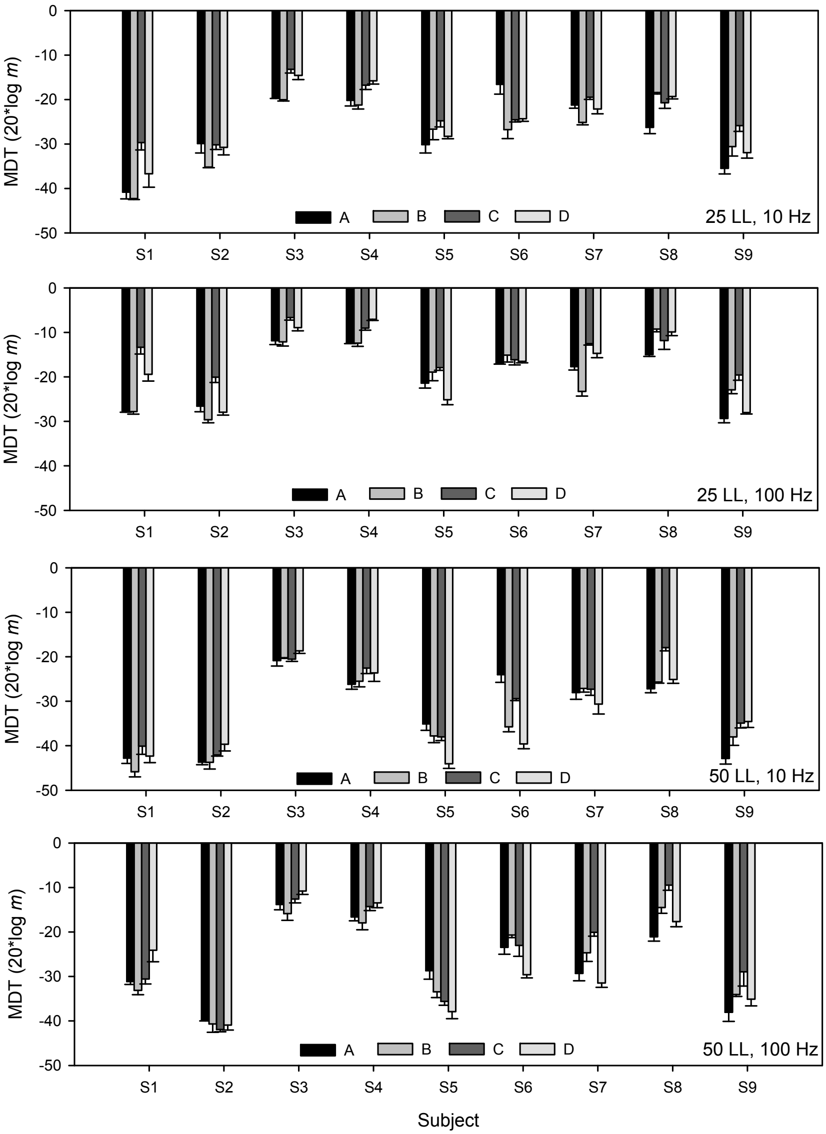

Figure 3.1. Single-channel MDTs for individual CI subjects. From top to bottom, the panels show 10-Hz MDTs at 25 LL, 100-Hz MDTs at 25 LL, 10-Hz MDTs at 50 LL, 100-Hz MDTs at 50 LL, respectively. The shaded bars show MDTs for the A, B, C, and D channels, respectively; the electrode-channel assignments are shown for each subject in Table 3.1. The error bars show the standard error.

Although the 3-way RM ANOVA showed a significant main effect of channel, there were individual differences in terms of the across-site variability in MDTs, with different best and worst channels for individual subjects. Additional 3-way ANOVAs were performed on individual subject data, with presentation level, modulation frequency and stimulation site as factors; the results are shown in Table 3.2. Significant effects were observed for presentation level in all 9 subjects, modulation frequency in 8 of 9 subjects, and stimulation site in 6 of 9 subjects. Post-hoc analyses showed that the best and worst stimulation sites differed among subjects.

Subject	Stimulation level				Modulation frequency				Stimulation site			
	dF, res	F	p	Post-hoc p<0.05	dF, res	F	p	Post-hoc p<0.05	dF, res	F	p	Post-hoc p<0.05
S1	1, 3	65	*0.004*	*50LL>25LL*	1, 3	304	*<0.001*	*10Hz>100Hz*	3,3	25	*0.012*	*A,B>C*
S2	1, 3	134	*<0.001*	*50LL>25LL*	1, 3	10	0.052		3,3	2	0.29	
S3	1, 3	26	*0.015*	*50LL>25LL*	1, 3	113	*0.002*	*10Hz>100Hz*	3,3	10	*0.044*	
S4	1, 3	278	*<0.001*	*50L >25LL*	1, 3	634	*<0.001*	*10Hz>100Hz*	3,3	41	*0.006*	*A,B>C, D*
S5	1, 3	213	*<0.001*	*50LL>25LL*	1, 3	47	*0.006*	*10Hz>100Hz*	3,3	8	0.058	
S6	1, 3	220	*<0.001*	*50L >25LL*	1, 3	166	*<0.001*	*10Hz>100Hz*	3,3	27	*0.011*	*A>D*
S7	1, 3	54	*0.005*	*50LL>25LL*	1, 3	10	*0.049*	*10Hz>100Hz*	3,3	5	0.103	
S8	1, 3	22	*0.019*	*50LL>25LL*	1, 3	143	*0.001*	*10Hz>100Hz*	3,3	17	*0.021*	*A>C*
S9	1, 3	256	*<0.001*	*50LL>25LL*	1, 3	94	*0.002*	*10Hz>100Hz*	3,3	58	*0.004*	*A>B, A,D>C*

Table 3.2. Results of three-way ANOVAs performed on individual subjects' single-channel MDT data. dF = degrees of freedom; res = residual error; F = F-ratio

Figure 3.2 shows the current level adjustment to the 4-channel stimulus needed to maintain equal loudness to the 500 pps, single-channel reference (electrode C at 25% and 50% DR). For the 4-channel stimuli, the current level adjustments were highly variable, ranging from 0.95 dB (subject S5 at the 50% DR reference) to 4.95 dB (subject S4 at the 25% DR reference). A one-way RM ANOVA showed no significant effect for reference level [$F(1,8)=2.398$, $p=0.160$], suggesting that loudness summation was similar at the relatively low and high presentation levels.

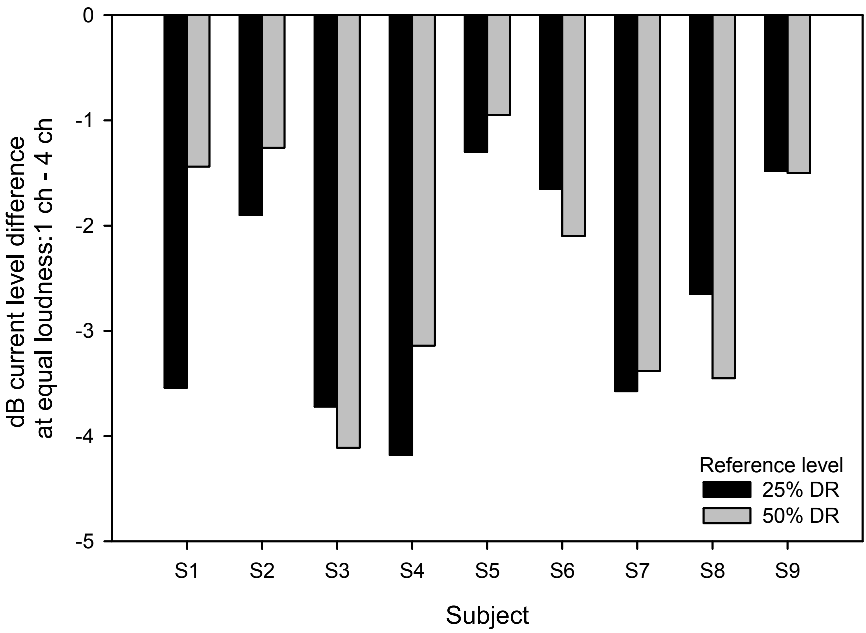

Figure 3.2. Loudness balancing between single- and multi-channel stimuli. The y-axis shows the current level adjustment needed to maintain equal loudness between 4-channel stimuli and the reference (single-channel, 500 pps, electrode C). The black bars show data referenced to 25% DR and the gray bars show data referenced to 50% DR. The error bars show the standard error.

Figure 3.3 shows individual subjects' multi-channel MDTs for the different listening conditions. The black bars show MDTs for the 4-channel loudness-balanced stimuli, which were as loud as the single-channel stimuli shown in Figure 1. The gray bars show MDTs for the 4-channel stimuli without loudness-balancing, which were louder than the single-channel stimuli shown in Figure 1 and the 4-channel loudness-balanced stimuli. As with the single-channel MDTs, multi-channel MDTs were generally better with the higher presentation level (50 LL) and the lower modulation frequency (10 Hz). In every case, 4-channel MDTs were poorer when current levels were reduced to match the loudness of the single-channel stimuli. A three-way RM ANOVA was performed on the data, with presentation level (25 LL, 50 LL), modulation frequency (10 Hz, 100 Hz), and loudness summation (4-channel with or without loudness-balancing) as factors. Results showed significant effects of presentation level [$F(1,8)=18.13$, $p=0.003$], modulation frequency [$F(1,8)=54.967$, $p<0.001$], and loudness summation [$F(1,8)=97.287$, $p<0.001$].

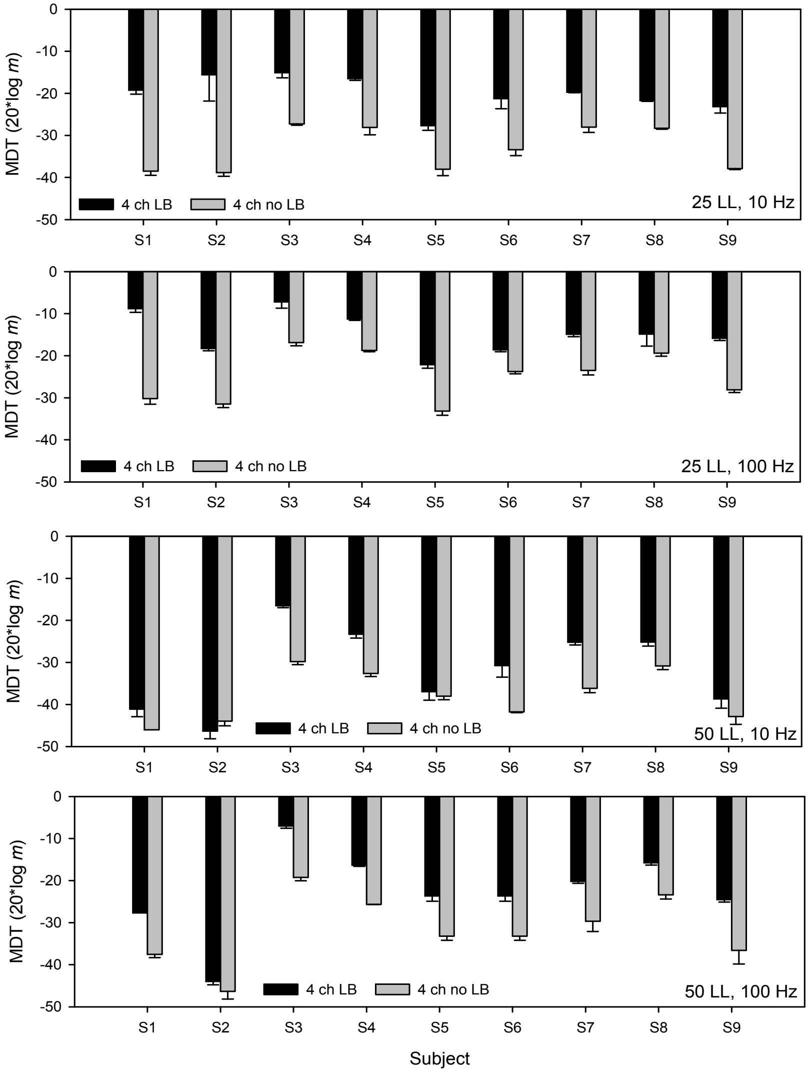

Figure 3.3. Multichannel MDTs for individual CI subjects. From top to bottom, the panels show 10-Hz MDTs at 25 LL, 100-Hz MDTs at 25 LL, 10-Hz MDTs at 50 LL, 100-Hz MDTs at 50 LL, respectively. The black bars show the MDTs for the 4-channel loudness-balanced stimuli (i.e., equally loud as the single-channel stimuli in Fig. 3.1) and the gray bars show MDTs for the 4-channel stimuli without loudness-balancing (i.e., louder than the single-channel stimuli in Fig. 3.1 and the 4-channel loudness-balanced stimuli). The error bars show the standard error.

Figure 3.4 shows boxplots for MDTs averaged across single channels or with the 4-channel loudness-balanced stimuli. Note that all stimuli were equally loud. Across all conditions, the average single-channel MDT was 3.13 dB better (lower) than with the 4-channel loudness-balanced stimuli; mean differences ranged from 0.70 dB for the 50 LL/10 Hz condition to 5.44 dB for the 25 LL/10 Hz condition. A Wilcoxon signed rank test showed that the average single-channel MDT was significantly better than that with the 4-channel loudness-balanced stimuli ($p=0.003$). Similarly, a ranked sign test showed that MDTs with the best single channel were significantly better than those with the 4-channel loudness-balanced stimuli ($p<0.001$). Finally, a ranked sign test showed that the difference between MDTs with the worst single channel and with the 4-channel loudness-balanced stimuli failed to achieve significance ($p=0.052$).

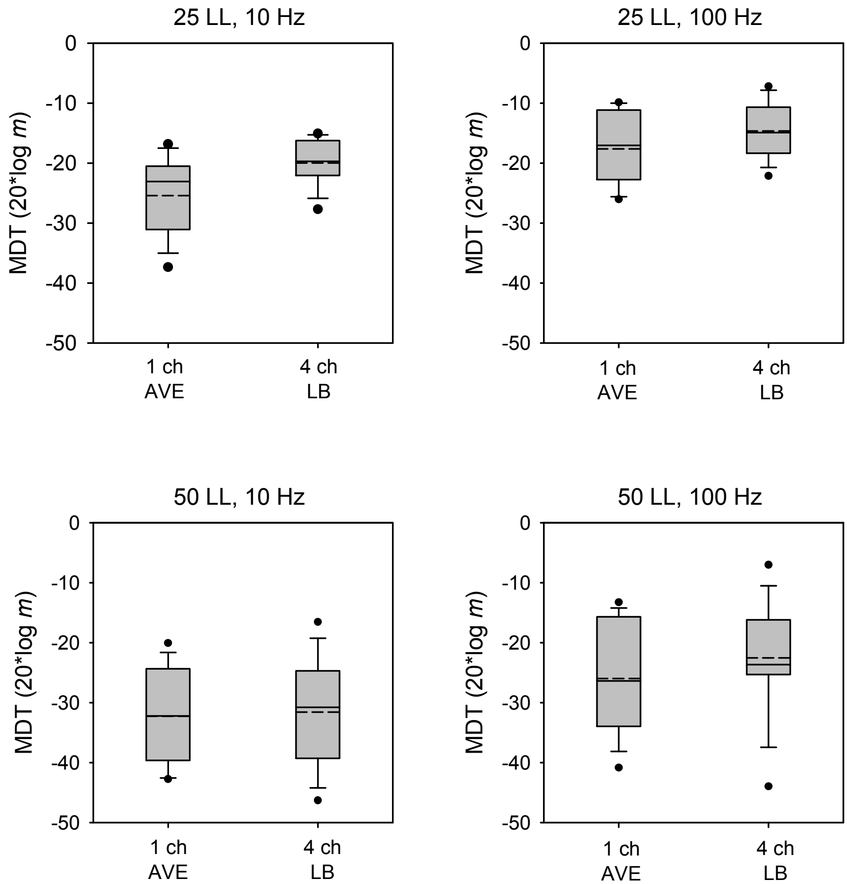

Figure 3.4. MDTs for equally loud single- and multi-channel stimuli. Box plots are shown for MDTs averaged across the best single channel or with the 4-channel loudness-balanced stimuli; note that all stimuli were equally loud. From left to right, the panels show data for the 25 LL/10 Hz, 25 LL/100 Hz, 50 LL/10 Hz, 50 LL/100 Hz conditions. In each box, the solid line shows the median, the dashed line shows the mean, the error bars show the 10th and 90th percentiles, and the black circles show outliers.

Figure 3.5 shows boxplots for MDTs with the best single channel or with the 4-channel stimuli with no loudness compensation. Thus, the 4-channel stimuli were louder than the single-channel stimuli. Across all conditions, the mean MDT was 3.01 dB better with the 4-channel stimuli than with the best single channel; mean differences ranged from 1.97 dB for the 50 LL/100 Hz condition to 3.97 dB for the 25 LL/10 Hz condition. A paired t-test across all conditions showed that MDTs were significantly better with the 4-channel stimuli than with the best single channel (p=0.001).

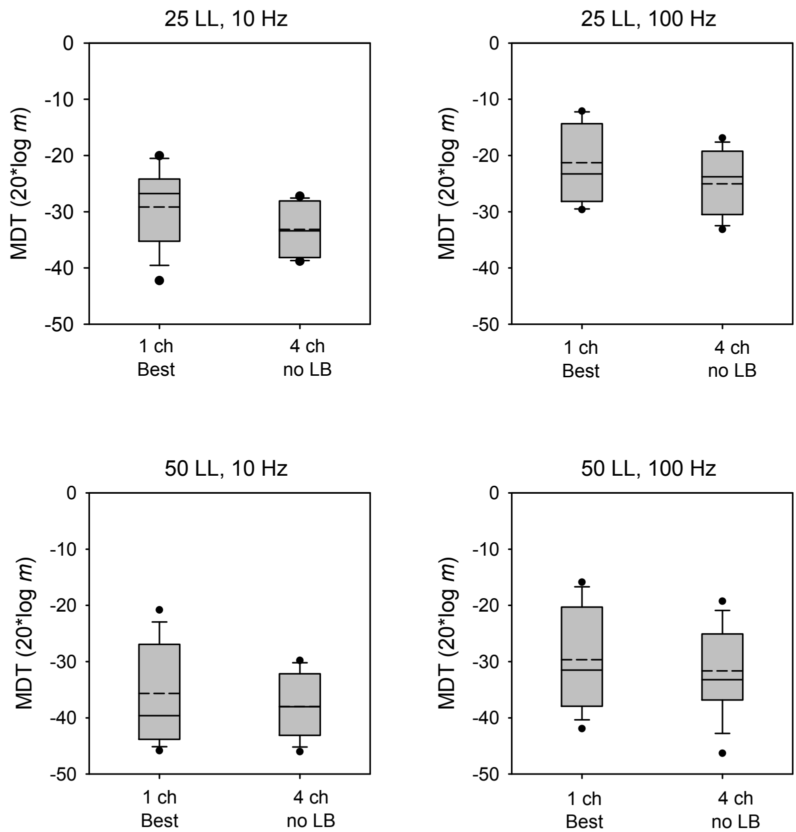

Figure 3.5. MDTs for single- and multi-channel stimuli without loudness summation compensation. Box plots are shown for MDTs with the best single-channel or with the 4-channel stimuli without loudness-balancing; note that the 4-channel stimuli without loudness-balancing were louder than the single-channel stimuli. From left to right, the panels show data for the 25 LL/10 Hz, 25 LL/100 Hz, 50 LL/10 Hz, 50 LL/100 Hz conditions. In each box, the solid line shows the median, the dashed line shows the mean, the error bars show the 10th and 90th percentiles, and the black circles show outliers.

As shown in Figure 3.1, across-site variability in MDTs differed greatly across subjects. It is possible that subjects with greater across-site variability may attend more to the single channel with the best modulation sensitivity when listening to the 4-channel stimuli. Similarly, subjects with less across-site variability may better integrate information across all channels in the 4-channel stimuli. The mean across-site variance in single-channel MDTs was calculated for individual subjects across the presentation level and modulation frequency test conditions, as in Garadat et al. (2012). Across all subjects, the mean variance was 10.08 dB2, and ranged from 3.91 dB2 (subject S4) to 19.07 dB2 (subject S1). Individual subjects' mean across-site variance was compared to the multi-channel advantage (with no loudness compensation) in modulation detection over the best single channel without loudness-balancing (i.e., 4-channel MDT – best single-channel MDT). Linear regression analysis showed no significant relationship between the degree of multi-channel advantage and across-site variance ($r^2=0.181$, $p=0.253$).

As shown in Figure 3.3, performance with 4-channel stimuli was much poorer when the current levels were reduced to match the loudness of single-channel stimuli. Figure 3.2 shows great inter-subject variability in terms of multi-channel loudness summation. It is possible that the degree of multi-channel loudness summation may be related to the deficit in multi-channel modulation sensitivity after compensating for loudness summation. The mean loudness summation across both presentation levels was calculated for individual subjects, and was compared to the difference in MDTs between 4-channel stimuli with and without loudness-balancing. Linear regression analysis showed no significant correlation between the degree of multi-channel loudness summation and the difference in MDTs between the 4-channel stimuli with or without loudness compensation ($r^2=0.014$, $p=0.79$).

Discussion

The present data suggest that, at equal loudness, MDTs were poorer with 4 channels than with a single channel, most likely due to the lower current levels in the 4-channel stimuli needed to maintain equal loudness to the single-channel stimuli. With no compensation for loudness multi-channel summation, MDTs were significantly better with 4-channel stimuli than with the best single channel, suggesting some multi-channel advantage. Below, we discuss the results in greater detail.

Effects of Presentation Level and Modulation Frequency

With single- or multi-channel stimulation, MDTs generally improved as the presentation level was increased and/or the modulation frequency was decreased, consistent with many previous studies (Shannon, 1992; Donaldson and Viemester, 2000; Fu, 2004;.Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007; Galvin et al., 2013). For both the single- and 4-channel stimuli, mean MDTs were 7.67 dB better with the 50 LL than with the 25 LL presentation level, and 7.07 dB better with the 10 Hz than with the 100 Hz modulation frequency.

Effect of Loudness Summation on Multi-channel MDTs

At equal loudness, 4-channel MDTs were significantly poorer than the average single-channel MDT (Fig. 3.4); 4-channel MDTs were also significantly poorer after compensating for multi-channel loudness summation (Fig. 3.3). In both cases, the deficits were presumably due to lower current levels on each channel needed to compensate for multi-channel loudness summation. MDTs are very level dependent, especially at lower presentation levels (Shannon, 1992; Donaldson and Viemester, 2000; Fu, 2004; Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007). The present data suggest that at equal loudness, single-channel estimates of modulation sensitivity may greatly over-estimate the functional sensitivity when multiple channels are stimulated. In clinical speech processors, current levels must often be reduced to accommodate multi-channel loudness summation. The present data suggests that such current level adjustments may worsen multi-channel modulation sensitivity.

Loudness summation was not significantly correlated with the difference in MDTs between 4-channel stimuli with or without loudness compensation. This may reflect individual subject variability in modulation sensitivity, especially at presentation low levels. Such variability has been reported in many studies (Donaldson and Viemester, 2000; Fu, 2004; Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007). Some subjects may have been more susceptible than others to the level differences between the 4-channel stimuli with and without loudness compensation.

Note that in the present study, we were unable to measure single-channel MDTs at the component channel stimulation levels used in the 4-channel loudness-balanced stimuli. After the current adjustment to accommodate multi-channel loudness summation, the component channel current levels were often too low (i.e., below detection thresholds) to measure single-channel MDTs.

Multi-channel loudness summation may also explain some of the advantage of multi-channel stimulation observed by Geurts and Wouters (2001) in AM frequency discrimination. Similar to their findings, the present data showed that multi-channel stimulation without loudness compensation offered a small but significant advantage over the best single channel. In Geurts and Wouters (2001) there was no level adjustment to equate loudness between the single- and multi-channel stimuli. If such a level adjustment had been applied to the multi-channel stimuli, AM frequency discrimination may have better with single than with multiple channels, as in the present study with modulation detection. Future studies may wish to examine how component channels contribute to AM frequency discrimination in a multi-channel context in which loudness summation does not play a role.

Contribution of Single Channels to Multi-channel MDTs

Across-site variability was not significantly correlated with the multi-channel advantage over the best single channel, suggesting that CI subjects combined information across channels, instead of relying on the channels with best temporal processing, even when there was great variability in modulation sensitivity across stimulation sites. This finding is in agreement with recent multi-channel MDI studies in CI users (Chatterjee, 2003; Kreft et al., 2013) that suggest that multi-channel envelope processing is more centrally than peripherally mediated.

Implications for Cochlear Implant Signal Processing

The present data suggest that accommodating multi-channel loudness summation, as is necessary when fitting clinical speech processors, may reduce CI users' functional modulation sensitivity. When high stimulation rates are used on each channel, the functional temporal processing may be further compromised, as the current levels must be reduced to accommodate summation due to high per-channel rates and multi-channel stimulation. Selecting a reduced set of optimal channels (ideally, those with the best temporal processing) to use within a clinical speech processor may reduce loudness summation, allowing for higher current levels to be used on each channel. Such optimal selection of channels has been studied by Garadat et al. (2012), who found better speech understanding in noise when only the channels with better temporal processing were included in the speech processor. In that study, subjects were allowed to adjust the speech processor volume for the experimental maps, which may have compensated for the reduced loudness associated with the reduced-electrode maps, possibly resulting in higher stimulation levels on each channel. Bilateral signal processing may also allow for fewer numbers of electrodes within each side, thereby reducing loudness summation, increasing current levels, and thereby improving temporal processing. The reduced numbers of channels on each ear may be combined, as the spectral holes on one side are filled in by the other. Such optimized “zipper processors’ have been explored by Zhou and Pfingst (2012), who found better speech performance in some subjects, presumably due to the increased functional spectral resolution. Using fewer channels within each speech processor may have also reduced loudness summation, resulting in higher current levels and better temporal processing.

Loudness summation and spatio-temporal channel interactions should be carefully considered to improve the spectral resolution and temporal processing for future CI signal processing strategies. It is possible that by selecting a fewer number of optimal electrodes (in terms of temporal processing and key spectral cues) within each stimulation frame would reduce the instantaneous loudness summation, allowing for higher current levels that might produce better temporal processing. Using relatively low stimulation rates (e.g., 250 – 500 Hz/channel) might help reduce channel interaction between adjacent electrodes. Zigzag stimulation patterns which maximize the space between electrodes in sequential stimulation (e.g., electrode 1, then 9, then 5, then 13, then 3, then 11, etc.) might also help to channel interaction.

Conclusions

Single- and multi-channel modulation detection was measured in CI users. Significant findings include:

1. Effects of presentation level and modulation frequency were similar for both single- and multi-channel MDTs; performance improved as the presentation level was increased or the modulation frequency was decreased.
2. At equal loudness, single-channel MDTs may greatly over-estimate multi-channel modulation sensitivity, due to the lower current levels needed to accommodate loudness summation in the latter.

3. When there was no level compensation for loudness summation, multi-channel MDTs were significantly better than MDTs with the best single channel.

4. There was great inter-subject variability in terms of multi-channel loudness summation. However, the degree of loudness summation was not significantly correlated with the deficit in modulation sensitivity when current levels were reduced to accommodate multi-channel loudness summation.

5. There was also great inter-subject variability in the across-site variance observed for single-channel MDTs. However, across-site variability was not significantly correlated with the multi-channel advantage over the best single-channel. This suggests that CI listeners combined information across multiple channels rather that attend primarily to the channels with the best modulation sensitivity.

Acknowledgments

We thank all implant subjects for their participation, Joseph Crew for help with data collection, as well as Monita Chatterjee, David Landsberger, Bob Shannon, Justin Aronoff, and Robert Carlyon for helpful comments. Work supported by NIH grant DC004993. Dr. Baskent was supported by VIDI grant 016.096.397 from the Netherlands Organization for Scientific Research (NWO) and the Netherlands Organization for Health Research and Development (ZonMw), and a Rosalind Franklin Fellowship from University of Groningen, University Medical Center Groningen.

Chapter 4

Modulation frequency discrimination with single and multiple channels in cochlear implant users

John J. Galvin III
Sandy Oba
Deniz Başkent
Qian-Jie Fu

This chapter is a modified version of a paper published in:
Hearing Research (2015); 324: 7-18

Abstract

Temporal envelope cues convey important speech information for cochlear implant (CI) users. Many studies have explored CI users' single-channel temporal envelope processing. However, in clinical CI speech processors, temporal envelope information is processed by multiple channels. Previous studies have shown that amplitude modulation frequency discrimination (AMFD) thresholds are better when temporal envelopes are delivered to multiple rather than single channels. In clinical fitting, current levels on single channels must often be reduced to accommodate multi-channel loudness summation. As such, it is unclear whether the multi-channel advantage in AMFD observed in previous studies was due to coherent envelope information distributed across the cochlea or to greater loudness associated with multi-channel stimulation.

In this study, single- and multi-channel AMFD thresholds were measured in CI users. Multi-channel component electrodes were either widely or narrowly spaced to vary the degree of overlap between neural populations. The reference amplitude modulation (AM) frequency was 100 Hz, and coherent modulation was applied to all channels.

In Experiment 1, single- and multi-channel AMFD thresholds were measured at similar loudness. In this case, current levels on component channels were higher for single- than for multi-channel AM stimuli, and the modulation depth was approximately 100% of the perceptual dynamic range (i.e., between threshold and maximum acceptable loudness). Results showed no significant difference in AMFD thresholds between similarly loud single- and multi-channel modulated stimuli.

In Experiment 2, single- and multi-channel AMFD thresholds were compared at substantially different loudness. In this case, current levels on component channels were the same for single- and multi-channel stimuli ("summation-adjusted" current levels) and the same range of modulation (in dB) was applied to the component channels for both single- and multi-channel testing. With the summation-adjusted current levels,

loudness was lower with single than with multiple channels and the AM depth resulted in substantial stimulation below single-channel audibility, thereby reducing the perceptual range of AM. Results showed that AMFD thresholds were significantly better with multiple channels than with any of the single component channels. There was no significant effect of the distribution of electrodes on multi-channel AMFD thresholds.

Overall, the results suggest that increased loudness due to multi-channel summation may contribute to the multi-channel advantage in AMFD, and that that overall loudness may matter more than the distribution of envelope information in the cochlea.

Introduction

In cochlear implants (CIs), low-frequency temporal envelope cues (<20 Hz) are important for speech understanding, while higher frequency envelope cues (80 – 300 Hz) are important for perception of voice pitch. Given the limited spectral resolution of the device, CI users strongly rely on temporal envelope cues for pitch-mediated speech tasks such as voice gender perception (Fu et al., 2004, 2005; Fuller et al., 2014), vocal emotion recognition (Luo et al., 2007), tonal language perception (Luo at al., 2008), and speech prosody perception (Chatterjee and Peng, 2007).

Temporal processing in CIs has been widely studied in terms of single-channel modulation detection thresholds (MDTs; Shannon, 1992; Busby et al., 1994; Chatterjee and Oba, 2005; Galvin and Fu, 2005, 2009; Pfingst et al., 2007; Won et al., 2011; Fraser and McKay, 2012; Green et al., 2012). Modulation detection is one of the few single-channel psychophysical measures that have been significantly correlated with speech perception for CI users (Cazals et al., 1994; Fu, 2002) and recipients of auditory brainstem implants (Coletti and Shannon, 2005), underscoring the importance of temporal processing to speech perception. Modulation detection has also been significantly correlated with modulation frequency discrimination (Chatterjee and Ozerbut, 2011), which is typically measured using envelope depths well above MDTs. The perception of changes in modulation frequency is highly relevant for perception of pitch cues in speech (e.g., voice gender, vocal emotion, lexical tones, prosody, etc.). Modulation frequency discrimination has been correlated with CI users' perception of lexical tones (Chatterjee and Peng, 2008; Luo et al., 2008), which depend strongly on perception of voice fundamental frequency (F0).

Previous CI studies have measured various aspects of amplitude modulation frequency discrimination (AMFD). Many studies have shown that, given a fixed amplitude modulation (AM) depth, single-channel AMFD thresholds generally improve as the current level is increased (Morris and Pfingst, 2000; Luo et al., 2008; Chatterjee and Ozerbut, 2011; Green et al., 2012). Geurts and Wouters (2001) found better single-channel AMFD with a fixed modulation frequency difference as the modulation depth was increased. However, Chatterjee and Peng (2008) found no consistent effect for modulation depths between 5% and 30% of the reference amplitude on single-channel AMFD thresholds. Efforts to enhance temporal envelope cues have shown mixed results for AMFD. Green et al. (2004) showed a small but significant advantage for perception of modulated frequency sweeps across multiple channels when the temporal envelope was sharpened ("sawsharp" enhancement). However, subsequently, Green et al. (2005) found poorer vowel recognition with the enhancement relative to the standard continuously interleaved sampling (CIS; Wilson et al., 1991) signal processing strategy, possibly due to its effect on spectral envelope cues. Hamilton et al. (2007) found that presenting modified temporal information to only one of six stimulated channels (rather than all channels as in Green et al., 2005), offered no clear advantage in a variety of speech recognition tasks. Landsberger (2008) found no significant difference in single-channel AMFD thresholds between sine, sawtooth, and sharpened sawtooth temporal envelopes. Kreft et al. (2010) found no significant difference in single-channel AMFD thresholds for pulse trains that were amplitude modulated by sine waves or by rectified sine waves, the latter of which was proposed to more closely resemble normal neural responses to low-frequency pure tones. Chatterjee and Ozerbut (2011) found some evidence of modulation tuning for AMFD thresholds, with increased sensitivity near 100 Hz, above and below which AMFD thresholds increased. When presented at a similar loudness level (i.e., 75% of the dynamic range, or DR), Green et al. (2012) showed no significant effect of carrier pulse rate on single-channel AMFD thresholds, despite better envelope representation with high carrier rates. Taken together, these single-channel studies suggest that, AMFD is strongly affected by current level and modulation depth, with modulation depth interacting with current level.

Although clinical CI speech processors provide multi-channel stimulation, very few studies have directly measured AMFD using multiple channels. Multi-channel envelope processing has mostly been measured using modulation detection interference (MDI) paradigms, in which CI users are asked to detect AM or discriminate between AM frequencies presented to one channel in the presence of competing AM on the same channel or other channels. Chatterjee (2003) found substantial modulation masking (defined as the difference in MDT between a dynamic and steady-state masker) even when masker channels were spatially remote from the target channel. Chatterjee and Oba (2004) found greater MDI for modulation detection when the modulation frequency of the interferer was lower than that of the target. Kreft et al. (2013) found a similar effect of masker-target modulation frequency for AMFD thresholds. In these studies, there was substantial off-channel masking, possibly due to the broad current spread associated with electric stimulation, and possibly due to envelope interactions beyond the auditory periphery.

Intuitively, multi-channel stimulation would be expected to offer some advantage in perception of coherent envelope information, relative to single-channel stimulation. Indeed, Geurts and Wouters (2001) found better AMFD thresholds with multiple channels than with any of the single component channels used for the multi-channel stimuli. However, no explicit adjustment was made for multi-channel loudness summation in Geurts and Wouters (2001). Work by McKay and colleagues (McKay et al., 2001; 2003) showed substantial multi-channel loudness summation independent of electrode spacing. As such, the multi-channel stimuli in Geurts and Wouters (2001) might have been louder than the single-channel stimuli, contributing to the multi-channel advantage. Previous studies (Morris and Pfingst, 2000; Luo et al., 2008; Chatterjee and Ozerbut, 2011; Green et al., 2012) have shown that single-channel AMFD improves with level (and by association, loudness). Interestingly, Galvin et al. (2014) found that multi-channel MDTs were better than MDTs with any of the single component channels. However, when the current levels were reduced in the multi-channel AM stimuli to match the loudness of the single-channel AM stimuli, multi-channel MDTs were significantly poorer than single-channel MDTs. As modulation detection is level-dependent, the reduced current levels required to accommodate multi-channel loudness summation resulted in poorer MDTs. It is unclear how multi-channel loudness summation may affect AMFD, while understanding perceptual mechanisms that may underlie multi-channel temporal processing is crucial and clinically relevant as CI speech processors are fit to accommodate multi-channel loudness summation.

In this study, single- and multi-channel AMFD was measured in CI users. Component electrodes were distributed to target relatively overlapping (narrow configuration) and non-overlapping neural populations (wide configuration). We hypothesized that AMFD would be better with the wide configuration due to multiple, relatively independent envelope cues, In Experiment 1, single- and multi-channel AMFD thresholds were measured at similar loudness. In this case, current levels were higher for single-channel AM stimuli than for multi-channel AM stimuli, due to multi-channel loudness summation. We hypothesized that for similarly loud AM stimuli, AMFD would be poorer with multiple than with single channels due to the reduced current levels needed to accommodate multi-channel loudness summation, similar to the MDT findings data from Galvin et al. (2014). In Experiment 2, single- and multi-channel AMFD thresholds were measured using the same summation-adjusted current levels for component channels. In this case, multi-channel AM stimuli were louder than the single-channel AM stimuli, due to multi-channel loudness summation. We hypothesized that, without adjustment for multi-channel loudness summation, AMFD would be better with multiple than with single channels, as in Geurts and Wouters (2001).

Methods

Participants

Five adult, post-lingually deafened CI users participated in this experiment. All were users of Cochlear Corp. devices and all had more than 2 years of experience with their implant device. Relevant subject details are shown in Table 4.1. Four of the 5 subjects previously participated in a related modulation detection study (Galvin et al., 2014). Subjects S1, S2, S3, and S5 were bilateral CI users; S1 and S3 were tested using the first implant while S2 and S5 were tested using the second implant. All subjects provided written informed consent prior to participating in the study, in accordance with the guidelines of the St. Vincent Medical Center Institutional Review Board (Los Angeles, CA), which specifically approved this study. All subjects were financially compensated for their participation.

Subject	Age at testing (yrs)	Age at implantation (yrs)	Duration of deafness (yrs)	Etiology	Device	Strategy
S1	70	60	23	Genetic	N24	ACE
S2	79	77	35	Otosclerosis	N5	ACE
S3	28	26	11	Acoustic Neuroma	Freedom	ACE
S4	67	59	20	Meniere's/ Otosclerosis	Freedom	ACE
S5	78	76	8	Unknown	N5	ACE

Table 4.1. CI subject demographics.

Stimuli.

All stimuli were 300-ms biphasic pulse trains; the stimulation rate was 2000 pulses per second (pps) per electrode. The relatively high stimulation rate was chosen to ensure adequate sampling of the maximum AM frequency tested (356 Hz) and to approximate the default cumulative stimulation rate across all channels used in Cochlear Corp. devices (8 maxima x 900 pps/channel = 7200 pps cumulative rate). The pulse phase duration was 25 μs and the inter-phase gap was 8 μs. Monopolar stimulation was used. Two sets of three electrodes were selected for multi-channel stimuli to represent different amounts of channel interaction: a “wide” configuration consisting of electrodes 4, 10, and 16 and a “narrow” configuration consisting of electrodes 9, 10, and 11. The wide configuration was expected to target relatively independent neural populations and the narrow configuration was expected to target overlapping neural populations. All stimuli were presented via research interface (Wygonski and Robert, 2002), bypassing subjects’ clinical processors and settings; custom software was used to deliver the stimuli and to record subject responses.

The electric dynamic range (DR) was first estimated for all single electrodes without AM. Absolute detection thresholds were initially estimated using a “counting” method, as is sometimes used for clinical fitting of speech processors. A number of 300-ms pulse train bursts (randomly selected between 2 and 5, with a 500 ms interval between bursts) were presented to the subject, who indicated how many bursts were heard. Stimulation initially began at sub-threshold levels and the current level was adjusted in 0.5 dB steps according to correctness of response (1-up/1 down). The detection threshold was the amplitude for the final of 4 reversals in current level. Maximum acceptable loudness (MAL) levels, defined as the “loudest sound that could be tolerated for a short time,” were initially estimated by slowly increasing the current level (in 0.2 dB steps) for 3 pulse train bursts until reaching MAL. Note that MALs are higher than comfort levels (C-levels) measured during clinical fitting of CI speech processors. Threshold and MAL levels were averaged across a minimum of two runs, and the DR was calculated as the difference in dB (re: 1 μA) between MAL and threshold.

Test electrodes were swept for loudness at 10% DR, 50% DR, and 100% DR (MAL) to ensure equal loudness, as is often done during clinical fitting of speech processors. The percent DR was calculated first in microamps and then converted to dB (re:

1 μA). During sweeping, 300 ms pulse trains were delivered to all electrodes (4, 9, 10, 11, and 16) in sequence (first from apex to base, and then from base to apex). The subject indicated which (if any) of the electrodes were louder or softer than the rest. If there were loudness differences across electrodes at 50% or 100% DR, the level of the different electrode was adjusted (up or down, as needed) by 0.4 dB (approximately 2 clinical units), and the electrodes were re-swept for loudness. If there were loudness differences across electrodes at 10% DR, the threshold level of the different electrode was adjusted (up or down, as needed) by 0.4 dB, and the electrodes were re-swept for loudness at 10% DR. After making all adjustments to obtain equal loudness, the final threshold, MAL and DR values for each electrode were recorded.

For the multi-channel stimuli, the component electrodes were optimally interleaved in time; the onset of each pulse was separated by 0.167 ms and the inter-pulse interval (between the offset of one pulse and the onset of the next pulse) was 0.109 ms. Because of loudness summation associated with multi-channel stimulation (McKay et al., 2001, 2003), the 3-channel stimuli were loudness-balanced to a common single-channel reference (electrode 10) presented at 50% DR (calculated in microamps then converted to dB re: 1 μA). The reference level of 50% DR was selected because the subsequent single-channel AMFD was measured for an AM depth of 100% DR (±50% DR re: reference of 50% DR). An adaptive two-alternative, forced-choice (2AFC), double-staircase procedure was used for loudness balancing (Jesteadt, 1980; Zeng and Turner, 1991); an ascending and descending track were randomly interleaved during each run. Stimuli were loudness-balanced without AM. In each trial for each track, two intervals were presented; the single-channel reference was randomly assigned to one interval and the multi-channel probe was assigned to the other. Subjects were asked to indicate which interval was louder, ignoring all other qualities of the stimuli. The current of the multi-channel probe was globally adjusted (in dB) according to subject response (2-down/1-up or 1-down/2-up, depending on the track), thereby adjusting the amplitude for each component electrode by the same ratio. The initial step size was 1.2 dB and the final step size was 0.4 dB. For each run, the final 8 of 12 reversals in current amplitude were averaged, and the mean of 2-3 runs was considered to be the loudness-balanced level. After adjustment for the multi-channel loudness summation, the current levels on the component electrodes were substantially reduced. These "summation-adjusted" current levels are indicated by an apostrophe throughout this paper (e.g., 4'). Note that the level adjustments for electrode 10 depended on the amount of summation associated with wide or narrow multi-channel configurations; hence the 10w' and 10n' designations.

Coherent sinusoidal AM was applied as a percentage of the carrier pulse train amplitude according to:

$$[f(t)][1 + m\sin(2 * \pi * fm * t)]$$

where *f(t)* is a steady-state pulse train, *m* is the modulation index, and *fm* is the modulation frequency. Note that modulation was applied both above and below the carrier reference level. A 10-ms onset and offset ramp in amplitude was applied to all AM stimuli. The initial modulation phase was 180 degrees for all stimuli. For the single-channel stimuli 4, 9, 10, 11, and 16, the modulation depth was between threshold and MAL (i.e., the entire DR). This maximum modulation depth was selected to provide strong envelope cues across different experimental conditions, as in Kreft et al. (2010, 2013).The same modulation depths (in dB) were used for the summation-adjusted component electrodes. Figure 4.1 illustrates the current levels and modulation depths for three electrodes (wide configuration) for subject S3 (see Table 4.2 for exact values). For the original single-channel AM stimuli (left part of Fig. 4.1), AM depth was between threshold and MAL (100% DR). For the multi-channel AM stimuli (middle part of Fig. 4.1), current levels were reduced to accommodate multi-channel loudness summation. AM depth on each channel was the same (in dB) as for the original single channels (9.03, 9.58, and 9.18 dB for electrodes 4, 10, and 16, respectively). The perceptual range of the AM was presumably similar between these similarly loud single- and multi-channel AM stimuli, although this was not explicitly measured. For the summation-adjusted single-channel AM stimuli (right part of Fig. 4.1), the same current levels and modulation range (in dB) were used as for the multi-channel stimuli. However, these single-channel AM stimuli were much softer than the multi-channel AM stimuli (and the original single-channel AM stimuli). While the range of modulation (in dB) was the same for all component channels (regardless of the current level or the number of channels stimulated), the perceptual range of modulation was likely much reduced for the single-channel summation-adjusted AM stimuli. Here, peak AM current levels was approximately 50% of the original single-channel DR and the minimum AM current levels were substantially below the original single-channel thresholds (solid horizontal lines). Thus, the single- and multi-channel AM stimuli on the left half of Figure 4.1 had similar overall loudness but different current levels, while the single- and multi-channel AM stimuli on the right half of Figure 4.1 had different overall loudness but the same current levels on each component channel.

Table 4.2 shows the test electrodes for each subject and condition, original threshold and MAL (in dB), summation-adjusted threshold and MAL (in DB), and the original DR (also the range of modulation for all AM stimuli, in dB). When measuring multi-channel AMFD, the current levels of the component channels were independently roved by ±1 dB to reduce any potential loudness differences among channels that may have escaped the initial loudness balancing procedure.

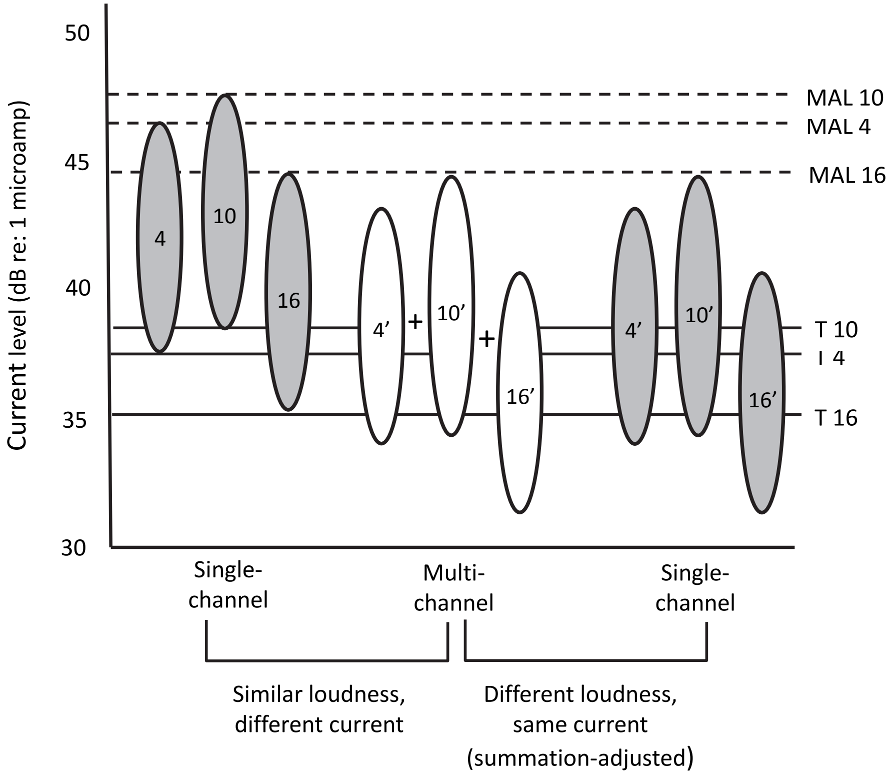

Figure 4.1. Illustration of the current levels and modulation depths used for each experimental condition, for subject S3. The ovals on the left side of the figure show the range of modulation for electrodes 4, 10, and 16 (original single-channel AM stimuli); the solid lines show the original thresholds (T) and the dashed lines show the original maximum acceptable loudness (MAL). These single-channel AM stimuli were similarly loud. The middle group of ovals shows current levels of the multi-channel AM stimuli after adjusting for multi-channel loudness summation. The right group of ovals shows the same summation-adjusted current levels for single-channel AM stimuli as used for the multi-channel AM stimuli. The left and middle groups of ovals were of similar loudness, but with different current levels, while the middle and right groups of ovals were of different loudness (multi-channel louder), but with the same current levels used on each component channel. Note also that the range of modulation (in dB) is the same for each component channel, regardless of experimental condition.

Subject	Configuration	Electrode	Single-channel (El x) Threshold	Single-channel (El x) MAL	Single-channel, multi-channel (El x') Threshold	Single-channel, multi-channel (El x') MAL	DR
S1	Wide	4	46.02	58.87	43.97	56.82	12.85
		10	45.85	60.28	43.80	58.23	14.44
		16	44.08	58.17	42.03	56.12	14.09
	Narrow	9	45.67	59.75	44.24	58.33	14.09
		10	45.85	60.28	44.42	58.86	14.44
		11	44.40	59.75	42.98	58.33	15.35
		AVE	45.31	59.52	43.57	57.78	14.21
		STD	0.84	0.84	0.91	1.06	0.81
S2	Wide	4	42.54	51.20	38.48	47.14	8.66
		10	40.98	51.36	36.92	47.30	10.38
		16	41.14	50.58	37.08	46.52	9.44
	Narrow	9	41.44	51.53	36.81	46.90	10.09
		10	40.98	51.36	36.35	46.73	10.38
		11	40.83	51.05	36.20	46.42	10.23
		AVE	41.32	51.18	36.97	46.84	9.86
		STD	0.63	0.34	0.81	0.35	0.69
S3	Wide	4	37.62	46.65	34.02	43.05	9.03
		10	38.17	47.75	34.57	44.15	9.58
		16	35.12	44.30	31.52	40.70	9.18
	Narrow	9	38.28	48.06	33.37	43.15	9.79
		10	38.17	47.75	33.26	42.84	9.58
		11	38.79	47.75	33.88	42.84	8.96
		AVE	37.69	47.04	33.43	42.79	9.35
		STD	1.31	1.43	1.05	1.13	0.34
S4	Wide	4	46.65	54.96	36.20	46.42	10.23
		10	46.49	55.75	34.02	43.05	9.03
		16	44.45	55.75	34.57	44.15	9.58
	Narrow	9	46.97	55.92	40.94	49.89	8.95
		10	46.49	55.75	40.46	49.72	9.26
		11	46.49	57.01	40.46	50.98	10.53
		AVE	46.25	55.86	37.77	47.37	9.60
		STD	0.90	0.66	3.20	3.31	0.65
S5	Wide	4	41.65	50.90	37.76	47.01	9.25
		10	42.12	52.94	38.23	49.05	10.82
		16	40.86	50.74	36.97	46.85	9.88
	Narrow	9	41.02	52.63	36.66	48.27	11.61
		10	42.12	52.94	37.76	48.58	10.82
		11	41.80	53.10	37.44	48.74	11.30
		AVE	41.60	52.21	37.47	48.08	10.61
		STD	0.54	1.09	0.57	0.93	0.89

Table 4.2. Threshold and MAL current levels in dB (re: 1 μA), with (El x; original single-channel levels) and without compensation for multi-channel loudness summation (El x'; summation-adjusted levels). For each experimental condition, AM was between these current levels. The DR also represents the range of modulation that was fixed for each electrode across conditions. For each subject, the mean and standard deviation of the threshold, MAL, and DR was calculated across all electrodes.

Procedure

AMFD was measured using a method of constant stimuli. The reference modulation frequency was 100 Hz; the probe modulation frequency was 101, 102, 104, 108, 116, 132, 164, 228, or 356 Hz. A 3AFC procedure was used. While AM frequency may affect loudness (Vandali et al., 2013) given a fixed AM depth, these effects were expected to be small for the presentation levels and AM depths used in this study. To minimize the effects of loudness difference across AM frequencies, the current of the stimulus in each interval was globally roved by ± 1 dB, similar to Chatterjee and Ozerbut (2012) and Kreft et al. (2010; 2013). Note that for multi-channel AM stimuli, this global roving was in addition to the component channel roving of ± 1 dB, which was performed to reduce any potential loudness differences among channels. Two of the present subjects were asked to loudness-balance single-channel AM stimuli with 100 Hz versus 356 Hz AM rates and 100% DR modulation depth. Results showed no clear or consistent differences in loudness between the 100 Hz and 356 Hz AM stimuli.

During each experimental trial, the probe was randomly assigned to one of the three intervals and the reference was assigned to the remaining two intervals. The subject was asked to respond which interval was different. Subjects were instructed that the loudness of each interval might vary and to ignore loudness differences. Each test run contained 5 reference-probe comparisons for each probe; the reference-probe comparisons were randomized within each run. Three to six test runs were conducted for each condition, depending on subjects' availability for testing, resulting in a minimum of 15 and a maximum of 30 comparisons for each reference-probe combination; S1 and S4 completed 5 runs, S2 and S3 completed 6 runs, and S5 completed 3 runs. No trial-by-trial feedback as to the correctness of the response was provided. The test order for the different single- and multi-channel stimuli was randomized within and across subjects. In Experiment 1, AMFD was measured for similarly loud single- and multi-channel AM stimuli for both the wide and narrow configurations. In Experiment 2, AMFD was measured for single- and multi-channel AM stimuli using the same summation-adjusted current levels for each component channel, whether tested in a single- or multi-channel context.

Results

Loudness balancing of single- and multi-channel non-AM stimuli

Figure 4.2 shows the current level adjustment needed to balance the loudness of the multi-channel non-AM stimuli to the single-channel non-AM reference (electrode 10 at 50% DR). The current level adjustment was calculated as the difference (in dB) between the single-channel reference and the multi-channel stimulus. Four out of the five subjects (S2 – S5) exhibited substantial multi-channel loudness summation (3.6 - 6.0 dB), while subject S1 exhibited less summation (1.4 – 2.0 dB). The mean level adjustment was 3.6 dB and 4.3 dB for the wide and narrow electrode combinations, respectively. Four of the 5 subjects exhibited greater multi-channel loudness summation for the narrow than for the wide configuration. A one-way repeated measures analysis of variance (RM ANOVA), with electrode configuration as the dependent factor (wide or narrow) and subject as the random/blocking factor, showed no significant effect of electrode configuration [$F(1,4) = 2.95$, $p = 0.161$]; note that power was low (0.19), due to the low number of subjects. This is in agreement with findings by McKay et al. (2001), who found that loudness summation was not significantly affected by distribution of electrodes within the multi-channel stimulus.

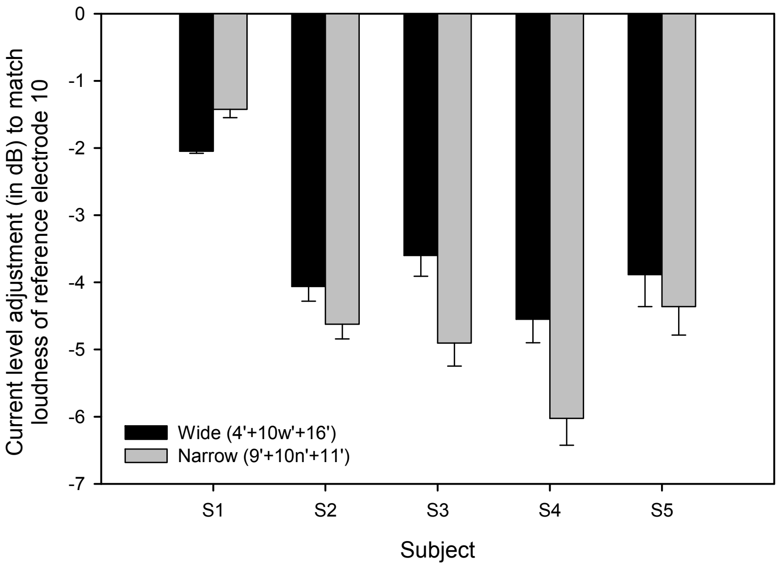

Figure 4.2. Loudness balancing between single- and multi-channel non-AM stimuli. The black and gray bars show the current level adjustments (in dB) needed to equate loudness to single-channel reference (electrode 10 at 50% DR) for the wide and narrow multi-channel configurations, respectively. The error bars show 1 standard error.

Experiment 1: AMFD with similarly loud single and multiple channels

Figure 4.3 shows AMFD (in percent correct) for similarly loud single- and multi-channel AM stimuli in the wide configuration, as a function of $\Delta F/F$. Due to multi-channel loudness summation, the current levels for the single-channel AM stimuli were higher than those for the multi-channel AM stimuli. The open circles show multi-channel data and the filled symbols show single-channel data. The data were fit with sigmoid functions using Sigmaplot 11.0 (Systat Software Inc). In most cases, AMFD with single- and multi-channel stimuli were quite similar. For subject S3, AMFD was somewhat better with multiple than with single channels. For subject S5, AMFD with the multiple channels was markedly poorer than with single channels. In most cases, AMFD was well above chance level when ΔF/F was greater than 0.1.

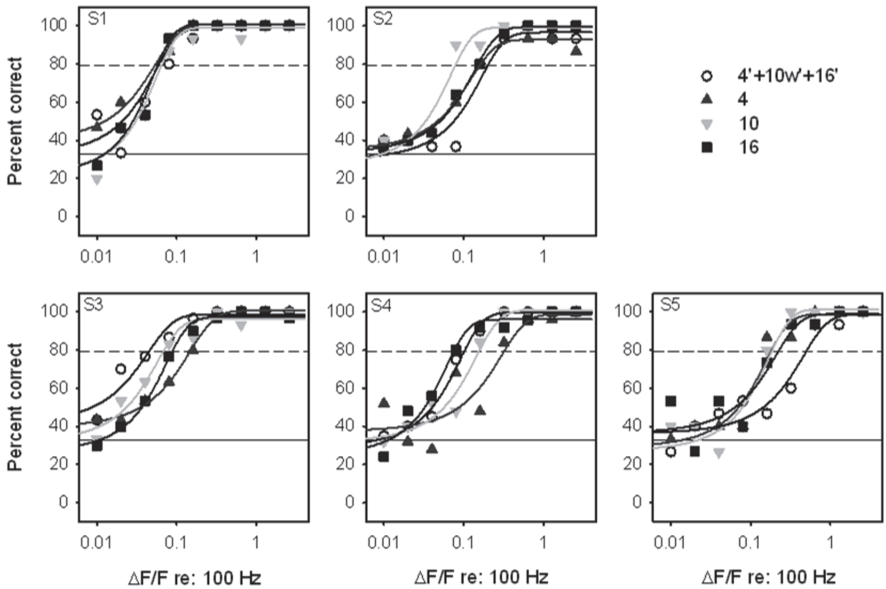

Figure 4.3. AMFD for the wide electrode configuration with similarly loud single- and multi-channel AM stimuli. Each panel shows individual subject data. The open circles show multi-channel AMDT data, and the filled upward triangles, downward triangles, and squares show single-channel data for the basal, middle, and apical electrodes, respectively. The solid lines through the data show sigmoid fits. The dashed horizontal line shows threshold (79.4% correct) and the solid horizontal line shows chance level (33.3% correct).

Figure 4.4 shows AMFD (in percent correct) for similarly loud single- and multi-channel AM stimuli in the narrow configuration, as a function of ΔF/F. Again, AMFD thresholds with single or multiple channels were quite similar, and were more similar than observed with the wide electrode configuration. Again, AMFD was well above chance level when ΔF/F was greater than 0.1.

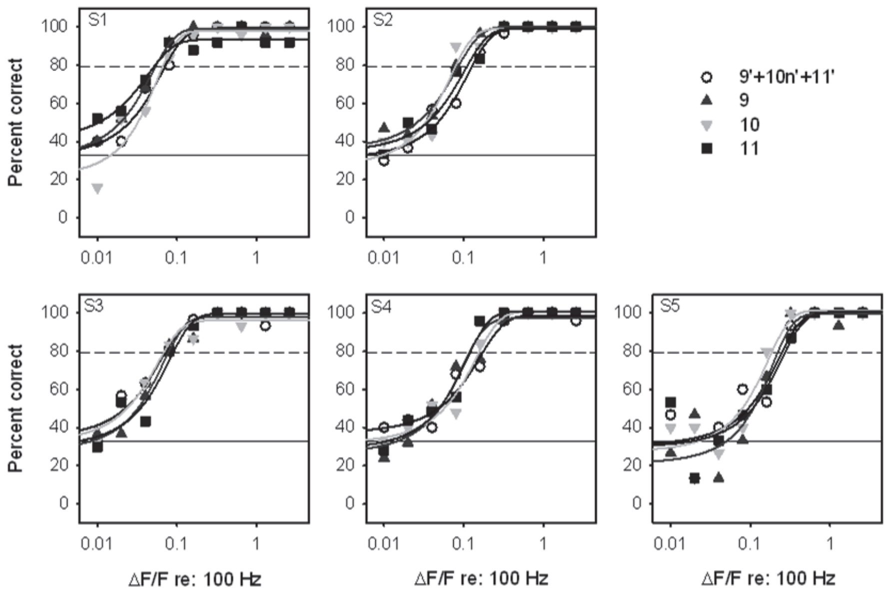

Figure 4.4. AMFD for the narrow electrode configuration with similarly loud single- and multi-channel AM stimuli. Each panel shows individual subject data. The open circles show multi-channel AMDT data, and the filled upward triangles, downward triangles, and squares show single-channel data for the basal, middle, and apical electrodes, respectively. The solid lines through the data show sigmoid fits. The dashed horizontal line shows threshold (79.4 % correct) and the solid horizontal line shows chance level (33.3% correct).

Linear interpolations of the sigmoid functions shown in Figures 4.3 and 4.4 were used to estimate the ΔF/F that corresponds to 79.4 % correct; this threshold is sometimes used for adaptive measurements of AMFD (3-down/1-up; Levitt, 1971). Figure 4.5 shows ΔF/F at threshold for individual subjects. The left and right panels show data for the wide and narrow combinations, respectively. As in Figures 4.3 and 4.4, the single- and multi- channel AM stimuli were similarly loud. In general, ΔF/F at threshold was quite similar across single- and multi-channel AM stimuli, with the exception of S5 who exhibited a highly elevated multi-channel threshold in the wide configuration. Absolute ΔF/F at threshold also varied across subjects. Multi-channel ΔF/F at threshold values ranged from 0.05 (S3, wide configuration) to 0.71 (S5, wide configuration), and single-channel threshold values ranged from 0.05 (S1, electrode 9) to 0.32 (S4, electrode 4). One-way RM ANOVAs were performed on the data in Figure 4.5, with stimulus (multi-channel and the three single channels) as the dependent factor and subject as the random/blocking factor. Because data were not normally distributed, a one-way RM ANOVA was performed on ranked data for the wide configuration. Results showed no significant effect of stimulus (Chi-square = 0.600 with 3 degrees of freedom; p = 0.896). For the narrow configuration, data were normally distributed. Results showed no significant effect of stimulus [$F(3,12) = 1.98$, $p = 0.170$].

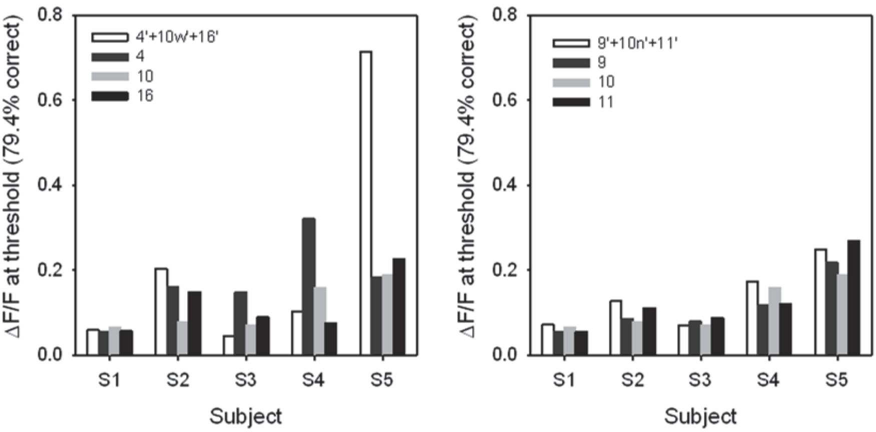

Figure 4.5. ΔF/F at threshold (79.4% correct) for individual subjects, for similarly loud single- and multi-channel AM stimuli. The left panel shows the wide electrode configuration and the right panel shows the narrow electrode configuration. The open bars show multi-channel data and the filled bars show single-channel data.

Figure 4.6 shows mean percent correct across all probe modulation frequencies for the wide (left panel) and narrow combinations (right panel), for single- and multi-channel AM stimuli. For multi-channel AM stimuli, mean values ranged from 57% correct (S5, wide configuration) to 86% correct (S3, wide configuration). For single-channel AM stimuli, mean values ranged from 64% correct (S5, electrode 9) to 83% correct (S1, electrode 9). One-way RM ANOVAs were performed on the data shown in Figure 4.6, with stimulus (multi-channel and the three single channels) as the dependent factor and subject as the random/blocking factor. There was no significant effect of stimulus on mean percent correct for the wide [$F(3,12) = 0.20$, $p = 0.893$] or narrow configurations [$F(3,12) = 0.06$, $p = 0.979$]. Note that in both these analyses, power was very low (alpha = 0.05).

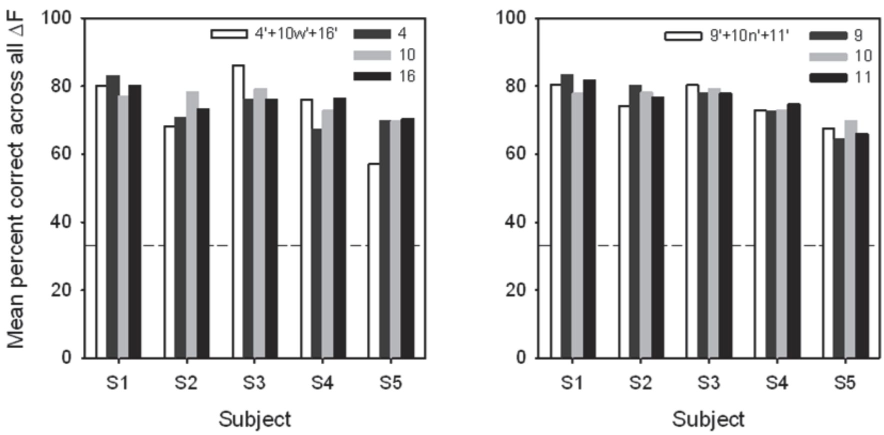

Figure 4.6. Mean percent correct AMFD across all probe modulation frequencies for individual subjects, for similarly loud single- and multi-channel AM stimuli. The left panel shows the wide electrode configuration and the right panel shows the narrow electrode configuration. The open bars show multi-channel data and the filled bars show single-channel data. The dashed line shows chance performance level (33.3% correct).

Experiment 2: AMFD with single or multiple channels using the same summation-adjusted current levels for the component channels

Figure 4.7 shows AMFD (in percent correct) for the wide configuration as a function of ΔF/F. The open circles show multi-channel data (same data is shown in Fig. 4.3) and the filled symbols show single-channel data. Note that the current levels for each component electrode were the same whether for single- or multi-channel AM stimuli and that the multi-channel AM stimuli were substantially louder than the single-channel AM stimuli. With the exception of subject S1, multi-channel AMFD was much better than single-channel AMFD for all subjects.

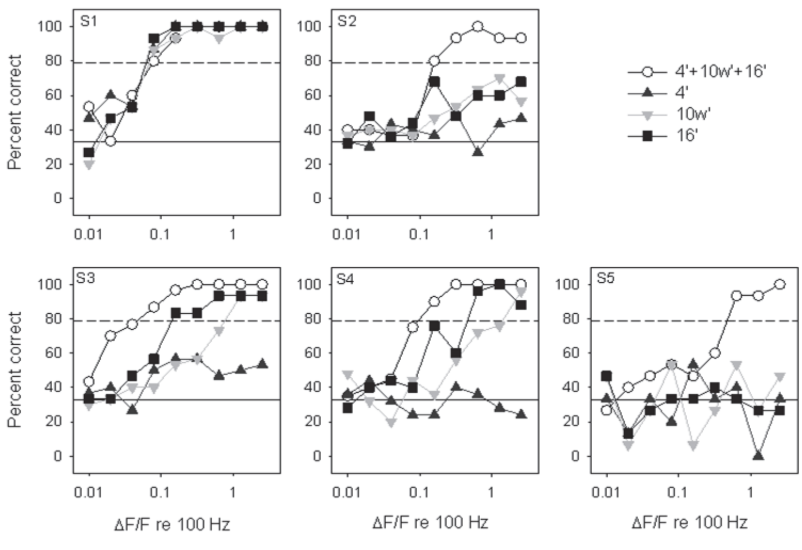

Figure 4.7. AMFD for the wide electrode configuration for single- and multi-channel AM stimuli using summation-adjusted current levels. Each panel shows individual subject data. The open circles show multi-channel AMDT data, and the filled upward triangles, downward triangles, and squares show single-channel data for the basal, middle, and apical electrodes, respectively. Because there was no adjustment for multi-channel loudness summation, multi-channel AM stimuli were louder than single-channel AM stimuli. The dashed horizontal line shows threshold (79.4 % correct) and the solid horizontal line shows chance level (33.3% correct).

Similar to Figure 4.7, Figure 4.8 shows AMFD (in percent correct) for the narrow configuration as a function as a function of ΔF/F. The open circles show multi-channel data (same data is shown in Fig. 4.4) and the filled symbols show single-channel data. Similar to the wide configuration, multi-channel AMFD with the narrow configuration was much better than single-channel AMFD for all subjects except S1. For subjects S2 and S4, single-channel AMFD was near chance level at all modulation frequencies.

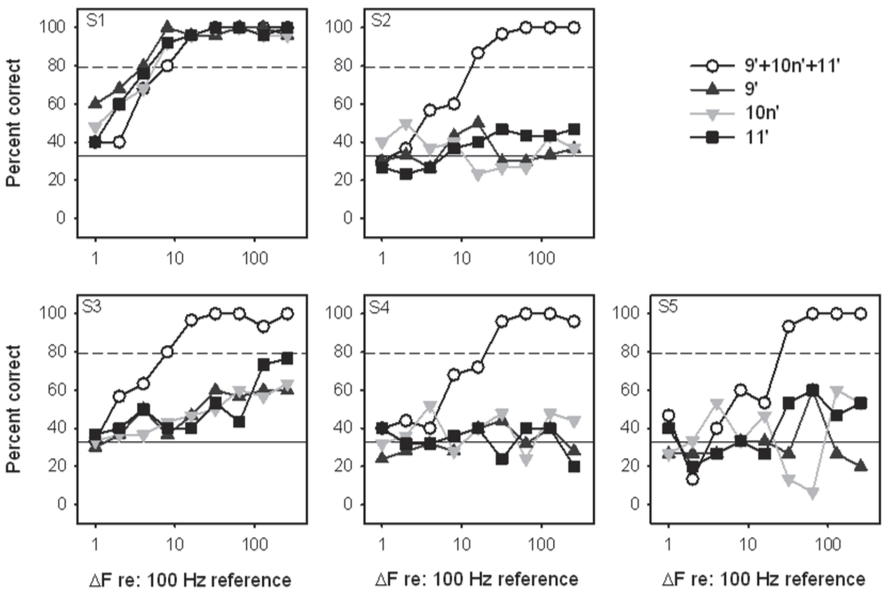

Figure 4.8. AMFD for the narrow electrode configuration for single- and multi-channel AM stimuli using summation-adjusted current levels. Each panel shows individual subject data. The open circles show multi-channel AMDT data, and the filled upward triangles, downward triangles, and squares show single-channel data for the basal, middle, and apical electrodes, respectively. Because there was no adjustment for multi-channel loudness summation, multi-channel AM stimuli were louder than single-channel AM stimuli. The dashed horizontal line shows threshold (79.4% correct) and the solid horizontal line shows chance level (33.3% correct).

Figure 4.9 shows mean percent correct across all probe modulation frequencies for the wide (left panel) and narrow combinations (right panel), for single- and multi-channel stimuli. The multi-channel data are the same as in Figure 4.6. With the exception of subject S1, mean percent correct AMFD was much better with multiple than with single channels. For multi-channel stimuli, mean values ranged from 57% correct (S5, wide configuration) to 86% correct (S3, wide configuration). For single-channel AM stimuli, mean values ranged from 30% correct (S5, electrode 9) to 88% correct (S1, electrode 4). One-way RM ANOVAs were performed on the data shown in each panel, with stimulus (multi-channel and the three single channels) as the dependent factor and subject as the random/blocking factor. For the wide configuration, there was a significant effect of stimulus on mean AMFD [$F(3,12) = 13.1$, $p < 0.001$]. Post-hoc Bonferroni pairwise comparisons showed that AMFD with 4'+10w'+16' was significantly better than with 4' or 10w' ($p < 0.05$), and significantly better with 16' than with 4' ($p < 0.05$). There were no significant differences among the remaining stimuli ($p > 0.05$). Because the distribution was not normal, a one-way RM ANOVA was performed on ranked data for the narrow configuration. There was a significant effect of stimulus on mean AMFD (Chi-square = 8.28 with 3 degrees of freedom, $p = 0.041$). Post-hoc pairwise comparisons (Tukey) showed that AMFD with 9'+10n'+11' was significantly better than with 9' ($p < 0.05$); there were no significant differences among the remaining stimuli ($p > 0.05$). A paired t-test showed no significant difference in mean multi-channel AMFD between the wide and narrow configurations ($p = 0.728$).

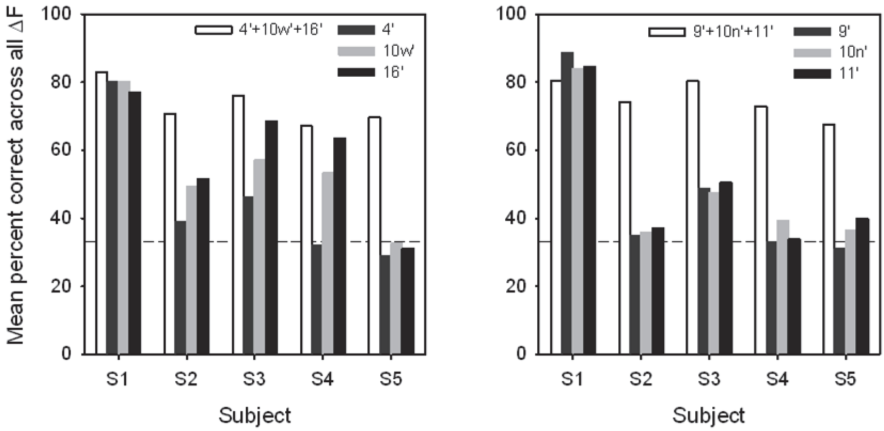

Figure 4.9. Mean percent correct AMFD across all probe modulation frequencies for data shown in Figures 4.7 and 4.8. The left panel shows the wide electrode configuration and the right panel shows the narrow electrode configuration. The open bars show multi-channel data and the filled bars show single-channel data. Because there was no adjustment for multi-channel loudness summation, multi-channel AM stimuli were louder than single-channel AM stimuli. The dashed line shows chance performance level (33.3% correct).

Discussion

There was no significant effect of the distribution of component channels in the multi-channel stimuli, contrary to the hypothesis that widely spaced channels would offer an advantage over narrowly spaced channels. When single- and multi-channel AM stimuli were similarly loud, there was no significant difference in AMFD, contrary to the hypothesis that the reduced current levels needed to accommodate multi-channel loudness summation would negatively affect multi-channel AMFD. With no adjustment for multi-channel loudness summation, AMFD was better with multiple channels than with any of the component single channels, consistent with our hypothesis. Below we discuss the results in greater detail.

Effects of loudness and multi-channel summation on single-and multi-channel AMFD

In Experiment 2, AMFD was measured using the same summation-adjusted current levels and the same range of modulation (in dB) on each component channel, whether tested in the single- or multi-channel condition. Because of multi-channel loudness summation, the multi-channel AM stimuli were generally louder than the single-channel AM stimuli. AMFD was much better with multiple channels than with any of the single component channels (see Figs. 4.7 and 4.8). This finding is in agreement with Geurts and Wouters (2001). It is unclear whether this multi-channel advantage is due to coherent envelope information delivered to multiple channels or to increased loudness. The single-channel data shown in Figures 4.3 and 4.4 may provide some insight. When the single-channel current levels were increased to match the loudness of the multi-channel stimuli, performance greatly improved. While this difference in single-channel AMFD thresholds may be due to current level, loudness also increased with level. Combined with the multi-channel data, this suggests that loudness, which increases with current level or with the number of channels (as well as with the cumulative number of pulses), may play a strong role in AMFD, whether with single or multiple channels.

One concern with the single-channel AMFD thresholds shown in Figures 4.7 and 4.8 is the potentially poor temporal envelope perception due to the reduced current levels. As shown in Table 4.2 and illustrated in Figure 4.1, the minimum AM current levels for summation-adjusted single channels were lower than the original single-channel thresholds. Given these reduced reference current levels, the large AM depth may have not have been sufficient to support AMFD. As such, the perceptual range of modulation was likely much reduced for the summation adjusted single-channel AM stimuli than for the multi-channel AM stimuli. It is also possible that the ± 1 dB level roving may have been a stronger cue across intervals than differences in AM frequency, contributing to poor AMFD.

Regardless of the source of poor AMFD with the summation-adjusted single channels, multi-channel loudness summation contributed strongly to the multi-channel advantage in AMFD. With any of the 3 summation-adjusted single AM channels, AMFD was often near chance level. When these channels were combined, AMFD was sharply improved. This may have been due to better perception of the AM range or to stronger perception of AM frequencies than loudness differences across intervals.

Note that subject S1 exhibited a different pattern of results than the other subjects (Figs. 4.7 and 4.8), as AMFD was similar for single- and multi-channel AM stimuli with the summation-adjusted current levels. As shown in Figure 4.2, subject S1 also exhibited much less multi-channel loudness summation than the other subjects. As such, there was less current adjustment for the single-channel AM stimuli shown in Figures 4.7 and 4.8. Consequently, single-channel AMFD was quite similar with or without the summation adjustment (i.e., the single-channel data in Fig. 4.3 versus Fig. 4.7, and Fig. 4.4 versus Fig. 4.8).

In Experiment 1, there were no significant differences among similarly loud single- and multi-channel AMFD. This highlights the importance of loudness on AMFD, rather than the distribution of envelope information in the cochlea. This finding is different from that of Geurts and Wouters (2001), who found better AMFD with multiple than with single AM channels. Several factors may contribute to these different findings. In Geurts and Wouters (2001), there was no adjustment for multi-channel loudness summation, and the modulation depth was considerably lower than in the present study. Stimuli were delivered through a research interface in the present study that allowed precise control of stimulation parameters, versus the experimental speech processors used in Geurts and Wouters (2001). Also, many more modulation frequencies were compared to the reference frequency in the present study than in Geurts and Wouters (2001), who only compared 150 Hz to 180 Hz ($\Delta F/F = 0.2$). The present data suggest no advantage in

AMFD for multiple AM channels over single AM channels when AM stimuli are similarly loud, at least for the AM depth and frequencies tested.

The effect of channel distribution on multi-channel AMFD

The distribution of channels did not significantly affect multi-channel AMFD thresholds. In Geurts and Wouters (2001), three adjacent electrodes were selected for multi-channel AM stimuli, similar to the narrow spacing in the present study. The narrow configuration targeted a limited region of neurons, for which single-channel AMFD thresholds would be expected to be more similar than for the wide configuration. If multi-channel AMFD thresholds were measured at lower overall loudness levels, some effect of electrode distribution may have emerged. The present findings are also in agreement with single-channel AMFD data from Green et al. (2012), who found no significant effect of carrier pulse rate when stimuli were presented at the same percent DR (and, presumably, at similar loudness). This suggests that the total number of pulses, whether delivered to a single channel or distributed across multiple channels, did not significantly affect AMFD thresholds, provided stimuli were similarly loud.

The lack of effect for the distribution of channels is somewhat in agreement with previous multi-channel MDI CI studies. Different from the present AMFD task in which coherent modulation was delivered to multiple channels, MDI measures detection or discrimination of one modulation frequency in the presence of another modulation frequency presented to the same or different channel. The spacing between electrodes is typically varied to explore the effect of overlapping neural populations on MDI. Richardson et al. (1998) found larger MDI for narrowly spaced than for widely spaced electrodes, suggesting that multi-channel envelope processing may depend on the degree of neural overlap among channels. However, Chatterjee (2003) found no clear effect of masker-probe separation for modulation masking (i.e., the difference in

MDI between a steady-state masker and an envelope masker with equivalent peak amplitudes). Chatterjee and Oba (2004) similarly found no clear effect of masker-probe separation for modulation masking. Kreft et al. (2013) found significant interference on AMFD when the masker and probe electrodes were widely separated. While the listening tasks may be different between the present and these previous studies, all seem to point toward a more centrally mediated envelope processing.

Differences between multi-channel MDT and AMFD

The present single- and multi-channel AMFD results are somewhat in contrast with previous amplitude modulation detection findings. In Galvin et al. (2014), when measured at the same loudness, multi-channel MDTs were significantly poorer than single-channel MDTs for the component electrodes used in the multi-channel stimuli. The authors argued that the reduced per-channel current levels needed to accommodate multi-channel loudness summation resulted in poorer multi-channel MDTs. Previous studies have shown that single-channel MDTs are highly level dependent, especially in the lower portion of the DR (Donaldson and Viemeister, 2001; Galvin and Fu, 2005, 2009; Pfingst et al., 2007). In this study, there was no significant difference between similarly loud single- and multi-channel AMFD thresholds, despite differences in current level between single- and multi-channel AM stimuli. Previous CI studies have shown that single-channel AMFD is level dependent (Luo et al., 2008; Kreft et al., 2010; Chatterjee and Ozerbut, 2011). The present data also showed that the mean percent correct in single-channel AMFD was better with higher current levels (Fig. 4.6 versus Fig. 4.9). Single-channel AMFD was generally poor with the lower, summation-adjusted current levels; when these channels were combined, AMFD sharply improved. The present results suggest that AMFD seems to depend more on the loudness of the stimulus (which varies with level, rate, or the number of channels), while MDT seems to depend more on the current level.

Differences in the listening task and stimuli – detecting modulation given weak envelope information (due to small AM depth and/or low presentation level) for MDT versus detecting a difference in AM frequency given strong envelope information (due to large AM depth and/or high presentation level) for AMFD – may also explain differences in the pattern of results between MDT and AMFD. Different mechanisms may also come into play for modulation detection and modulation frequency discrimination. When discriminating between AM and non-AM stimuli with the same reference amplitude, there are potential loudness cues associated with the peak amplitude of the AM stimulus (McKay and Henshall, 2010; Fraser and McKay, 2012). Given sufficient modulation depth and/or presentation level, such peak AM loudness cues do not seem to play a strong role in modulation frequency discrimination.

Limitations to the present study

In this study, a 3AFC discrimination task was used ("which interval is different?"), as in Chatterjee and Peng (2008), Chatterjee and Ozerbut (2011), Luo et al., (2008, 2010), Deroche et al. (2012, 2014). Other AMFD studies in CI users have used a 2AFC procedure (Geurts and Wouters, 2001; Green et al., 2012; Kreft et al., 2011, 2013). In the 3AFC procedure, there is no assumption of regarding the perceptual difference between the reference and probe modulation frequencies (e.g., pitch, timbre, loudness, or some other quality). These perceptual qualities may differ greatly, depending on the reference modulation frequency, as low (<50 Hz) and high frequencies (>300 Hz) may not give strong pitch percepts. In the present study, given the 100 Hz reference AM frequency (which would likely elicit a fairly strong pitch percept), AMFD thresholds may have on pitch differences or some other quality, such as loudness. The loudness balancing, roving, and instructions to ignore loudness differences across intervals presumably reduced the contribution of loudness cues to the present AMFD thresholds. In Experiment 1, the range of AMFD thresholds was comparable to those found in previous studies that used a 2AFC procedure (e.g., Green et al., 2012; Kreft et al., 2012, 2013).

Loudness balancing was performed using non-AM pulse trains, rather than the AM stimuli used for AMFD. Given that current levels were swept for equal loudness at 10%, 50% and 100% DR, it seems unlikely that there would be great differences in loudness at, for example, 30% DR or 70% DR. It is possible that the loudness of AM stimuli with 100% AM depth may have differed across single channels and/or AM rates, but the effect of AM on loudness would likely be consistent across single channels. If there were indeed loudness differences across single channels when AM was applied, the current level roving (± 1 dB independent level roving for each channel in the multi-channel AM stimuli; ± 1 dB global level roving for each of the 3 intervals during each trial of AMFD) helped to reduce such loudness differences.

For similarly loud single- and multi-channel AM stimuli, the overall loudness was not explicitly measured. However, subjects did not report that the AM stimuli were too soft or too loud, although the summation-adjusted single-channel AM stimuli were substantially softer. It is unclear how overall loudness might affect single- and multi-channel AMFDs, assuming sufficient envelope cues for all stimuli. Such an experiment would require sufficient modulation depth (e.g., 20% of reference amplitude, depending on the current/loudness level), but not necessarily the maximal modulation depth used in this and other studies (e.g., Kreft et al., 2010, 2013).

In Experiment 2, the poor AMFD with the summation-adjusted single-channel AM stimuli were presumably due to low current levels, which could not support AMFD even with the large AM depth used. As shown in Table 4.2 and Figure 4.1, minimum AM current levels would likely have been inaudible. Another approach would be to use a smaller AM depth that would ensure stimulation above single-channel threshold, even after reducing current levels to accommodate multi-channel loudness summation. In such a design, it would be necessary to keep the range of modulation (in dB) constant across stimuli to examine the effects of multi-channel loudness summation on AMFD. Most likely, this approach would produce similar findings as in the present study: poor single-channel AMFD due to low current levels can be improved with multi-channel stimulation, due to the increased loudness associated with multi-channel summation.

Clinical implications

Clinical fitting of CIs must accommodate multi-channel loudness summation. The present results suggest that AMFD with multiple channels is largely unaffected by this accommodation, provided sufficient modulation depth and/or presentation levels. However, modulation detection is negatively affected by the reduced current levels needed to accommodate multi-channel loudness summation (Galvin et al., 2014). Amplification of envelope information, whether by increasing the modulation depth (envelope expansion) or by increasing current levels, may improve perception of envelope cues. There is likely to be a trade-off between amplification of envelope cues and increased noise levels for some listening environments. Selectively amplifying envelope information that is likely to be weakly represented (e.g., consonant information presented to basal electrodes) may help improve perception of envelope cues without globally increasing noise levels. The present study suggests that delivery of coherent envelope information to multiple channels may also improve perception of envelope cues, primarily due to increased loudness associated with multi-channel summation.

Conclusions

Single- and multi-channel AMFD thresholds were measured relative to 100 Hz AM in 5 CI subjects, with and without current level adjustments for multi-channel loudness summation. The electrical range of modulation was constant across AM stimuli, but the perceptual range of modulation was most likely reduced for the quieter, summation-adjusted single-channel AM stimuli. Key findings include:

1. When single- and multi-channel AM stimuli were similarly loud, there was no significant difference in AMFD thresholds. This finding is somewhat different than for modulation detection (Galvin et al., 2014), in which multi-channel MDTs were significantly poorer than those for similarly loud single channels.

2. When the same summation-adjusted current levels were used for the component channels in single- or multi-channel AM stimuli, AMFD was significantly better with multiple channels than with any of the single component channels. The poor single-channel AMFD may have been due to the lower current level, poor perception of the modulation range (which included substantial sub-audible stimulation) or to level roving (which may have obscured differences in AM frequency).

3. There was no significant effect of the distribution of electrodes for multi-channel AMFD thresholds.

4. The present results suggest that loudness, whether due to current level or the number of channels stimulated, may play a strong role in modulation frequency discrimination.

Acknowledgements

We thank all of the CI subjects for their tireless participation. We also thank Monita Chatterjee for very helpful comments. John Galvin, Sandy Oba, and Qian-Jie Fu were supported by NIH grant DC004993. Deniz Baskent was supported by VIDI grant 016.096.397 from the Netherlands Organization for Scientific Research (NWO) and the Netherlands Organization for Health Research and Development (ZonMw), and a Rosalind Franklin Fellowship from University of Groningen, University Medical Center Groningen.

Chapter 5

Envelope interactions in multi-channel amplitude modulation frequency discrimination by cochlear implant users

John J. Galvin III
Sandy Oba
Monita Chatterjee
Qian-Jie Fu
Deniz Başkent

This chapter is a modified version of a paper published in:
PLoS One, 10(10):e0139546

Abstract

Previous cochlear implant (CI) studies have shown that single-channel amplitude modulation frequency discrimination (AMFD) can be improved when coherent modulation is delivered to additional channels. It is unclear whether the multi-channel advantage is due to increased loudness, multiple envelope representations, or to component channels with better temporal processing. Measuring envelope interference may shed light on how modulated channels can be combined.

In this study, multi-channel AMFD was measured in CI subjects using a 3-alternative forced-choice, non-adaptive procedure ("which interval is different?"). For the reference stimulus, the reference AM (100 Hz) was delivered to all 3 channels. For the probe stimulus, the target AM (101, 102, 104, 108, 116, 132, 164, 228, or 256 Hz) was delivered to 1 of 3 channels, and the reference AM (100 Hz) delivered to the other 2 channels. The spacing between electrodes was varied to be wide or narrow to test different degrees of channel interaction.

Results showed that CI subjects were highly sensitive to interactions between the reference and target envelopes. However, performance was non-monotonic as a function of target AM frequency. For the wide spacing, there was significantly less envelope interaction when the target AM was delivered to the basal channel. For the narrow spacing, there was no effect of target AM channel. The present data were also compared to a related previous study in which the target AM was delivered to a single channel or to all 3 channels. AMFD was much better with multiple than with single channels whether the target AM was delivered to 1 of 3 or to all 3 channels. For very small differences between the reference and target AM frequencies (2-4 Hz), there was often greater sensitivity when the target AM was delivered to 1 of 3 channels versus all 3 channels, especially for narrowly spaced electrodes.

Besides the increased loudness, the present results also suggest that multiple envelope representations may contribute to the multi-channel advantage observed in previous AMFD studies.

The different patterns of results for the wide and narrow spacing suggest a peripheral contribution to multi-channel temporal processing. Because the effect of target AM frequency was non-monotonic in this study, adaptive procedures may not be suitable to measure AMFD thresholds with interfering envelopes. Envelope interactions among multiple channels may be quite complex, depending on the envelope information presented to each channel and the relative independence of the stimulated channels.

Introduction

Given the limited spectral resolution of cochlear implants (CIs), temporal envelopes convey important speech cues for CI users. As such, CI users' temporal processing capabilities may contribute to their speech understanding. Single-channel amplitude modulation detection (AMD) has been extensively measured in CI users (Donaldson and Viemeister, 2000; Chatterjee and Robert, 2001; Fu, 2004; Galvin and Fu, 2005, 2009; Pfingst et al., 2007; Chatterjee and Yu, 2010; Chatterjee and Ozerbut, 2011; Fraser and McKay, 2012; Green et al., 2013) and has been correlated with CI users' speech performance (Cazals et al., 1994; Fu, 2004). Similarly, CI users' single-channel amplitude modulation frequency discrimination (AMFD) has been correlated with CI users' prosody perception (Chatterjee and Peng, 2008: Deroche et al., 2012, 2014) and tonal language perception (Luo et al., 2008). However, in everyday listening with clinical processors, CI users must process multiple temporal envelopes.

Because of multi-channel loudness summation, current levels on individual channels must often be reduced in clinical processors to provide a comfortable operating range (McKay et al., 2001, 2003; Drennan et al., 2006). Single-channel AMD and AMFD have been shown to depend on current level (Donaldson and Viemeister, 2000; Chatterjee and Robert, 2001; Fu, 2004; Galvin and Fu, 2005, 2009; Pfingst et al., 2007; Luo et al., 2008; Chatterjee and Yu, 2010; Chatterjee and Ozerbut, 2011; Green et al., 2013). At the same loudness, multi-channel AMD has been shown to be significantly poorer than single-channel AMD, due to the reduced current levels needed to compensate for multi-channel loudness summation (Galvin et al., 2014). However, at the same loudness, single- and multi-channel AMFD thresholds have not been shown to be significantly different (Galvin et al., 2015), despite differences in current level. Previous studies have also shown that single-channel AMD thresholds can vary across stimulation site [6], though no clear effect of across-site variability has been shown for multi-channel AMD (Galvin et al., 2014). For AMFD, It is unclear how across-site variability may affect multi-channel perception. Thus, many factors may contribute to CI users' multi-channel temporal envelope processing: listening task (envelope detection vs. envelope frequency discrimination), current level, multi-channel loudness summation, across-site differences in temporal processing, etc.

One issue when measuring AMD is the contribution of potential loudness cues associated with amplitude modulated (AM) stimuli (McKay and Henshall, 2010). As such, it is unclear whether AMD represents CI users' temporal processing limits or their sensitivity to loudness cues in AM stimuli. While there are methods to limit the contribution of potential AM loudness cues (Chatterjee and Ozerbut, 2011; Fraser and McKay, 2012; Galvin et al., 2013), such current level adjustments and/or roving may introduce too much variability in AMD thresholds. As such, discrimination of AM frequency, rather than detection of AM, may better represent temporal processing limits of CI users. AMFD is typically measured using AM depths that are well above AMD threshold. Loudness differences across AM frequency are inconsistent and typically small [24]. Accordingly, less current level compensation and jitter is needed when measuring AMFD than for AMD, resulting in a potentially less "noisy" measure of CI users' temporal processing.

AMFD has been shown to be better when the target AM was delivered to multiple channels than to any of the single component channels used for the multi-channel stimuli (Geurts and Wouters, 2001; Galvin et al., 2015). As noted above, when single- and multi-channel stimuli are similarly loud and at a comfortably loud presentation level, no significant difference in AMFD was observed (Galvin et al., 2015). It is unclear how across-site variability might contribute to the multi-channel advantage in AMFD. When single-channel AMFD was measured at summation-adjusted current levels, performance was near chance-level (Galvin et al., 2015), obscuring across-site differences in performance. Thus, when multi-channel loudness summation is considered, it may be difficult to observe how channels are combined when discriminating coherent AM delivered to multiple channels.

Many previous studies have explored how competing envelopes may interfere with CI users' ability to detect or discriminate target AM. For AMD, significant amounts of "envelope masking" (the difference in AMD threshold between a modulated and steady state masker) have been observed even when the target AM channel is spatially remote from the masker channel (Chatterjee, 2003: Chatterjee and Oba, 2004). As such, central processes are thought to contribute strongly to CI users' temporal envelope perception. Similarly, AMFD thresholds have been shown to be greatly elevated in the presence of competing envelope information, even when the target and masker channels are spatially remote (Chatterjee and Ozerbut, 2009; Kreft et al., 2013). In general, CI users seem unable to segregate even large AM frequency differences between the target and masker channel. In these previous studies, presentation levels for the target AM channel were relatively high, thus ensuring good baseline single-channel AMD or AMFD thresholds. Also in these studies, there was typically no adjustment for multi-channel loudness summation. Because the multi-channel stimuli only contained 2 channels, and because of the relatively high presentation levels, multi-channel loudness summation would not be expected to significantly contribute to the pattern of results observed. However, when a larger number of channels are considered along with the attendant loudness summation, baseline single-channel thresholds at summation-adjusted levels would most likely be much poorer than observed in previous AMD or AMFD studies. Indeed, at summation adjusted levels, single-channel AMFD was recently shown to be at near chance-level (Galvin et al., 2015). And while widely spaced channels have been used in some previous studies (Chatterjee and Ozerbut, 2009; Kreft et al., 2013), there have been few comparisons of envelope interference between widely and narrowly spaced channels. If interference were to occur at the edges of the spread of excitation from multiple channels, less interference would be expected for widely spaced electrodes. At reduced summation-adjusted current levels, the spread of excitation would be less broad (Chatterjee and Shannon, 1998; Chatterjee et al., 2006), which might reduce channel interaction, especially for widely spaced channels. In these previous studies, it is also unclear how across-site differences in temporal processing may have contributed to the degree of interference between the masker and target channels, as temporal processing was not typically measured for masker channels. One previous AMD study showed no clear relationship between the envelope sensitivity of the masker channel and the amount of envelope masking produced by the masker channel (Chatterjee and Oba, 2007).

Taken together, results from these previous studies suggest that multi-channel envelope perception may affected by the information in each channel (coherent or competing AM), multi-channel loudness summation, across-site difference in temporal processing, and the spatial overlap in the spread of excitation from each component channel. In this study, AMFD was measured using multi-channel stimuli in which the target AM was delivered to 1 of 3 channels and the reference AM was delivered to the other 2 channels. The component channels were either widely or narrowly spaced to explore different degrees of channel interaction. The target AM channel was varied to explore across-site differences in temporal processing. To examine how AM discrimination was affected by the type of envelope information delivered to multiple channels, the present data were compared to those from a previous related study in which the target AM was delivered to a single channel or to all 3 channels (Galvin et al., 2015). In all cases, whether with single or multiple channels, AMFD data was compared using summation-adjusted current levels to explore temporal processing at the reduced current levels that might be used in clinical processors. Comparing AMFD with single and multiple channels at the same summation adjusted current levels provided an opportunity to examine the effects of loudness and the type of information delivered to each channel on AM discrimination.

Methods

Subjects

Five adult, post-lingually deafened CI users participated in this study. All were users of Cochlear Corp. devices and all had more than 2 years of experience with their implant device. Relevant subject details are shown in Table 5.1. All 5 subjects previously participated in a related AMFD study (Galvin et al., 2015). All subjects provided written informed consent prior to participating in the study, in accordance with the guidelines of the St. Vincent Medical Center Institutional Review Board (Los Angeles, CA), which specifically approved this study. All subjects were financially compensated for their participation.

Stimuli

Stimuli were similar to those used in Galvin et al. (2015). All stimuli were 300-ms biphasic pulse trains; the stimulation mode was monopolar, the stimulation rate was 2000 pulses per second (pps) per electrode, the pulse phase duration was 25 μs and the inter-phase gap was 8 μs. The relatively high stimulation rate was selected to encode the highest target AM frequency (356 Hz) and to approximate the cumulative stimulation rate used in some clinical processors. The spacing between electrodes was varied to represent different amounts of channel interaction; electrodes were either widely (El 4, 10, and 16) or narrowly spaced (EL 9, 10, and 11). The component electrodes of the multi-channel stimuli were optimally interleaved in time; the inter-pulse interval (between the offset of one pulse and the onset of the next) was 0.109 ms. All stimuli were presented via research interface (Wygonski and Robert (2002), bypassing subjects' clinical processors and settings; custom software was used to deliver the stimuli and to record subject responses.

Subject	Age at testing (yrs)	Age at implantation (yrs)	Duration of deafness (yrs)	Etiology	Device	Strategy
S1	70	60	23	Genetic	N24	ACE
S2	79	77	35	Otosclerosis	N5	ACE
S3	28	26	11	Acoustic Neuroma	Freedom	ACE
S4	67	59	20	Meniere's/ Otosclerosis	Freedom	ACE
S5	78	76	8	Unknown	N5	ACE

Table 5.1. CI subject demographics. N24 = Nucleus 24; N5 = Nucleus 5; ACE = Advanced combination encoder

Several steps were taken to determine the current levels for the component electrodes in the multi-channel stimuli and to ensure similar loudness across component electrodes and the wide and narrow spacing conditions, and are more fully described in Galvin et al. (2015). First, the dynamic range (DR) was estimated for single electrodes without AM. Absolute detection thresholds (Ts) were estimated using a "counting" method, as is sometimes used for clinical fitting of speech processors. During each threshold measurement, a number of pulse train bursts (between 2 and 5 bursts) were presented to the subject, who responded by reporting how many bursts were heard. Depending on the correctness of response, the current level was adjusted in 0.5 dB steps; the current level after 4 reversals was considered the threshold. Maximum acceptable loudness (MAL) levels were estimated by slowly increasing the current level (in 0.2 dB steps) for three pulse train bursts until reaching MAL. Threshold and MAL levels were averaged across a minimum of two runs, and the DR was calculated as the difference in current between MAL and T levels. After the initial DR estimation, all electrodes were swept for equal loudness at

10% DR, 50% DR, and at MAL (100% DR). During loudness sweeping, 300 ms pulse trains were delivered to each electrode in sequence (at either 10% DR, 50% DR or MAL, depending on the sweep), first from apex to base, and then from base to apex. The subject indicated which (if any) of the electrodes were louder or softer than the rest; the current level was adjusted to those electrodes as needed, and the electrodes were then re-swept for loudness. After making all adjustments, the final threshold, MAL and DR values for each electrode were recorded.

When the three component electrodes were combined using the above single-channel current levels, multi-channel stimulation would be expected to be substantially louder due to summation (McKay et al., 2001, 2003; Drennan et al., 2006). Multi-channel stimuli were loudness-balanced to a common single-channel reference (EL 10) presented at 50% DR. An adaptive two-alternative, forced-choice (2AFC), double-staircase procedure was used for loudness balancing (Jestead, 1980; Zeng and Turner, 1991). Stimuli were loudness-balanced without AM. The amplitude of the 3-channel probe was globally adjusted (final step size = 0.4 dB) according to subject response (2-down/1-up or 1-down/2-up, depending on the track), thereby adjusting the current level for each component electrode by the same ratio. For each run, the final 8 of 12 reversals in current level were averaged, and the mean of 2-3 runs was considered to be the loudness-balanced level. The mean current level reduction to the multi-channel stimuli across the wide and narrow combinations was 3.95 dB (range = 1.6 to 6.0 dB), relative to the single-channel reference. Refer to Galvin et al. (2015) for additional details regarding the loudness balance procedure, and for the amount of current level reduction needed to compensate for multi-channel loudness summation for each subject. Figure 5.1 shows the summation-adjusted DRs for widely spaced electrodes for subject S3. Note that the summation-adjusted current levels were well below the original single-channel T (solid lines) and MAL levels (dashed lines).

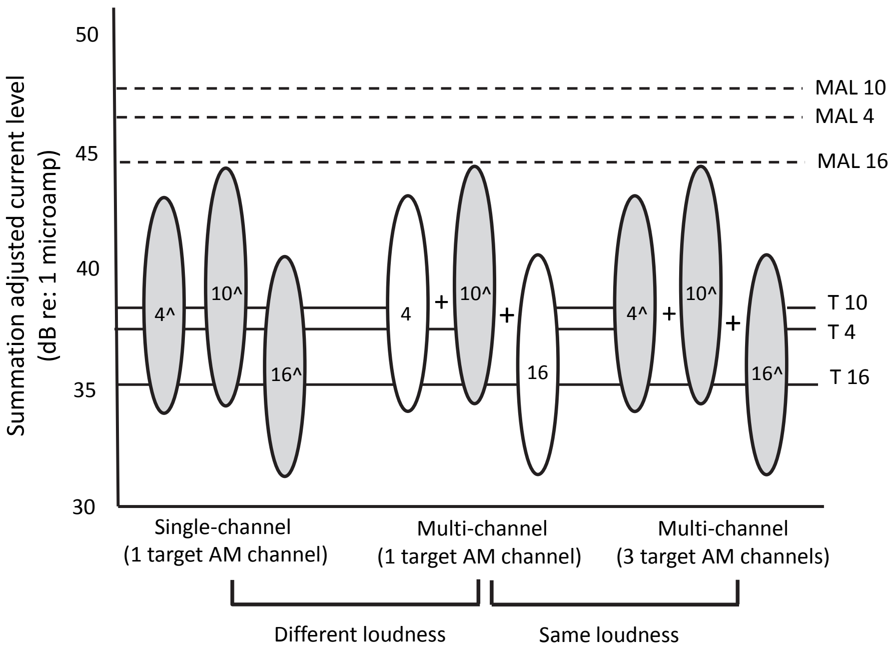

Fig. 5.1. Illustration of the summation-adjusted current levels and DRs used for subject S3. The solid and dashed lines show the original single-channel T and MAL levels before adjusting for multi-channel loudness summation, respectively. The ovals represent the summation-adjusted DRs, and also represent the AM depth used to measure AMFD (i.e., between the summation-adjusted T and MAL levels); the number within each oval indicates the electrode. The ovals on the left side of the figure show single-channel stimuli; the ovals in the middle and right side of the figure show multi-channel stimuli. The filled ovals indicate the channels that received the target AM frequency and the white ovals indicate the channels that received the reference AM frequency.

Table 5.2 shows the test electrodes for each subject and condition and the current levels for summation-adjusted T levels (minimum AM current level), MAL levels (maximum AM current level), DR (which corresponds to the range of AM), and 50% DR (which corresponds to the reference current level used to calculate AM depth). Because of the previous loudness sweeping with single electrodes, component electrodes were presumed to be similarly loud at the summation-adjusted T, MAL, and 50% DR current levels. When measuring multi-channel AMFD, the current levels of the component channels were independently roved by ±1 dB from trial to trial to reduce potential cross-channel loudness differences.

For the multi-channel stimuli, the basal, middle, and apical channels were sequentially interleaved in the sequentially interleaved. Sinusoidal AM was then applied to the multi-channel stimulus according to:

$$[f(t)][1 + m\sin(2 * \pi * fm * t)]$$

where *f(t)* is a steady-state pulse train, m is the modulation index, and *fm* is the modulation frequency. A 10-ms onset and offset was applied to all stimuli. The initial modulation phase was 180 degrees for all stimuli. For each channel, the modulation index was calculated relative to the reference current level (50% DR, in microamps) to target minimum and maximum current levels at T and MAL, respectively.

Throughout this paper, the caret symbol (^) indicates the channel that received the target AM. The reference AM frequency was 100 Hz; the target AM frequency was 101, 102, 104, 108, 116, 132, 164, 228, or 356 Hz. During AMFD testing, the reference stimulus contained the reference frequency delivered to all 3 channels. The probe stimulus contained the target AM frequency delivered to one channel and the reference AM frequency delivered to the other two channels. Figure 5.2 shows examples of the reference and probe stimuli. The envelope patterns are very similar between the 100 Hz reference and the 102 Hz target, but very different between the 100 Hz reference and the 132 Hz target. When the target AM was delivered to only 1 of 3 channels, there is very little difference in the 102 Hz temporal pattern compared to when the target AM was delivered to all 3 channels. However, the difference in the 132 Hz temporal pattern was quite large when the target AM was delivered to 1 of 3 channels or to all 3 channels.

			microamps				dB (re: 1 microamp)			
Subject	**Spacing**	**El**	**T**	**MAL**	**DR**	**50% DR**	**T**	**MAL**	**DR**	**50% DR**
S1	**Wide**	4	158	693	535	426	43.97	56.82	12.85	52.58
		10	155	816	661	485	43.80	58.23	14.44	53.72
		16	126	640	513	383	42.03	56.12	14.09	51.67
	Narrow	9	163	825	662	494	44.24	58.33	14.09	53.87
		10	166	877	711	522	44.42	58.86	14.44	54.34
		11	141	825	684	483	42.98	58.33	15.35	53.67
S2	**Wide**	4	84	228	144	156	38.48	47.14	8.66	43,85
		10	70	232	162	151	36.92	47.30	10.38	43.58
		16	71	212	140	142	37.08	46.52	9.44	43.02
	Narrow	9	69	221	152	145	36.81	46.90	10.09	43.24
		10	66	217	151	141	36.35	46.73	10.38	43.01
		11	65	209	145	137	36.20	46.42	10.23	42.74
S3	**Wide**	4	50	142	92	96	34.02	43.05	9.03	39.66
		10	54	161	108	107	34.57	44.15	9.58	40.62
		16	38	108	71	73	31.52	40.70	9.18	37.27
	Narrow	9	47	144	97	95	33.37	43.15	9.79	39.57
		10	46	139	93	92	33.26	42.84	9.58	39.31
		11	49	139	89	94	33.88	42.84	8.96	39.47
S4	**Wide**	4	65	209	145	137	36.20	46.42	10.23	42.73
		10	50	142	92	96	34.02	43.05	9.03	39.66
		16	54	161	108	107	34.57	44.15	9.58	40.62
	Narrow	9	111	312	201	212	40.94	49.89	8.95	46.52
		10	105	306	201	206	40.46	49.72	9.26	46.27
		11	105	354	249	230	40.46	50.98	10.53	47.22

Table 5.2. Summation-adjusted current levels. Values are shown for threshold (T), maximum acceptable loudness (MAL), dynamic range (DR), and 50% DR. The AM depth was between T and MAL (100% DR), and the reference current level was 50% DR.

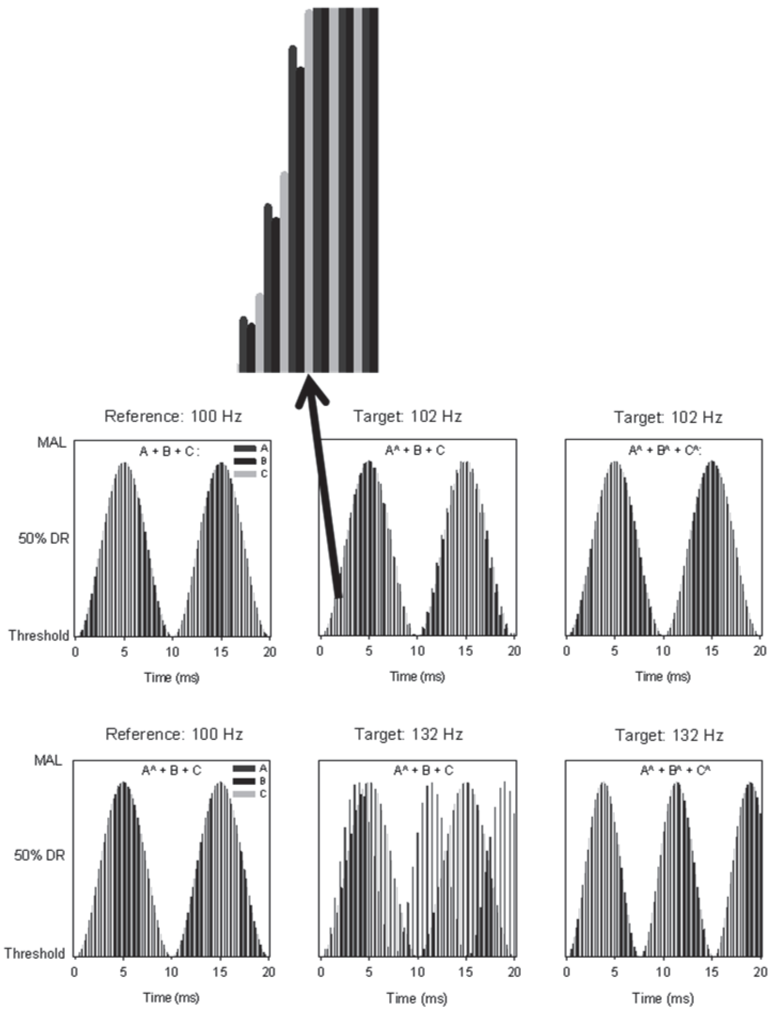

Fig. 5.2. Examples of experimental stimuli. The reference stimuli are shown in the left column and the probe stimuli are shown in the middle and right columns. The top row shows probe stimuli with the 102 Hz target AM frequency and the bottom row shows probe stimuli with the 132 Hz target AM frequency. The left column shows the reference AM frequency delivered to all 3 channels, the middle column shows the target AM frequency delivered to 1 of 3 channels (with the reference AM delivered to the other 2 channels) , and the right column shows the target AM frequency delivered to all 3 channels. The x-axis shows time (in ms). The y-axis shows the nominal summation-adjusted current levels. The figure accurately shows the timing of the pulse trains and order of interleaving over a 20 ms range.

Figure 5.3 illustrates the test conditions in terms of the electrode spacing. For the wide spacing, channels were expected to relatively independent; for the narrow spacing, channels were expected to interact.

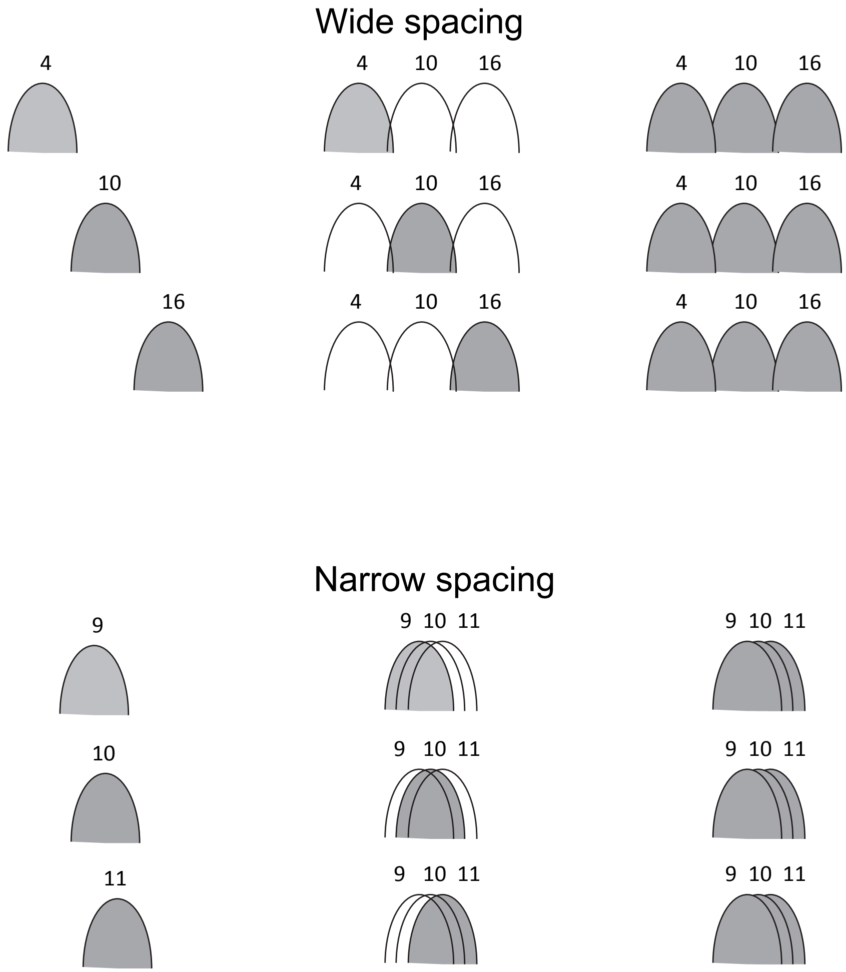

Fig. 5.3. Illustration of electrode spacing conditions. The wide spacing is shown at top and the narrow spacing is shown at bottom. The gray regions indicate the target AM frequency channels and the white regions indicate the reference AM frequency channels. The target AM was delivered to a single channel (left), 1 of 3 channels (middle), or to all 3 channels (right).

Procedure

A 3AFC non-adaptive procedure was used to measure AMFD ("which interval is different?"), as in Galvin et al. (2015). During each trial of the run, the probe stimulus (in which the target AM was delivered to 1 channel and the reference AM was delivered to the other 2 channels) was randomly assigned to 1 of the 3 intervals and the reference stimulus (in which the reference AM frequency was delivered to all 3 channels) was assigned to the remaining 2 intervals. The subject was asked to respond which interval was different. Because AM frequency may affect loudness given a fixed AM depth (Vandali et al., 2013), the current level in each interval was globally roved by ± 1 dB to protect against potential loudness differences across AM frequencies as in previous studies (Chatterjee and Ozerbut, 2011; Kreft et al., 2010, 2013; Galvin et al., 2015). This roving was in addition to the independent roving to current levels across channels in the multi-channel stimuli. Each test run contained 5 reference-probe comparisons for each probe; the reference-probe comparisons were randomized within each run. Three to six test runs were conducted for each condition, depending on subjects' availability for testing. No trial-by-trial feedback as to the correctness of the response was provided. The test order was randomized within and across subjects.

Results

Figure 5.4 shows percent correct AMFD for the wide spacing condition when the target AM was delivered to a single channel (black circles), 1 of 3 channels (white triangles) or to all 3 channels (black squares). The circle and square data are from Galvin et al. (2015). and are shown for comparison purposes. Performance was generally best when the target AM was delivered to all 3 channels. When the target AM was delivered to only a basal channel, with (white triangles) or without the additional 100-Hz reference channels (black circles), performance was generally poor. At relatively low target AM frequencies (102-104 Hz), there were several instances where multi-channel performance was better when the target AM was delivered to the middle or apical channels, rather than to all 3 channels.

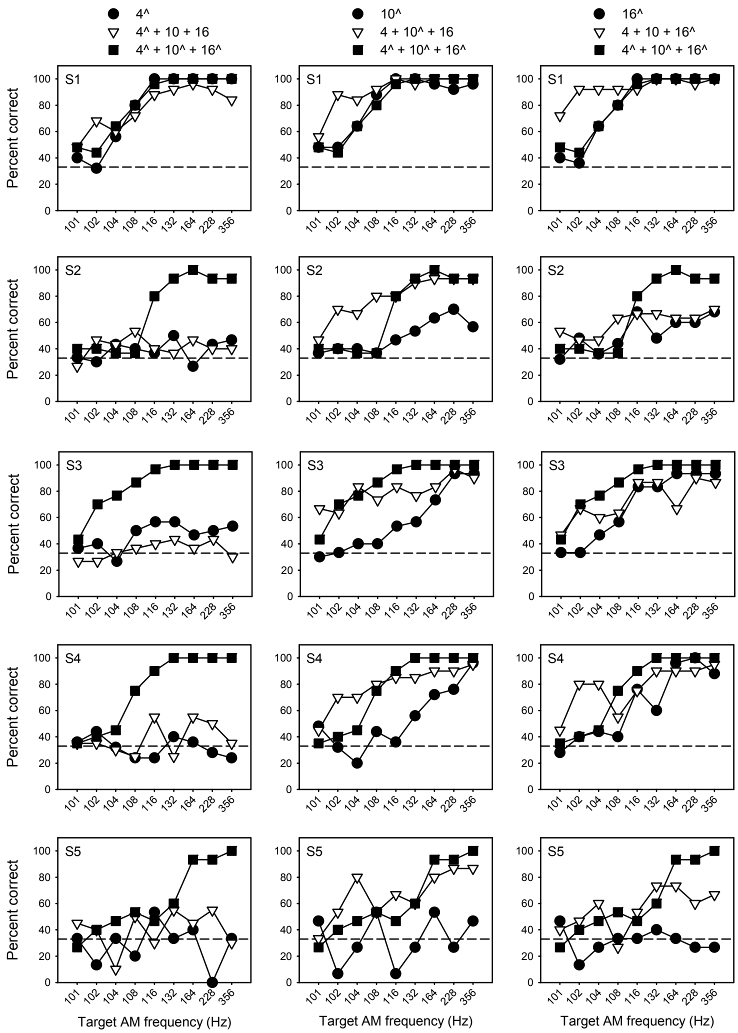

Fig. 5.4. Percent correct AMFD at each probe frequency with wide electrode spacing. Each row shows individual subject data. Each column shows data when the target AM was delivered to only the basal (left), middle (middle), or apical channel (right), or to all 3 channels. The white triangles show data from the present study, in which the target AM was delivered to 1 of 3 channels. The black symbols show data from Galvin et al. (2015), in which the target AM was delivered to a single channel (circles) or to all 3 channels (squares). The caret symbol (^) indicates the target AM channel(s). The dashed line shows chance level performance (33% correct).

Similarly, Figure 5.5 shows percent correct AMFD for the narrow spacing condition when the target AM delivered to a single channel, 1 of 3 channels or to all 3 channels; again, the black circle and square data are from Galvin et al. (2015). Different from the wide spacing condition, multi-channel performance was similar when the target AM was delivered to 1 of 3 channels, regardless of target AM channel. At low target AM frequencies (102-104 Hz), performance was markedly better when the target AM was delivered to 1 of 3 channels rather than to all 3 channels. At high target AM frequencies (> 132 Hz), performance tended to be better when the target AM was applied to all 3 channels.

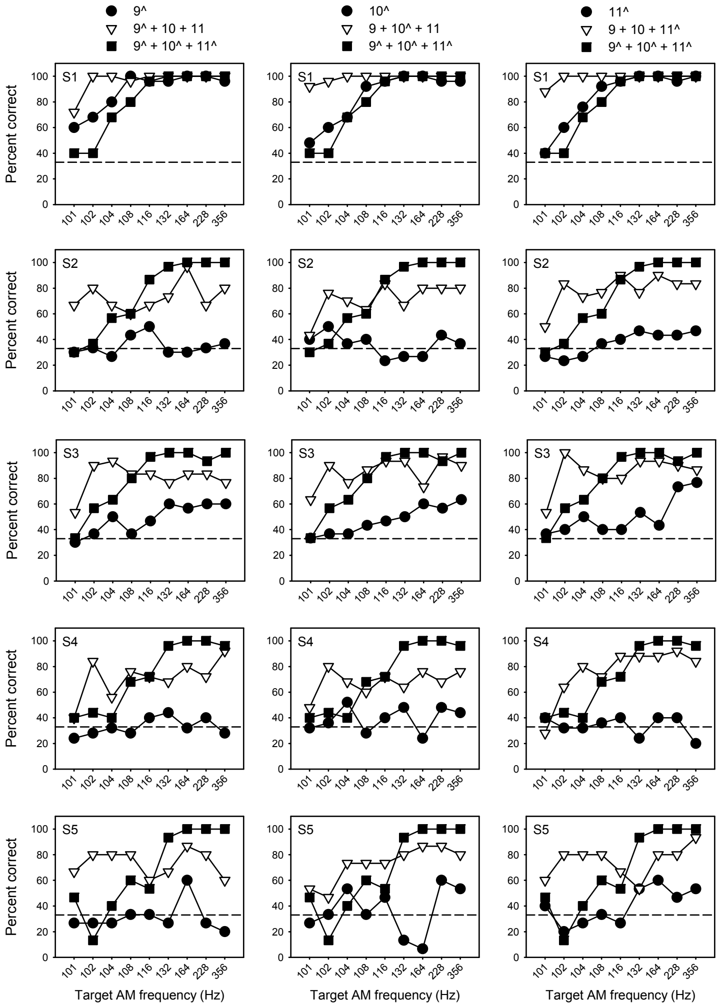

Fig. 5.5. Percent correct AMFD at each probe frequency with narrow electrode spacing. Each row shows individual subject data. Each column shows data when the target AM was delivered to only the basal (left), middle (middle), or apical channel (right), or to all 3 channels. The white triangles show data from the present study, in which the target AM was delivered to 1 of 3 channels. The black symbols show data from a Galvin et al. (2015), in which the target AM was delivered to a single channel (circles) or to all 3 channels (squares). The caret symbol (^) indicates the target AM channel(s). The lower solid line shows chance level performance (33% correct) and the upper dashed line shows a threshold of 79.4% correct.

A three-way repeated-measures analysis of variance (RM ANOVA) was performed on the data collected for the present study (i.e., the white triangle data in Figs. 5.4 and 5.5), with electrode spacing (wide or narrow), target AM channel (relatively basal, middle, or apical), and target AM frequency (101, 102, 104, 108, 116, 132, 164, 228, and 356 Hz) as factors. Results showed no significant main effects of electrode spacing [$F(1,7) = 3.345$, $p = 0.105$], target AM channel [$F(2,14) = 2.07$, $p = 0.161$], or target AM frequency [$F(8,56) = 1.01$, $p = 0.442$]. However, there were significant interactions between electrode spacing and target AM channel [$F(2,14) = 7.52$, $p = 0.006$], and between electrode spacing and target AM frequency [$F(8,56) = 2.53$, $p = 0.020$]. Because of these interactions, subsequent separate two-way RM ANOVAs were performed on the white triangle data from Figs. 5.4 and 5.5, with target AM channel and target AM frequency as factors. The results are shown in Table 5.3. For both the wide and narrow spacing, there were significant effects of target AM channel and target AM frequency ($p < 0.05$ in both cases). For the wide mode, AMFD was significantly poorer when the target AM was delivered to the basal channel ($p < 0.05$). In many cases, AMFD was significantly poorer with the 101 Hz target AM ($p < 0.05$).

Spacing	Factor	dF, res	F-ratio	*p*	Post-hoc (Bonferroni, p < 0.05)
	AM ch	2, 64	22.74	< 0.001	Apical, middle > basal
Wide	AM freq	8, 64	23.16	< 0.001	132, 164, 228, 356 > 101, 102, 104, 108
					102, 104, 108, 116 > 101
	AM ch X AM freq	16, 64	2.11	0.018	Middle: 104, 108, 116, 132, 164 > 101
					Apical: 132, 346 > 101, 108;
					116, 164, 228 > 101
	AM ch	2, 64	4.75	0.044	
Narrow	AM freq	8, 64	12.00	< 0.001	102, 104, 108, 116, 164, 228, 356 > 101
	AM ch X AM freq	16, 64	0.76	0.719	Basal: 102, 164 > 101
					Middle: 116, 164, 228, 356 > 101
					Apical: 102, 104, 108, 116, 132, 164,
					228, 356 > 101

Table 5.3. Results of the separate two-way RM ANOVAs performed on the white circle data (i.e., the target AM delivered to 1 of 3 channels) shown in Figs. 5.4 and 5.5. AM ch = target AM channel (relatively basal, middle, or apical); AM freq = target AM frequency (101, 102, 104, 108, 116, 132, 164, 228, or 356 Hz); dF, res = degrees of freedom, residual error.

In Figs. 5.4 and 5.5, AMFD across target AM frequency was often non-monotonic when the target AM was delivered to a single channel (black circles) or to 1 of 3 channels (white triangles). As such, it is difficult to estimate AMFD threshold. Fig. 5.6 shows mean percent correct AMFD (across all target AM frequencies) for the wide and narrow electrode spacing when the target AM was delivered to a single channel, 1 of 3 channels or to all 3 channels. Note that the data when the target AM was presented to a single channel or to all 3 channels are from Galvin et al. (2015) and are presented for comparison purposes. For the wide spacing, mean performance was generally poorer when the target AM was delivered to a single channel (hatched bars), and poorest when delivered to a single basal channel (hatched dark grey bars). Average performance was similar when the target AM was delivered to a single basal channel (hatched dark grey bars) or to the basal channel with the 100 Hz reference delivered to the apical and middle channel (solid dark grey bars). For the narrow spacing, mean percent correct was near chance-level when the target AM was delivered to any of the single channels (hatched bars), except for subject S1. Performance sharply improved when the target AM was delivered to 1 of 3 channels or to all 3 channels.

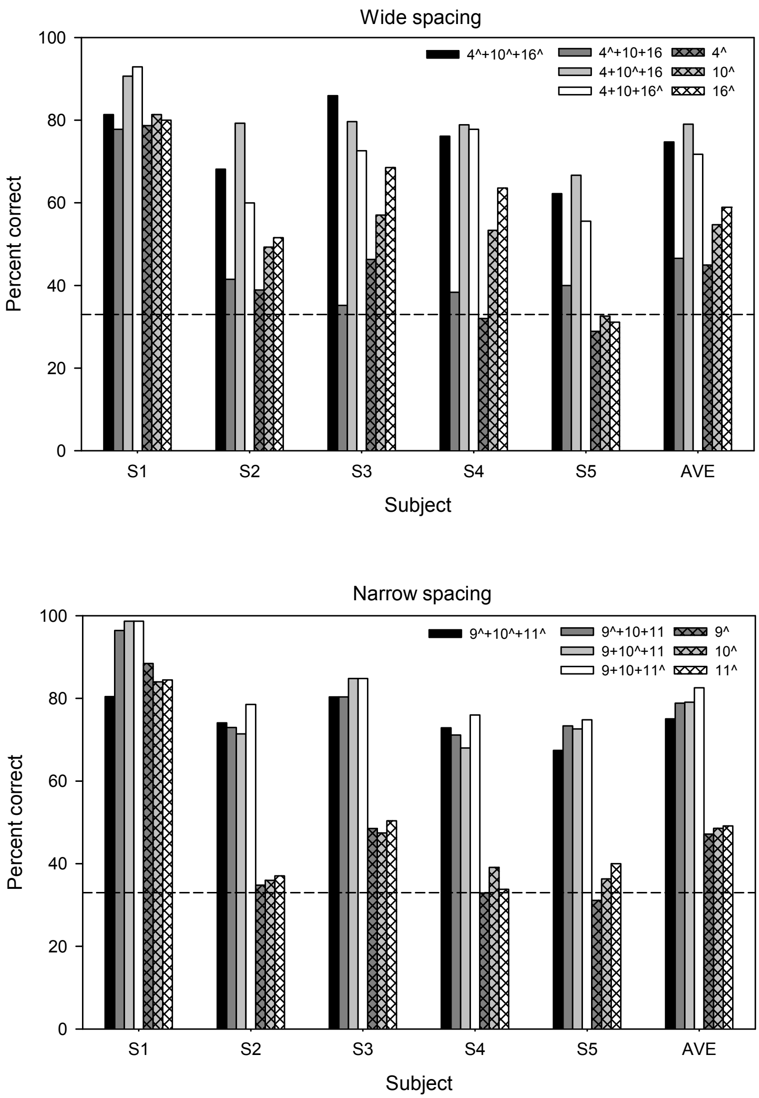

Fig. 5.6. Mean percent correct AMFD across all probe frequencies. Individual and average data is shown. The top and bottom panels show mean AMFD for the wide and narrow spacing, respectively. The black bars show performance when the target AM was delivered to all 3 channels and the hatched bars show performance when the target AM was delivered to a single channel; data are from a previous related study (Galvin et al., 2015). The caret symbol (^) indicates the target AM channel(s). The dashed line shows chance-level performance (33.3% correct).

A three-way RM ANOVA, with electrode spacing, target AM channel, and target AM condition (single channel, 1 of 3 channels, or all 3 channels) was performed on the mean AMFD data in Fig. 5.6. Again, note that the data when the target AM was delivered to a single channel or to all 3 channels are from a previous related study [21]. Results showed no significant effects for electrode spacing [$F(1,7) = 1.41$, $p = 0.253$], target AM channel [$F(2,14) = 1.49$, $p = 0.259$], or target AM condition [$F(2,14) = 1.99$, $p = 0.129$]. There were significant interactions between electrode spacing and target AM condition [$F(2,14) = 6.41$, $p = 0.011$], between electrode spacing and target AM channel [$F(2,14) = 6.34$, $p = 0.011$], between target AM condition and target AM channel [$F(4,28) = 6.84$, $p = 0.001$], and among all three factors [$F(4,28) = 6.31$, $p = 0.001$]. Because of these interactions, subsequent separate two-way RM ANOVAs were performed for the each target AM condition shown (single-channel, 1 of 3 channels, or all 3 channels) in Fig. 5.6, with electrode spacing and target AM channel as factors. The results are shown in Table 5.4. When the target AM was delivered to a single channel or to all three channels, there was no significant effect of electrode spacing. When the target AM was delivered to 1 of 3 channels, there were significant effects of electrode spacing and target AM channel ($p <0.05$), largely driven by the poorer mean AMFD when the target AM was delivered to the basal channel in the wide spacing.

As shown in Figs 5.4-5.6, AMFD improved greatly when two channels were added to a single target AM channel, whether with 2 reference or 2 target AM channels. At some target AM frequencies, the improvement in AMFD with multi-channel stimulation was sometimes greater when 2 reference AM channels were added rather than 2 target AM channels. Fig. 5.7 shows the mean difference in percent correct (across subjects) when the target AM was delivered to 1 of 3 channels or to all 3 channels, relative to when the target AM was delivered to a single channel (i.e., the mean difference between the white triangle data and the black circle and square data from Figs. 5.4 and 5.5). Again, note that the data when AM was delivered to a single channel or to all 3 channels are from Galvin et al. (2015). Values greater than zero indicate that performance was better with the multi-channel stimuli; values less than zero indicate that performance was better with single-channel stimuli. Note that performance with a single channel (black circle data in Figs. 5.4 and 5.5) was often quite poor and often near chance level, especially for the narrow spacing. In general, multi-channel performance was better than single-channel performance. One exception was the pattern of results for the multi-channel stimuli relative to single electrode 4 (top left panel of Fig. 5.7). When 2 target AM channels were added, performance sharply improved with AM frequency; there was little effect when 2 reference AM channels were added to single electrode 4. At low target AM frequencies (102-104 Hz), performance was often better when 2 reference AM channels rather than 2 target AM channels were added to the single channel. At higher target AM frequencies (>132 Hz), performance was often better when 2 target AM channels rather than 2 reference AM channels were added to the single channel. In general, there was a near monotonic improvement in performance with target AM frequency when the target AM was delivered to all 3 channels (black squares in Fig. 5.7). When the target AM was delivered to only 1 of 3 channels (white triangles in Fig. 5.7), performance also improved, but without a consistent relationship to target AM frequency.

Target AM condition	**Factor**	**dF, res**	**F-ratio**	***p***	**Post-hoc (Bonferroni, p < 0.05)**
	Spacing	1, 8	1.23	0.330	
Single channel	AM ch	2, 8	6.26	0.023	Apical > basal
	Spacing x AM ch	2, 8	3.65	0.075	Wide: apical > basal
	Spacing	1, 8	29.57	0.006	Narrow > Wide
1 of 3 channels	AM ch	2, 8	23.85	< 0.001	Apical, middle > basal
	Spacing x AM ch	2, 8	19.34	< 0.001	Wide: apical, middle > basal
					Basal: narrow > wide
All 3 channels	Spacing	1, 4	0.02	0.905	

Table 4. Results of RM ANOVAs performed on the mean AMFD data shown in Fig. 6. Separate analyses were performed for the different target AM channel conditions (i.e., the target AM delivered to a single channel, 1 of 3 channels, or all three channels). Spacing = wide or narrow, AM ch = target AM channel (relatively basal, middle, or apical); dF, res = degrees of freedom, residual error

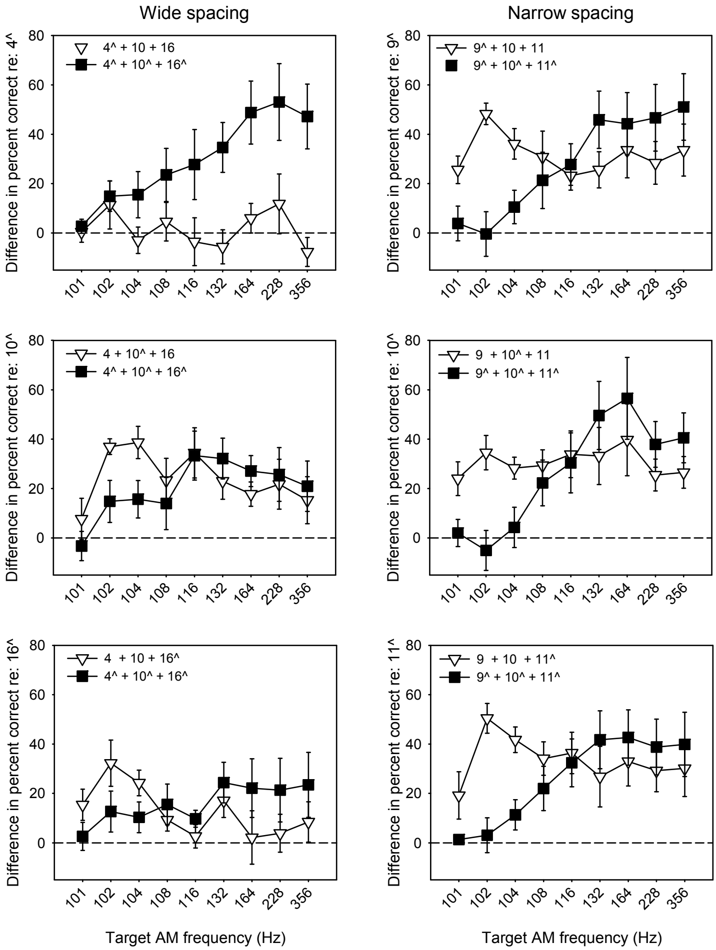

Fig. 5.7. Mean difference in percent correct (across subjects) when the target AM was delivered to 1 of 3 channels or to all 3 channels, relative to a single channel. The left and right columns show data for the wide and narrow spacing conditions, respectively. The top, middle, and bottom rows show data relative to the single basal, middle, or apical channel, respectively. The caret symbol (^) indicates the target AM channel(s). The black squares show the mean difference when 2 target AM channels were added to the single target AM channels; the white triangles show performance when 2 reference AM channels were added to the single target AM channels. The error bars show the standard error. Data for the single-channel reference and when the target AM was delivered to all 3 channels (black squares) are from Galvin et al. (2015).

A four-way RM ANOVA was performed on the difference data shown in Fig. 5.7, with electrode spacing, additional channel type (2 reference AM or 2 target AM), target AM channel, and target AM frequency as factors. The results are shown in Table 5.5. While there were no significant main effects, there were significant interactions between electrode spacing and additional channel type, additional channel type and target AM channel, additional channel type and target AM frequency, and among electrode spacing, additional channel type, and target AM channel ($p < 0.05$ in all cases).

Factor	dF, res	F-ratio	*p*
Spacing	1, 8	2.77	0.134
Add ch	1, 8	1.12	0.321
AM ch	2, 16	0.96	0.403
AM freq	8, 64	1.01	0.442
Spacing X Add ch	1, 8	6.77	0.032
Spacing X AM ch	2, 16	2.60	0.105
Spacing X AM freq	8, 64	1.34	0.242
Add ch X AM ch	2, 16	7.26	0.006
Add ch X AM freq	8, 64	7.67	< 0.001
AM ch X AM freq	16, 128	1.61	0.074
Spacing X Add ch X AM ch	2, 16	6.83	0.007
Spacing X Add ch X AM freq	8, 64	1.81	0.091
Spacing X Am ch X AM freq	16, 128	1.13	0.334
Add ch X AM ch X AM freq	16, 128	1.62	0.073
Spacing X Add ch X AM ch x AM freq	16, 128	1.23	0.265

Table 5. Results from a four-way RM ANVOA performed on the difference data shown in Fig. 5.7. Spacing = wide or narrow, Add ch = type of channels added to the single target AM channel (2 reference AM or 2 target AM); AM ch = target AM channel (relatively basal, middle, or apical); AM freq = target AM frequency (101, 102, 104, 108, 116, 132, 164, 228, or 356 Hz); dF, res = degrees of freedom, residual error.

Because of the interaction shown in the previous four-way RM ANOVA, separate two-way RM ANOVAs were performed on the difference data shown in each panel of Fig. 5.7, with added channel type (2 reference or 2 target AM channels) and target AM frequency as factors. The results are shown in Table 5.6. Adding 2 target AM channels was significantly better than adding 2 reference AM channels only relative to single target AM channel 4 ($p < 0.05$; top left panel of Fig. 5.7). For the narrow spacing, the difference in AMFD was significantly better when adding 2 reference AM channels than when adding 2 target AM channels, only for 102 and 104 Hz ($p < 0.05$). Relative to single AM channel 10, the difference in AMFD was significantly greater when adding 2 reference AM channels than when adding 2 target AM channels, for 102 Hz and only for the wide spacing ($p < 0.05$), Although there appeared to be a greater difference in AMFD when adding 2 target AM channels for frequencies > 132 Hz, there was no significant effect, except relative to single channel 4 (top left panel of Fig. 5.7).

Another series of separate two-way RM ANOVAs were performed on the data shown in Fig. 5.7, this time with electrode spacing (wide or narrow) and target AM frequency as factors. Data was analyzed separately for conditions when 2 reference AM channels or 2 target AM channels were added to the single AM channel. Data was also analyzed separately for the basal, middle, and apical single-channel references. The results are shown in Table 5.7. There was a significant effect for electrode spacing only when 2 reference AM channels were added ($p < 0.05$), with a greater difference for the narrow than for the wide spacing. For the basal and middle single-channel references, there was no significant effect of target AM frequency when 2 reference AM channels were added ($p > 0.05$). When 2 target AM channels were added, there were significant differences between relatively high and low target AM frequencies ($p < 0.05$), especially for the narrow combination, but no significant differences between the wide and narrow spacing ($p > 0.05$). It should be noted that for the analyses presented in Tables 5.6 and 5.7, power was sometimes quite low due to the small number of subjects.

Spacing	Single-channel ref	Factor	dF, res	F-ratio	*p*	Post-hoc (Bonferroni, p < 0.05)
Wide	4	Added ch	1, 32	12.8	0.023	2 target AM > 2 ref AM
	4	AM freq	8, 32	1.7	0.135	
	4	Added ch x AM freq	8, 32	6.3	< 0.001	2 target AM: 164, 228 > 101 256: 2 target AM > 2 ref AM
	10	Added ch	1, 32	2.0	0.234	
	10	AM freq	8, 32	1.7	0.148	
	10	Added ch X AM freq	8, 32	7.1	< 0.001	102: 2 ref AM > 2 target AM
	16	Added ch	1, 32	0.5	0.531	
	16	AM freq	8, 32	0.6	0.790	
	16	Added ch X AM fred	8, 32	5.4	< 0.001	
Narrow	9	Added ch	1, 32	1.3	0.318	
	9	AM freq	8, 32	2.9	0.014	356 > 101
	9	Added ch x AM freq	8, 32	25.8	< 0.001	2 target AM: 132, 164, 228, 356 > 101, 102, 104; 102, 104: 2 ref AM > 2 target AM
	10	Added ch	1, 32	1.0	0.372	
	10	AM freq	8, 32	2.4	0.035	
	10	Added ch X AM freq	8, 32	27.3	< 0.001	2 target AM: 132, 164 > 101, 102, 104; 356 > 102; 102: 2 ref AM > 2 target AM
	11	Added ch	1, 32	7.4	0.053	
	11	AM freq	8, 32	1.7	0.133	
	11	Added ch X AM freq	8, 32	15.0	< 0.001	2 target AM: 132, 164 > 101, 102; 356 > 101

Table 5.6. Results of two-way RM ANOVAs performed on data shown in Fig. 5.7, with the type of added channel and target AM frequency as factors. Single-channel ref = single channel reference used to calculate performance difference between single- and multi-channel AMFD scores; Added ch = type of channel added to the single AM channel (2 target AM or 2 reference AM channels); AM freq = target AM frequency (101, 102, 104, 108, 116, 132, 164, 228, or 356 Hz); dF, res = degrees of freedom, residual error.

Added channels	Single-channel ref	Factor	dF, res	F-ratio	*p*	Post-hoc (Bonferroni, p < 0.05)
	A	Spacing	1, 32	27.8	0.006	Narrow > Wide
	A	AM freq	8, 32	1.4	0.237	
	A	Spacing X AM freq	8, 32	1.0	0.437	
2 reference AM	B	Spacing	1, 32	7,8	0.049	Narrow > Wide
	B	AM freq	8, 32	1.4	0.249	
	B	Spacing X AM freq	8, 32	0.9	0.949	
	C	Spacing	1, 32	10,9	0.030	Narrow > Wide
	C	AM freq	8, 32	2.5	0.029	
	C	Spacing X AM freq	8, 32	0.9	0.548	
2 target AM	A	Spacing	1, 32	0.3	0.636	
	A	AM freq	8, 32	8.1	< 0.001	164, 228, 356 > 101, 102, 104; 132 Hz > 101,102 Hz
	A	Spacing X AM freq	8, 32	0.6	0.619	
	B	Spacing	1, 32	2.7	0.179	
	B	AM freq	8, 32	6.8	< 0.001	132, 164 > 101, 102, 104; 116, 228, 356 > 101
	B	Spacing X AM freq	8, 32	2.5	0.032	Narrow: 132, 164 > 101, 102, 104; 116, 228, 356 > 101
	C	Spacing	1, 32	2.5	0.188	
	C	AM freq	8, 32	4.1	0.002	132, 164, 256 > 101
	C	Spacing X AM freq	8, 32	15.0	< 0.001	Narrow: 132 > 101; 164 > 101, 102

Table 5.7. Results of two-way RM ANOVAs performed on data shown in Fig. 5.7, with electrode spacing and target AM frequency as factors. Single channel ref = single channel used to calculate performance differences between single- and multi-channel AMFD scores; Spacing = wide or narrow; AM freq = target AM frequency (101, 102, 104, 108, 116, 132, 164, 228, or 356 Hz); dF, res = degrees of freedom, residual error.

Discussion

The present data showed substantial envelope interaction that was not consistently related to difference in AM frequency between the target and reference AM channels. The envelope interactions were greater when channels were narrowly spaced than when widely spaced (i.e., when there was greater channel interaction in terms of spread of excitation). However, there was substantial interference even among widely spaced channels, suggesting a central component to perception of multiple temporal envelopes. Below, we discuss the present results in greater detail.

Channel and envelope interactions

When the target AM was delivered to only 1 of 3 channels, there was not a consistent relationship between AMFD and the differences in AM frequency across multiple channels. However, CI subjects were very sensitive to the presence of envelope interactions at all target AM frequencies. This result is somewhat consistent with previous studies that showed sometimes highly elevated AMFD thresholds when a single masker AM channel was combined with a single target AM channel (Chatterjee and Ozerbut, 2009; Kreft et al., 2013).

CI subjects were extremely sensitive to very small AM frequency differences (2-4 Hz) between component channels, especially when channels were narrowly spaced. For the 20 ms segments shown in the top row of Fig. 5.2, the overall temporal envelope appears to be quite similar whether the 102 Hz target AM was delivered to 1 of 3 channels (middle panel) or to all 3 channels. However, given the 300-ms stimulus duration, the 100-Hz reference and 102-Hz target AM would have been out of phase, which may have provided a strong perceptual cue. The bottom row of Fig. 5.2 shows strong differences between the target and reference channels when the 132 Hz target AM was delivered to 1 of 3 channels. Yet performance was quite similar between the 102 and 132 Hz target AM frequencies when the target AM was delivered to 1 of 3 channels (see Figs. 5.4 and

5.5). Because the present multi-channel stimuli were interleaved in time, and because the interference was greater for narrowly spaced channels, CI subjects may have attended to interactions between envelopes at the neural level (rather than the envelopes themselves interacting). Given the relatively high stimulation rate (2000 pps/channel), neurons responding to stimulation from one channel might not have fully recovered before receiving stimulation from the second and third channels. As such, the temporal envelopes from each channel may have been combined in the overlapping neural region. When the difference in AM rate across channels was small, this may have produced some irregularity in the probe stimuli. Such percepts associated with low AM rate differences were also observed in previous AMFD studies with interferers (Chatterjee and Ozerbut 2009; Chatterjee and Kulkarni, 2015). In this study, this percept persisted for larger AM frequency differences. As such, the present results do not reflect CI subjects' ability to discriminate target AM frequency in the presence of competing AM channels, but rather their sensitivity to envelope interactions that did not strongly depend on AM frequency differences. This sensitivity may have been somewhat elevated for low AM rate differences between the target and reference AM channels.

The present data showed greater interference among narrowly spaced channels than widely spaced channels, similar to previous studies (Chatterjee and Ozerbut, 2009; Kreft et al., 2013). Still, there was significant interference among widely spaced channels. The spread of excitation might be expected to be reduced given the low summation-adjusted stimulation levels on each channel. The present data suggests that interactions most likely occurred where these channels overlapped, which would have been more pronounced with narrow spacing. In Galvin et al. (2015), there was no significant effect of electrode spacing when the target AM was applied to all 3 channels. When the target AM was applied to only 1 of 3 channels, there was a significant effect of electrode spacing, suggesting that the effect of channel interaction may depend on the type of envelope information delivered to each channel. When the envelope information was the same on all channels, the degree of channel interaction had little effect. When the envelope information was different across channels, the degree of channel interaction mattered greatly. Thus the peripheral pattern may matter more when processing competing rather than coherent envelopes.

The present data also suggest that using adaptive procedures to measure AMFD with interferers may not produce meaningful threshold data. In this study, there was no monotonic relationship between the target AM frequency and performance when the target AM was delivered to 1 of 3 channels. As such, AMFD thresholds derived from an adaptive pitch ranking procedure, as used in some previous studies (Chatterjee and Ozerbut, 2009; Kreft et al., 2013) may not accurately reflect perception of frequency differences between component AM channels. Also, adaptive procedures in an AMFD task may not test very low frequency differences between the reference and probe AM rates, as thresholds often are 10% or more of the reference rate (Kreft et al., 2015). A non-adaptive procedure as used in the present study allows the psychometric function to be directly measured. As such, any non-monotonicities in the psychometric function may be observed. In the present results, the non-adaptive procedure revealed non-monotonic pattern of results when the target AM was delivered to 1 of 3 channels. Given the present pattern of results, it is unclear whether the sometimes greatly elevated thresholds reported in previous AMFD studies (Chatterjee and Ozerbut, 2009; Kreft et al., 2013) fully reflect CI users' ability to perceive target AM rates in the presence of interferers. Even lower thresholds reported for some masked conditions may not reflect the nature of the envelope interactions (Chatterjee and Ozerbut, 2009; Kreft et al., 2013), as the present data suggest a dip in the masked threshold function when the target AM frequency difference was between 8 and 32 Hz. The present data suggest substantial interference even when the difference between the target and reference channel AM rates was quite small, and that this interference persisted even when the difference in AM rates was nearly 2 octaves, especially for the narrowly spaced electrodes.

In this study, only monopolar stimulation was used. Monopolar stimulation is associated with broader absolute current spread compared to more focused stimulation modes such as bipolar or tripolar, in which the current is restricted within the cochlea (e,g., Bierer and Middlebrooks, 2002; Snyder et al., 2004). However, bipolar and tripolar also require much greater current to achieve comparable loudness to monopolar stimulation (e.g., Litvak et al., 2007; Landsberger and Srinivassan, 2009). Because higher current levels are associated with greater spread of excitation (Chatterjee and Shannon, 1998), the spread of excitation may be comparable across monopolar and focused stimulation modes when measured at the same loudness. While some studies have shown some reduced spread of excitation for some CI users with focused stimulation (e.g., Landsberger et al., 2012), these small advantages were observed for single-channel measures. When multi-channel stimulation is considered, these small advantages most likely would not persist. Thus, in this study, it is unlikely that monopolar stimulation was a limiting factor in the present pattern of results.

Multi-channel loudness summation and temporal envelope processing

In Galvin et al. (2015), single-channel AMFD with summation-adjusted current levels was quite poor (see circle data in Figs 5.4 and 5.5). In that study, increasing the current of a single channel or adding channels with coherent AM greatly improved performance. In this study, adding channels with different AM to the target AM was easily perceived, though none of the single channels could convey temporal envelope information when presented in isolation. In both cases, there was greater temporal envelope sensitivity with multiple channels. In Galvin et al. (2015), the multi-channel advantage was explained by the increased loudness rather than by multiple representations of the temporal envelope. However, the present data suggest that envelope information may have been combined across channels. Loudness summation may still play a role in multi-channel envelope processing, as envelopes may not be effectively combined across channels until achieving some criterion loudness (e.g., comfortably loud). Thus, the present results also support previous work (Geurts and Wouters, 2011; Galvin et al., 2015) in which the multi-channel advantage in AMFD was explained by the multiple representations of envelope information.

One exception to the present pattern of results is subject S1 (top row in Figs. 5.4 and 5.5). Subject S1 experienced the least amount of multi-channel loudness summation. Consequently, single-channel and multi-channel AMFD were very similar (Galvin et al., 2015). Note that S1 was also the most sensitive to envelope interactions, exhibiting the highest scores of all subjects at all target AM frequencies. Subject S1 did not have the largest DRs or lowest T levels (see Table 5.2), so absolute current levels do not explain the greater sensitivity to envelope interactions. Subject S1 exhibited similar effects of electrode spacing as the other subjects, so it is unlikely that there was markedly different channel interaction.

Contributions of individual channels to multi-channel envelope processing

In Galvin et al. (2015), it was difficult to observe across-site differences in single-channel AMFD. At the summation-adjusted levels, performance was too poor and at comfortably loud levels, performance was too good. As such, contributions of individual channels to the multi-channel percept could not be observed when coherent AM was delivered to all 3 channels. One motivation for the present study was to vary the stimulation site of the target AM channel when the target was delivered to 1 of 3 channels. Across-site differences in this manipulation might reveal channels that strongly or weakly interacted with the others. Channels with better temporal processing might be more resistant to the interferers. Alternatively, channels with poorer temporal processing might interact weakly with channels with better temporal processing.

In the wide spacing, there was little interaction when the target AM was delivered to EL 4 and the reference AM was delivered to EL 10 and EL 16. Indeed, performance was quite similar when the target AM was delivered to EL 4 (circle and triangle data in left column of Fig. 5.4), whether or not the reference AM was delivered to EL 10 and EL 16. Given that AMFD was generally poorest when the target AM was delivered to single EL 4, it is unlikely that good temporal processing made EL 4 more resistant to the interferers. Interestingly, when the target AM was delivered to EL 10 and EL 16 (square data in left column of Fig. 5.4), performance sharply improved. Taken together, these patterns of results suggest that performance was largely driven by EL 10 and EL 16, whether or not coherent AM was delivered to the additional channels. It seems likely that for these subjects and stimuli, temporal processing was poor for EL 4 and thus contributed weakly to multi-channel envelope processing. Such an observation is consistent with previous studies that have suggested better temporal processing in the apical region of the cochlea (Middlebrooks and Snyder, 2010; Macherey et al., 2011), although no strong or consistent advantage has been shown for apical electrodes (Baumann and Nobbe, 2004; Carlyon et al. 2010). This result is not consistent with previous studies that have shown no significant effect of interferer location on AMFD (Chatterjee and Ozerbut, 2009; Kreft et al., 2013). Note that in these studies, only 2 channels were stimulated (1 target and 1 interferer), and the target AM was typically delivered to an electrode in the middle of the array. In this study, the stimulation site of the target AM channel was varied across all 3 channels, which may have revealed the different pattern of results.

In the wide spacing, when the target AM was delivered to EL 10 or 16 (middle and bottom left panels of Fig. 5.7, respectively), there was substantial interaction, especially for low target AM frequencies. Interestingly, the largest interaction was observed when the target AM was delivered to EL 10. It is unclear whether this indicates better temporal processing on EL 10 (which might give rise to stronger interaction) or interactions with the spread of excitation from both EL 4 and EL 16. When the target AM was delivered to EL 4 or 16, either would have primarily interacted with EL 10, as the spread of excitation from the most spatially remote electrode would have produced much less interference.

There was a significant interaction between electrode spacing and target AM channel for mean AMFD when the target AM was delivered to only 1 of 3 channels (see Table 5.4). With the wide spacing, mean AMFD was significantly better when the target AM was delivered to the apical or middle channels, rather than the basal channel. With the narrow spacing, there was no significant difference among target AM channels, most likely because of the strong overlap in the spread of excitation among ELs 9, 10, and 11.

Limitations to the present study

A 3AFC task was used in this study to measure AMFD, similar to many previous studies (Chatterjee and Peng, 2008; Chatterjee and Ozerbut, 2011; Deroche et al., 2013, 2014; Galvin et al., 2015), rather than a 2AFC pitch ranking task (Geurts and Wouters, 2001; Kreft et al., 2010, 2013; Green et al., 2013). CI subjects were very sensitive to the channel interactions in this study. As discussed above, a 2AFC adaptive procedure may not be appropriate given the present non-monotonic functions when AMFD was measured with interferers. One alternative would be to measure pitch ranking with interferers using a 2AFC non-adaptive procedure.

The AM depth used in this study was much deeper than typically used in previous AMFD or MDI studies, which is typically some value above MDT (Chatterjee and Ozerbut, 2011). This large AM depth may have contributed to the present pattern of results. It is unclear whether the present pattern of results would hold with smaller AM depths. Also, the summation-adjusted current levels used in this study were quite low, providing very poor single-channel AMFD. Most previous studies have measured AMFD or MDI at higher loudness levels (Geurts and Wouters, 2001; Chatterjee and Ozerbut, 2009; Kreft et al., 2013), which provides good AMFD even with a smaller AM depth than used in this study. However, the present summation-adjusted current levels are likely to be more comparable to those used in clinical processors. AMFD measured at these summation-adjusted levels may be more representative of the temporal processing limits within each channel. With multi-channel stimulation, AMFD greatly improves due to increased loudness and/or combined coherent AM information. Unfortunately, channels with different envelope information interact as well, resulting in poor perception of the target AM.

Implications for CI users

The present results demonstrate the importance of reducing channel interaction in CIs. Envelope interference was reduced in the present wide spacing, relative to the narrow spacing. Results from the previous studies suggest that CI users may benefit from redundant envelope cues presented on multiple channels. As such, similar envelope cues could be delivered to adjacent channels while dissimilar envelope cues could be delivered to spatially remote channels; in such an approach, adequate and accurate representation of the spectral envelope should still be maintained. High rates may further increase channel interaction (Middlebrooks, 2004). As such, lower stimulation rates may improve channel independence and reduce envelope interference. Finally, given the effects of loudness summation on multi-channel envelope processing, it might be advisable to stimulate fewer channels per stimulation cycle. Fewer channels in each cycle may require higher current levels to maintain adequate loudness. The higher current levels may in turn improve temporal processing for each channel and subsequently improve multi-channel envelope perception.

Conclusions

In this study, multi-channel AMFD was measured using stimuli in which the target AM was delivered to 1 channel and the reference AM was delivered to 2 channels. The spacing between electrodes was varied to be wide or narrow, thereby testing the effect of relative channel interaction on multi-channel AMFD. The stimulation site of the target AM channel was varied to test single-channel contributions to the multi-channel AMFD. The present data were compared to data from a previous study in which the target AM was delivered to a single channel or to all 3 channels; in all cases, AMFD was measured using reduced current levels on each channel to accommodate multi-channel loudness summation. Key findings include:

1. CI subjects were very sensitive to multi-channel envelope interference, especially when electrodes were narrowly spaced.

2. When only the target AM was delivered to 1 of 3 channels, there was not a consistent relationship with target AM frequency. The non-monotonic functions suggest that a non-adaptive procedure, as used in this study may be more appropriate than adaptive pitch ranking tasks used in previous studies that measured AMFD with interfering envelopes.

3. When electrodes were widely spaced, there was little interaction among channels when the target AM was delivered to the most basal channel, possibly due to poorer temporal processing in the basal electrode. The most envelope interaction was observed when the target AM was delivered to the middle electrode and the reference AM was delivered to the apical and basal electrodes, which may have maximized interactions at the edges of the spread of excitation.

4. Data from Galvin et al. (2015) showed that single-channel AMFD was very poor at summation-adjusted current levels. When multiple channels were added that contained coherent AM, AMFD improved greatly. When multiple channels were added that contained different AM from the target, CI subjects were very sensitive to envelope interactions. Thus, channels that were not capable of transmitting envelope cues could be combined to deliver envelope information that was easily perceived. This suggests that listeners combined envelope information across channels, in addition to benefitting from the increased loudness associated with multi-channel summation.

Acknowledgements

We thank all of the CI subjects for their tireless participation. We also thank Joseph Crew for helpful comments. John Galvin, Sandy Oba, and Qian-Jie Fu were supported by NIH grant DC004993. Monita Chatterjee was supported by NIH grant R01 DC014233.Deniz Baskent was supported by VIDI grant 016.096.397 from the Netherlands Organization for Scientific Research (NWO) and the Netherlands Organization for Health Research and Development (ZonMw), and a Rosalind Franklin Fellowship from University of Groningen, University Medical Center Groningen.

Chapter 6

General discussion

The four studies presented here shed light on different aspects of CI users' perception of AM delivered to single and/or multiple channels. AMD and AMFD reflect somewhat different aspects of AM perception, yet share some qualities. Both are sensitive to current level, but AMFD appears to also be sensitive to loudness. As such, the data in the AMD and AMFD studies showed somewhat different effects of multi-channel loudness summation. Multi-channel AMD worsened as the current levels were reduced to accommodate summation; multi-channel AMFD was unaffected. This difference is results may also reflect differences in the perceptual task. In AMD, listeners were asked to detect the presence of modulation, while in AMFD, listeners were asked to discriminate between AM frequencies for temporal envelopes that were well above MDT. While both measures have correlated with speech perception (Cazals et al., 1994; Fu, 2004; Chatterjee and Peng, 2008: Luo et al., 2008; Deroche et al., 2012, 2014), they represent different, but related aspects of envelope perception in speech (e.g., envelope saliency versus sensitivity to changes in envelope rate for salient envelopes).

One major factor in these studies was loudness summation. In our previous single-channel AMD studies (Galvin and Fu, 2005; 2009), single-channel MDTs were poorer for high-rate carriers than for equally loud low-rate carriers, due to the current level reduction needed to accommodate multi-pulse integration. At a fixed current level, single-channel MDTs were better with the high-rate carriers, presumably due to the increased loudness. Similarly, in the present second study, MDTs were poorer for multi-channel stimuli than for equally loud single-channel stimuli, again due to the current level reductions needed to accommodate multi-channel loudness summation. With no adjustment for summation, performance was better with multiple than with single-channels. In the third and fourth studies, at low presentation levels, single-channel AMFD was very poor; when multiple channels were added, AMFD sharply improved. However, when single- and multi-channel stimuli were presented at the same loudness, AMFD was quite similar, suggesting that overall loudness contributed strongly to the performance, rather than the distribution of envelope information. It is worth noting that subjects differed in terms of loudness summation. Some subjects exhibited very little summation, and envelope perception was quite similar between single and multi-channel stimulation. Other subjects exhibited much greater summation, and were more greatly affected by the current level reductions.

While previous studies have shown significant across-site differences in single-channel envelope perception (e.g., Pfingst et al., 2007), it was difficult to observe the contributions of single channels to the multi-channel percept. In Chapter 3, multi-channel AMD did not appear to be limited by the channel with the best or worst temporal envelope processing. Rather, CI subjects appeared to combine information across multiple channels. In Chapter 4, it was difficult to observe across-site differences in single channel AMFD at the relatively low and high presentation levels, due to floor and ceiling performance effects. In Chapter 5, there was some evidence of greater multi-channel envelope interaction when the target AM was delivered to the apical or middle channels rather than to the basal channel for widely spaced electrodes. This finding gives some support to previous studies that suggest better temporal processing in the apex (e.g., Middlebrooks and Snyder, 2010).

The distribution of envelope information had no significant effect on multi-channel AMFD when the target AM was delivered to all 3 channels (Chapter 4). When the target AM was delivered to only 1 of 3 channels, there was greater envelope interaction for narrowly spaced channels than for widely spaced channels (Chapter 5). For these experiments, subjects were asked to discriminate between target AM frequencies. In Chapter 4, the target AM was delivered to all 3 channels. As such, discrimination was likely based on rate differences between the reference and probe stimuli. In Chapter 5, the reference AM was delivered to all 3 channels for the reference stimulus and the target AM was delivered to 1 of 3 channels for the probe stimulus. As such, listeners may have been very sensitive to the envelope interactions between the reference and target AM in the probe stimulus. This interaction was very easy to detect when the difference between the reference and target AM was small (2-4 Hz), and when channels were narrowly spaced.

Data from these studies highlight potential difficulties when extrapolating multi-channel performance from single-channel measures. Data from Chapter 3 suggests that for equally loud stimuli, single-channel measures may greatly overestimate multi-channel AMD, due to the reduced current levels needed to accommodate multi-channel loudness summation. At summation-adjusted current levels, single-channel AMFD was quite poor (Chapter 4); adding coherent AM to other channels sharply improved performance. Single-channel data from Chapter 4 showed little across-site variability in AMFD, due to floor and ceiling effects associated with the current levels tested. However, multi-channel envelope interactions (Chapter 5) showed some evidence that performance was affected by the channel that received the target AM. Overall, the present data suggest that multi-channel loudness summation and channel interaction should be important considerations when evaluating perceptual limits of temporal envelope processing. When current levels are reduced to accommodate multi-channel loudness summation, detection of temporal envelopes may worsen but discrimination of envelope frequency may not. When coherent envelope information is distributed to multiple channels, electrode spacing may not matter; when different envelope information is distributed to multiple channels, envelope interference may be increased for narrowly spaced channels, especially when the differences in temporal envelopes are small (i.e., "neural beating effects"). Such findings may not be readily apparent when measuring single-channel AMD or AMFD.

The present studies required a research interface to directly control stimulation parameters such as current level, stimulation rate, electrode(s) to be stimulated, AM depth and frequency, etc. It was vital to bypass CI users' clinical processors, where the stimulation pattern would have been distorted relative to the acoustic input signal. These studies also required careful stimulus control to balance loudness across component electrodes and across single- and multi-channel stimuli. The present studies also gave rise to improved methods for measuring perception of temporal envelopes. In the first two studies, AMD was adaptively measured using a novel method to control for potential loudness cues associated with peak AM amplitude. This control presumably provided better estimates of envelope detection, rather than peak amplitude sensitivity in the AM stimulus. In the last two studies AMFD was measured using a non-adaptive 3AFC task, which allowed for psychometric functions to be fit to the data, which in turn revealed a non-monotonic function when different temporal envelopes were applied to the multi-channel probe stimulus. This non-monotonic function suggests that data from previous AMFD studies with masking electrodes (Chatterjee and Ozerbut, 2009; Kreft et al., 2013) may not have reflected meaningful detection thresholds. However, in this and previous studies, it is clear that CI users have great difficulty segregating envelope information within multi-channel, temporally complex stimuli. One other advantage for the non-adaptive procedure is that listeners are not performing the task at or near threshold for most of the test, as is the case in adaptive procedures. This results in a less stressful test run, with relatively easy and difficulty comparisons randomly mixed together. The non-adaptive procedure also required a similar amount of time to complete as the adaptive procedure, provided that the range of probe values was optimized. Where possible, non-adaptive procedures may be preferable to adaptive procedures for many psychophysical measures with CI users, especially when interactions between multiple channels may be complex.

Given the limited functional spectral resolution experienced by CI users, temporal envelope cues are important for speech recognition. The present studies suggest that multi-channel envelope perception may be affected by a number of factors, including loudness summation, channel interaction, and the type of envelope information presented on each channel. In terms of clinical implications, these factors should be considered for clinical fitting of speech processors. Fewer channels in each stimulation frame may reduce multi-channel loudness summation, which would allow for higher current levels on each channel, which in turn may improve modulation detection (as shown in the second study). To reduce channel interaction, it would be best to maximize the spatial distance among channels stimulated in each frame. Competing envelope interaction is reduced with widely spaced channels, as shown in the fourth study. Even with widely spaced channels, the stimulation rate should be sufficiently low (e.g., 500 pps/channel) to reduce temporally induced channel interaction. Lower stimulation rates have been shown to provide better single-channel AMD (Galvin and Fu, 2005, 2009; Pfingst et al., 2007). Because of reduced multi-pulse integration, lower rates would also require higher current levels on each channel which, as noted above, would improve multi-channel AMD. Selecting channels with good temporal modulation has been shown to improve speech performance (Zhou and Pfingst, 2012; Garadat et al.. 2013). Alternatively, compressing the electric DR for electrodes with poor temporal processing has also been shown to improve speech performance (Zhou and Pfingst, 2013). This approach may be help to preserve spectral envelope information while maintaining good temporal processing when selecting a subset of electrodes to stimulate within each frame. The results from the third study suggest that AMFD can be improved when coherent temporal envelope information is delivered to multiple channels. Within a speech processor strategy, it might be beneficial to deliver coherent envelope cues to multiple channels, especially if the peak amplitude of the envelope is relatively low. To the extent that temporal envelope cues can be perceived within dynamic spectral envelope cues, CI fitting should be optimized to limit multi-channel loudness summation and channel interaction.

Chapter 7: References

Arora K, Vandali A, Dowell R, Dawson P (2011) Effects of stimulation rate on modulation detection and speech recognition by cochlear implant users. Int J Audiol. 2011;50:123-32.

Baumann U, Nobbe A. Pulse rate discrimination with deeply inserted electrode arrays. Hear Res. 2004;196:49–57.

Bierer JA, Middlebrooks JC. Auditory cortical images of cochlear-implant stimuli: dependence on electrode configuration. J Neurophysiol. 2002;87:478–92.

Busby PA, Tong Y, Clark GM. The perception of temporal modulations by cochlear implant patients. J Acoust Soc Am. 1993;94:124-31.

Carlyon RP, Lynch C, Deeks JM. Effect of stimulus level and place of stimulation on temporal pitch perception by cochlear implant users. J Acoust Soc Am. 2010;127: 2997–3008 .

Cazals Y, Pelizzone M, Saudan O, Boex C. Low-pass filtering in amplitude modulation detection associated with vowel and consonant identification in subjects with cochlear implants. J Acoust Soc Am. 1994;96:2048-54.

Chatterjee M. Modulation masking in cochlear implant listeners: envelope versus tonotopic components. J Acoust Soc Am. 2003;113:2042-53.

Chatterjee M, Kulkarni A. AM rate discrimination by cochlear implant listeners in on- and off-channel, modulated masking. *2015 Midwinter Meeting of the Association for Research in Otolaryngology,* Baltimore, MD. 2015;152

Chatterjee M, Robert ME. Noise enhances modulation sensitivity in cochlear implant listeners: stochastic resonance in a prosthetic sensory system? J Assoc Res Otolaryngol. 2001;2:159-71

Chatterjee M, Oba SI. Across- and within-channel envelope interactions in cochlear implant listeners. J Assoc Res Otolaryngol. 2004;5:360-75.

Chatterjee M, Oba SI. Noise improves modulation detection by cochlear implant listeners at moderate carrier levels. J Acoust Soc Am. 2005;118:993-1002.

Chatterjee M, Oberzut C. Multi-channel interactions in amplitude modulation detection and discrimination by cochlear implant listeners. *2009 Conference on Implantable Auditory Prostheses,* Lake Tahoe, CA. 2009;218.

Chatterjee M, Oberzut C. Detection and rate discrimination of amplitude modulation in electrical hearing. J Acoust Soc Am. 2011;130:1567-80.

Chatterjee M, Peng SC. Processing F0 with cochlear implants: Modulation frequency discrimination and speech intonation recognition. Hear Res. 2008; 235:143-56.

Chatterjee M, Shannon RV. Forward masked excitation patterns in multielectrode electrical stimulation. J Acoust Soc Am. 1998;103:2565–72

Chatterjee M, Yu J. A relation between electrode discrimination and amplitude modulation detection by cochlear implant listeners. J Acoust Soc Am; 2010; 127:415-26.

Chatterjee M, Galvin JJ 3rd, Fu QJ, Shannon RV. Effects of stimulation mode, level and location on forward-masked excitation patterns in cochlear implant patients. J Assoc Res Otolaryngol. 2006;7:15-25.

Colletti V, Shannon RV. Open set speech perception with auditory brainstem implant? Laryngoscope 2005; 115:1974-78.

Dai H, Micheyl C. On the choice of adequate randomization ranges for limiting the use of unwanted cues in same-different, dual-pair, and oddity tasks. Atten Percept Psychophys. 2010;72:538-547

Dau T, Kollmeier B, Kohlrausch A. Modeling auditory processing of amplitude modulation. I. Detection and masking with narrow-band carriers. J Acoust Soc Am. 1997a;102:2892-2905.

Dau T, Kollmeier B, Kohlrausch A. Modeling auditory processing of amplitude modulation. II. Spectral and temporal integration. J Acoust Soc Am. 1997b;102:2906-19.

Deroche ML, Lu HP, Limb CJ, Lin YS, Chatterjee M. Deficits in the pitch sensitivity of cochlear-implanted children speaking English or Mandarin. Front. Neurosci. 2014. doi: 10.3389/fnins.2014.00282.

Deroche, M., Zion DJ, Schurman JR, Chatterjee M. Sensitivity of school-aged children to pitch-related cues. J Acoust Soc Am. 2012;131: 2938-47.

Donaldson GS, Viemeister NF. Intensity discrimination and detection of amplitude modulation in electric hearing. J Acoust Soc Am. 2000;108:760–3.

Drennan WR, Pfingst BE. Current-level discrimination in the context of interleaved, multichannel stimulation in cochlear implants: effects of number of stimulated electrodes, pulse rate, and electrode separation. J Assoc Res Otolaryngol. 2006;7:308-16.

Fraser M, McKay CM. Temporal modulation transfer functions in cochlear implantees using a method that limits overall loudness cues. Hear Res. 2012;283:59-69.

Friesen LM, Shannon RV, Baskent D, Wang X. Speech recognition in noise as a function of the number of spectral channels: comparison of acoustic hearing and cochlear implants. J Acoust Soc Am. 2001; 110(2):1150-63.

Fu QJ. Temporal processing and speech recognition in cochlear implant users Neuroreport 2002.13:1635–40.

Fu QJ, Chinchilla S, Galvin JJ. The role of spectral and temporal cues in voice gender discrimination by normal-hearing listeners and cochlear implant users. J Assoc Res Otolaryngol. 2004;5:253-60.

Fu QJ, Chinchilla S, Nogaki G, Galvin JJ 3rd. Voice gender identification by cochlear implant users: the role of spectral and temporal resolution. J Acoust Soc Am. 2005;118:1711-8.

Fuller CD, Gaudrain E, Clarke JN, Galvin JJ, Fu QJ, Free RH, Başkent D. Gender categorization is abnormal in cochlear implant users. J Assoc Res Otolaryngol. 2014;15:1037-48.

Galvin JJ 3rd, Fu QJ. Effects of stimulation rate mode and level on modulation detection by cochlear implant users. J Assoc Res Otolaryngol. 2005;6:269–79.

Galvin JJ 3rd, Fu QJ. Influence of stimulation rate and loudness growth on modulation detection and intensity discrimination in cochlear implant users. Hear Res. 2009;250:46–54.

Galvin JJ, Fu QJ, Oba S, Başkent D. A method to dynamically control unwanted loudness cues when measuring amplitude modulation detection in cochlear implant users. J Neurosci Methods. 2014;222: 207–12.

Galvin JJ 3rd, Oba S, Fu QJ, Başkent D. Single- and multi-channel modulation detection in cochlear implant users. PLoS One. 2014;11; 96:e99338.

Galvin JJ 3rd, Oba S, Başkent D, Fu QJ. Modulation frequency discrimination with single and multiple channels in cochlear implant users. Hear Res. 2015;324:7-18.

Garadat SN. Zwolan TA, Pfingst BE. Across-site patterns of modulation detection: relation to speech recognition J Acoust Soc Am. 2012;131:4030-41.

Garadat SN, Zwolan TA, Pfingst BE. Using temporal modulation sensitivity to select stimulation sites for processor MAPs in cochlear implant listeners. Audiol Neurootol. 2013;18:247-60.

Geurts L, Wouters J. Coding of the fundamental frequency in continuous interleaved sampling processors for cochlear implants. J Acoust Soc Am. 2001;109:713-26.

Green, T, Faulkner, A, Rosen, S. Enhancing temporal cues to voice pitch in continuous interleaved sampling cochlear implants. J Acoust Soc Am. 2004;116, 2298-2310.

Green T, Faulkner A, Rosen S, Macherey O. Enhancement of temporal periodicity cues in cochlear implants: effects on prosodic perception and vowel identification. J Acoust Soc Am. 2005;118:375-85.

Green T, Faulkner A, Rosen S. Variations in carrier pulse rate and the perception of amplitude modulation in cochlear implant users. Ear Hear. 2012;33:221-30.

Hamilton N, Green T, Faulkner A. Use of a single channel dedicated to conveying enhanced temporal periodicity cues in cochlear implants: effects on prosodic perception and vowel identification. Int J Audiol. 2007;46:244-53.

Jesteadt W. An adaptive procedure for subjective judgments. Percept Psychophys. 1980;28:85-8.

Kreft HA, Oxenham AJ, Nelson DA. Modulation rate discrimination using half-wave rectified and sinusoidally amplitude modulated stimuli in cochlear-implant users. J Acoust Soc Am. 2010;127:656-9.

Kreft HA, Nelson DA, Oxenham AJ. Modulation frequency discrimination with modulated and unmodulated interference in normal hearing and in cochlear-implant users. J Assoc Res Otolaryngol. 2013;4:591-601.

Landsberger DM. Effects of modulation wave shape on modulation frequency discrimination with electrical hearing. J Acoust Soc Am. 2008;124:EL21-7.

Landsberger DM, Srinivasan AG. Virtual channel discrimination is improved by current focusing in cochlear implant recipients. Hear Res. 2009;254:34–41.

Landsberger DM, Padilla M, Srinivasan AG. Reducing current spread using current focusing in cochlear implant users. Hear Res. 2012; 284:16-24.

Levitt H. Transformed up-down methods in psychoacoustics. J Acoust Soc Am. 1971;49: Supp 2, 467.

Litvak LM, Spahr AJ, Emadi G. Loudness growth observed under partially tripolar stimulation: model and data from cochlear implant listeners. J Acoust Soc Am. 2007;122:967–81.

Luo X, Fu QJ, Wei CG, Cao KL. Speech recognition and temporal amplitude modulation processing by Mandarin-speaking cochlear implant users. Ear Hear. 2008;29:957-70.

Luo X, Fu QJ, Galvin JJ. Vocal emotion recognition by normal-hearing listeners and cochlear implant users. Trends Amplific. 2007;11:301-15.

Luo X, Galvin JJ, Fu QJ. Effects of stimulus duration on amplitude modulation processing with cochlear implants. J Acoust Soc Am. 2010;127:EL23-9.

Macherey O, Deeks JM, Carlyon RP. Extending the limits of place and temporal pitch perception in cochlear implant users. J Assoc Res Otolaryngol. 2011;12:233–51

McKay CM, Henshall KR. Amplitude modulation and loudness in cochlear implantees J Assoc Res Otolaryngol. 2010;11:101-111.

McKay CM, Remine MD, McDermott HJ. Loudness summation for pulsatile electrical stimulation of the cochlea: effects of rate, electrode separation, level, and mode of stimulation. J Acoust Soc Am. 2001; 110:1514-24.

McKay CM, Henshall KR, Farrell RJ, McDermott HJ. A practical method of predicting the loudness of complex electrical stimuli. J Acoust Soc Am. 2003;113:2054-63.

Middlebrooks JC. Effects of cochlear-implant pulse rate and inter-channel timing on channel interactions and thresholds. J Acoust Soc Am. 2004;116:452–68

Middlebrooks JC, Snyder RL. Selective electrical stimulation of the auditory nerve activates a pathway specialized for high temporal acuity. J Neurosci; 2010;30:1937–46.

Pfingst BE, Xu L, Thompson CS. Effects of carrier pulse rate and stimulation site on modulation detection by subjects with cochlear implants. J Acoust Soc Am. 2007;121:2236–46.

Rosen S. Temporal information in speech: acoustic, auditory and linguistic aspects. Philos Trans R Soc Lond B Biol Sci. 1992;336:367-73.

Shannon RV. Temporal modulation transfer functions in patients with cochlear implants J Acoust Soc Am. 1992;91:2156-64.

Shannon RV, Zeng FG, Kamath V, Wygonski J, Ekelid M. Speech recognition with primarily temporal cues. Science 1995; 270(5234):303-4.

Shannon RV, Fu QJ, Galvin J 3rd. The number of spectral channels required for speech recognition depends on the difficulty of the listening situation. Acta Otolaryngol Suppl. 2004;552:50-4.

Snyder RL, Bierer JA, Middlebrooks JC. Topographic spread of inferior colliculus activation in response to acoustic and intracochlear electric stimulation. J Assoc Res Otolaryngol. 2004;5:305–22.

Vandali A, Sly D, Cowan R, van Hoesel R. Pitch and loudness matching of unmodulated and modulated stimuli in cochlear implantees. Hear Res. 2013;302:32-49.

Wilson BS, Finley CC, Lawson DT, Wolford RD, Eddington DK, Rabinowitz WM. Better speech recognition with cochlear implants. Nature 1991;352:236-8.

Won JH, Drennan WR, Nie K, Jameyson EM, Rubinstein JT. Acoustic temporal modulation detection and speech perception in cochlear implant listeners. J Acoust Soc Am 2011; 130:376-388.

Wygonski J, Robert M. House Ear Institute Nucleus Research Interface (HEINRI) specification - internal materials. 2002.

Zeng FG, Turner CW. Binaural loudness matches in unilaterally impaired listeners Quarterly J. Exp. Psych; 1991; 43: 565-583.

Zhou N, Pfingst BE. Psychophysically based site selection coupled with dichotic stimulation improves speech recognition in noise with bilateral cochlear implants. J Acoust Soc Am. 2012 Aug;132(2):994-1008.

Zhou N, Pfingst BE. Effects of site-specific level adjustments on speech recognition with cochlear implants. Ear Hear. 2014;35:30-40.

Chapter 8: Summary

Chapter 1 - Introduction

Cochlear implants (CIs) provide hearing to profoundly deaf individuals by electric stimulation of the remaining auditory neurons. In typical CI signal processing, pulse trains are delivered to 12 – 22 implanted electrodes, modulated by the temporal envelope extracted from the acoustic frequency analysis band associated with each electrode. As such, CIs provide both spectral cues associated with the place of stimulation in the cochlea and temporal cues, which represent envelope information below 50 Hz and (to some extent) periodicity cues below 500 Hz. There is a tradeoff between spectral and temporal cues, with temporal cues becoming more important as the number of spectral channels is reduced. Because of electrode interactions, CI users seem able to access only about 8 spectral channels, despite being provided with many more channels.

Given the limited functional spectral resolution, temporal envelope information is important for CI speech perception. As opposed to other measures of temporal processing (e.g., gap detection, pulse rate discrimination, etc.), single-channel measures of temporal envelope perception – amplitude modulation (AM) detection and AM frequency discrimination – have been correlated with various measures of speech perception in CI users. But in everyday use of their device, CI users receive multi-channel stimulation. While one could measure perception of speech envelopes directly, top-down processes related to speech pattern perception may obscure the psychophysical limits of temporal envelope perception. It is important to know these limits to improve and/or optimize CI signal processing. While single-channel temporal envelope perception has been correlated to speech perception in CI users, it is unclear how it is affected by multi-channel stimulation. For example, AM detection and discrimination are known to be level-dependent. Because of multi-channel loudness summation, current levels must be reduced to obtain a functional DR with multi-channel stimulation. These reduced current levels may affect AM detection. In previous studies, significant differences in AM detection have been observed across single electrodes; it is unclear how these across-site differences might contribute to the multi-channel percept (e.g., attending to the best channel, interference by the worst channel, etc.).

While single-channel temporal envelope perception has been widely studied by CI researchers, there is little research regarding AM detection and discrimination with multiple channels. One major difficulty in such research is access and expertise with CI research interfaces, which bypass CI users' clinical processors and allow for precise control over stimulation parameters such as current level, electrode selection, etc. It is important to directly control these parameters and avoid the effects of microphone, gain control, analysis filters, and amplitude mapping, all of which can distort psychophysical measures and may only reflect the limits of the clinical processor. It is important to understand the limits of perception so that clinical processing can be improved and optimized for CI users. Thus, the present experiments were conducted with a research interface. The present studies also included novel experimental procedures, at least in relation to previous AM detection and discrimination studies.

In this thesis, we explore temporal envelope perception with single and multiple channels. In the first and second studies, we measure single- and multi-channel AM detection. In the third and fourth studies, we measure single- and multi-channel AM frequency discrimination. While AM detection and discrimination are related measures, they reflect different aspects of temporal envelope perception. For example, one can discriminate between AM frequencies only if the AM is sufficiently deep (i.e., well above the AM detection threshold). Similarly, the ability to detect AM may not fully reflect the ability to track changes in AM frequency (which are important to perceive voice pitch, speech prosody, etc.).

Chapter 2 - A method to dynamically control unwanted loudness cues when measuring amplitude modulation detection in cochlear implant users
(Galvin et al., 2014; J Neurosci Methods. 222:207-212).

When measuring AM detection, listeners can use an envelope cue (related to the AM rate) and a loudness cue (related to the peak current level of the AM stimulus). Because we were most interested in listeners' sensitivity to the envelope cue, it is important to control for the peak AM loudness cue. We designed and evaluated a method to dynamically control for peak AM loudness cues while adaptively measuring single-channel AM detection.

To address this issue, we designed and evaluated a method to adaptively measure MDTs (as in many previous CI studies) while controlling for potential loudness cues in the AM stimulus. When adaptively measuring MDTs, the AM depth is adjusted from trial to trial according to the correctness of the response. As such, the potential loudness cue associated with the AM depth may also change from trial to trial, with greater loudness cues associated with larger AM depths. For this study, we designed and evaluated a method adjust the current level of the non-AM stimulus to match the loudness of the AM stimulus for each trial of an adaptive MDT measurement (i.e., a dynamic control for AM loudness cues during an adaptive MDT test). First, AM stimuli at relatively low and high stimulation rates (500 and 2000 pulses per second, or pps), AM frequencies (10 or 100 Hz) and AM depths (5, 10, 20, and 30% of the reference amplitude) were loudness-balanced to non-AM stimuli. During the loudness-balancing procedure, the current level of the non-AM stimulus was adjusted according to subject response, eventually converging on the level that matched the loudness of the AM stimulus. AM loudness compensation functions were then derived for individual subjects, and these functions were used to dynamically adjust the current level of the non-AM stimulus to match the loudness of the AM stimulus during the subsequent AM detection task. Current levels across stimuli were also roved within each trial to further obscure potential AM loudness cues. MDTs were adaptively measured with and without the loudness compensation function at various stimulation rates, AM frequencies, and presentation levels.

Results showed that MDTs generally worsened with the method used to control potential AM loudness cues. The effects of AM rate and presentation level on MDTs were similar with and without the loudness compensation algorithm. However, the effect of stimulation rate on MDTs was significant only when there was no control for AM loudness cues. Previous studies have shown significantly poorer MDTs with high rather than low stimulation rates; however, there was no control for potential AM loudness cues in these studies. The present data suggest that AM loudness cues may have contributed to previous studies' pattern of results. Thus, in an AM detection task, it is important to control for potential AM loudness cues to isolate listeners' sensitivity to temporal envelope (i.e., changes in amplitude over time rather than peak AM loudness). However, it should also be noted that various amounts of current level adjustments associated with the present AM loudness compensation and level roving may have simply added variability to the MDT measurements, especially at very low AM depths where the current level differences across stimuli may have been greater than those associated with the AM depth, which was the parameter of concern.

Chapter 3 - Single- and multi-channel modulation detection in cochlear implant users
(Galvin et al., 2014; PLoS One, 9(6):e99338)

While single-channel AM detection has been extensively studied in CI users, multi-channel AM detection may be more relevant to the real-life listening experience of CI users, given that CI users' temporal envelope processing involves multiple channels in everyday listening. Further, during clinical fitting of CIs, multi-channel loudness summation must be accommodated to ensure comfortable operating levels. Because previous single-channel studies have shown that MDTs depend strongly on stimulation level, it is unclear how accommodating multi-channel loudness summation may affect multi-channel modulation detection. Also, previous studies have shown across-site variability in single-channel MDTs; as such, it is unclear how these differences in single-channel AM detection contribute to the multi-channel percept. For example, do CI users cue to the channel with the best or worst temporal envelope sensitivity?

To address these issues, we measured single- and multi-channel AM detection in CI users for a range of AM frequencies and presentation levels. Multi-channel stimuli were comprised of 4 maximally spaced electrodes; the component electrodes were loudness-balanced to one another. The multi-channel stimuli were then loudness-balanced to a single-channel reference; current levels on each channel were reduced to accommodate multi-channel loudness summation. MDTs were then adaptively measured for single- and multi-channel stimuli, with and without adjustment for multi-channel loudness summation. With the adjustment, single- and multi-channel stimuli were equally loud; without the adjustment, multi-channel stimuli were louder than the single-channel stimuli. MDTs were measured using the dynamic control for AM loudness cues, as described in Chapter 2.

Results showed that both single- and multi-channel modulation detection were significantly affected by presentation level and AM rate, consistent with previous single-channel studies. Results showed significant across-site variability in single-channel MDTs, also consistent with previous studies. At equal loudness, MDTs were <u>significantly poorer</u> with multiple than with single channels, due to the reduced current levels needed to accommodate multi-channel loudness summation. With no compensation for multi-channel loudness summation, MDTs were <u>significantly better</u> with multiple channels than with even the best single channel, due to increased loudness and/or to multiple representations of temporal envelope information across channels. The pattern of results indicated that CI subjects optimally combined envelope information across channels, rather than cue to channel with the best or worst MDT, or to the average MDT across channels. The results also suggest that single-channel measures may over-estimate CI users' multi-channel AM detection when multi-channel loudness summation is considered. Note that the current level adjustments needed to compensate for multi-channel loudness summation differed greatly across subjects. While the amount of summation adjustment was not significantly correlated with multi-channel MDTs, subjects with the least amount of adjustment tended to have better multi-channel MDTs, most likely due to the higher current levels on each channel.

Chapter 4 - Modulation frequency discrimination with single and multiple channels in cochlear implant users
(Galvin et al., 2015; Hear Res, 324: 7-18)

Detection of AM and discrimination of AM frequency represent two aspects of temporal envelope perception that have been correlated with various types of speech perception in CI users (e.g., phoneme recognition, prosody perception, lexical tone perception). To discriminate AM frequency, one must be able to reliably perceive AM (i.e., AM depths well above MDT). As explained in Chapter 3, when measuring AM detection, peak AM loudness cues may contribute to MDTs, making it difficult to isolate envelope detection from loudness discrimination. Efforts to control for these AM loudness cues may introduce unwanted variability in MDT measures. Given sufficient AM depth, AM frequency discrimination is less susceptible to AM loudness cues and thus may better reflect CI users' temporal envelope perception. AM frequency discrimination is also not susceptible to loudness cues associated with multi-pulse integration when measuring stimulation rate discrimination (another measure of temporal processing that is significantly correlated with AM frequency discrimination). Also, CI users must often perceive dynamic changes in AM frequency associated with voice pitch, vowel-consonant transitions, etc.

Similar to AM detection, AM frequency discrimination in CI users has been shown to be significantly affected by reference AM rate and current level. AM frequency discrimination thresholds have been previously shown to be better with multiple than with single channels, presumably due to the multiple representations of envelope information across channels. However, as noted in Chapter 3, multi-channel loudness summation can greatly affect multi-channel AM detection. Thus, it is important to control for multi-channel loudness summation to determine the source of any multi-channel advantage in temporal envelope perception. Also, it is important to understand how channel interaction may contribute to multi-channel envelope perception. When the same temporal envelope is presented to multiple channels, widely spaced channels may provide relatively independent representations while narrowly spaced channels would not. Across-site differences in temporal envelope processing may also interact with electrode spacing in multi-channel perception. Temporal envelope perception may be more similar across narrowly spaced channels that target the name neural region, but may be different across widely spaced channels that target different neural regions. It is unclear how multi-channel AM frequency discrimination might be affected by loudness summation and electrode spacing.

To address these issues, we measured single- and multi-channel AM frequency discrimination in CI users with and without current level adjustments to compensate for multi-channel loudness summation. The electrode spacing was varied to be wide or narrow to target different neural regions. Single-channel stimuli were first loudness-balanced to one another; multi-channel stimuli were then loudness-balanced to the single-channel stimuli by reducing the current levels on each channel. AM frequency discrimination was measured with the maximum modulation depth (i.e., modulation was between detection threshold and maximum acceptable loudness), ensuring very salient envelope information. The reference AM rate was 100 Hz; the target AM rate was 101, 102, 104, 108, 116, 132, 164, 228, or 356 Hz. For multi-channel stimuli, coherent AM was applied across all 3 channels. AM frequency discrimination was measured for single- and multi-channel stimuli at the same loudness; single-channel AM frequency discrimination was also measured at the reduced summation-adjusted current levels used for the multi-channel stimuli. Rather than using an adaptive procedure (as in the above AM detection studies in Chapters 2 and 3), a non-adaptive procedure (method of constant stimuli was used to measure AM frequency discrimination. The non-adaptive procedure allowed for psychometric functions to be fit to the data, which can be used to characterize temporal envelope perception beyond threshold, especially when thresholds are difficult to obtain (e.g., due to low current levels).

Results showed that, with no compensation for multi-channel loudness summation, AMFD thresholds were significantly better with multiple than with single channels, consistent with previous studies. Note that in this condition, multi-channel stimuli were much louder than the single-channel stimuli. When single- and multi-channel stimuli were equally loud, there were no significant differences in AM thresholds. This finding was not consistent with previous studies that demonstrated a multi-channel advantage for AM frequency discrimination. The present AM frequency discrimination results were also markedly different from the above multi-channel MDT data (Chapter 3), in which AM detection was significantly poorer with multiple than with single channels at equal loudness. No significant differences in multi-channel AM frequency discrimination were observed between widely and narrowly spaced electrodes. It is worth noting that at the summation-adjusted current levels, single-channel AM frequency discrimination was quite poor. When multiple channels were added, performance sharply improved, whether due to the increased loudness or to the multiple representations of envelope information across the cochlea. These results suggest that loudness may be the strongest factor in CI users'

AM frequency discrimination, rather than current level and/or the number of channels stimulated.

Chapter 5 - Envelope interactions in multi-channel amplitude modulation frequency discrimination by cochlear implant users (Galvin et al., 2015; PLoS One, 10(10):e0139546)

In Chapter 4, there was no multi-channel advantage in AM frequency discrimination when single- and multi-channel stimuli were equally loud. The results also showed AM frequency discrimination could be greatly improved when multiple channels with coherent AM were added to a poorly performing single AM channel. It is unclear whether this multi-channel advantage was due to increased loudness, multiple envelope representations, or component channels with better temporal processing. It was difficult to observe single-channel contributions to multi-channel perception in Chapter 4. At the reduced, summation-adjusted current levels, single-channel performance was generally poor. At the higher current levels, performance was very good. In both cases, there was little difference in single-channel AM frequency discrimination across stimulation sites. A different approach would be to deliver the target AM to only 1 of 3 channels (rather than to all 3 channels as in Chapter 4). In this manipulation, the multi-channel stimulus would be sufficiently loud, allowing channel-specific contributions to multi-channel AM frequency to be observed. Also, while AM frequency discrimination may be enhanced by delivering coherent AM to multiple channels, CI users frequently encounter different envelope information delivered to multiple channels and experience difficulty in segregating these envelopes. Thus, it seems important to study multi-channel perception of coherent AM (as in Chapter 4) and multi-channel perception of competing envelope information (as in this study), as both involve combining envelope information across channels.

In this study, multi-channel AM frequency discrimination was measured using the same CI subjects, similar stimuli, and the same procedures as in the previous study in Chapter 4. For the reference stimulus, 100 Hz AM was delivered to all 3 channels. Different from the previous study, in which coherent AM was delivered to all 3 channels, the target AM (which ranged from 101-356 Hz) was delivered to 1 of 3 channels and the reference AM (100 Hz) was delivered to the other 2 channels. As in the previous study, the spacing between electrodes was varied to be wide or narrow to test different degrees of channel interaction.

Results showed that CI subjects were highly sensitive to interactions between the reference and target envelopes, with AM frequency discrimination similar to and sometimes better than observed in the previous study (Chapter 4). However, different from the previous study, multi-channel AM frequency discrimination was non-monotonic as a function of target AM frequency. For the wide spacing, there was significantly less envelope interaction when the target AM was delivered to the basal channel (rather than delivered to apical or middle channel), suggesting some site-specific effects when channels were combined. For the narrow spacing, there was no effect of target AM channel, due to the limited neural region that was stimulated.

The present AM frequency data were also compared to those from Chapter 4 (in which the target AM was delivered to all 3 channels). For very small differences in AM frequency (2 – 4 Hz), there was often greater sensitivity when the target AM was delivered to 1 of 3 channels rather than to all 3 channels, especially for narrowly spaced electrodes. Because the multiple channels were interleaved in time, this suggests some sort of interaction at the neural level. For relatively large differences in AM frequency (> 32 Hz), there was a small (but not significant) advantage when the target AM was delivered to all 3 channels, rather than to 1 of 3 channels. In the previous study, single-channel AM frequency discrimination was poor at the reduced summation-adjusted current levels. Performance greatly improved when 2 AM channels were added to the single AM channel, whether the additional channels contained coherent

AM (as in Chapter 4) or non-coherent AM (as in this study), relative to the target AM channel.

Both the present and the previous study used a discrimination task ("which of the 3 intervals is different?") to measure AM frequency discrimination; the target AM of the probe stimulus was varied in both studies. However, in Chapter 4, subjects most likely cued to differences in pitch between the reference and probe stimuli. In this study, subjects most likely cued to the envelope interactions between the reference and probe stimuli (rather than the pitch difference). Sensitivity to these different cues was reflected in the different psychometric functions between the previous and present study. Because the effect of target AM frequency was non-monotonic in this study, adaptive procedures (as used in most previous AM studies with CI users) may not be suitable to measure AM frequency discrimination with interfering envelopes. Thus, the non-adaptive procedure used in this study and the associated psychometric functions allowed for insights regarding envelope interactions that may not have been revealed with an adaptive procedure (as is used in many previous AM frequency discrimination studies).

The present results also suggest that multiple envelope representations (in addition to overall loudness) may contribute to the multi-channel advantage observed in previous AM frequency discrimination studies. The different patterns of results for the wide and narrow spacing suggest a peripheral contribution to multi-channel temporal processing. Overall, envelope interactions among multiple channels appear to quite complex, depending on the envelope information presented to each channel and the relative independence of the stimulated channels.

Chapter 6 – General discussion

These series of experiments shed light on differences in CI users' temporal envelope perception between single- and multi-channel stimulation, as well as some differences between

AM detection and discrimination. Historically, single-channel measures have been more common in CI research, mostly because of difficulties associated with multi-channel experiments. Many CI researchers also do not have access to or sufficient experience with research interfaces to directly control stimulation parameters. CI research interfaces, as opposed to clinical processors, should be used to measure CI users' psychophysical limits, as signals will be distorted by the clinical processors' amplitude mapping, frequency allocation, AGC, etc. This is especially relevant when measuring temporal envelope perception, as changes in a signal's AM depth are most likely distorted by clinical processor settings. It is also difficult to target electrodes with a clinical processor, due to the analysis filter bandwidths which can result in multi-electrode stimulation when targeting a single electrode. While some CI researchers have access to research interfaces, relatively few have pursued multi-channel psychophysics. As such, there often remains a disconnect between many single-channel measures (e.g., electrode pitch ranking, stimulation rate discrimination, detection thresholds, DR, etc.) and multi-channel speech performance. Some of this may be explained by top-down processes associated that allow for robust speech recognition despite distortions to the speech patterns and/or psychophysical deficits. But relatively little is known regarding psychophysical perception with three or more channels. It is important to know the CI users' psychophysical limits in both single- and multi-channel contexts to improve and/or optimize CI signal processing. As such, the studies presented here are a first step toward better understanding CI users' perceptual limits in a broader context.

Overall, these studies show that multi-channel loudness summation should be strongly considered when extrapolating temporal envelope perception with multiple channels from single-channel measures. CI users may differ greatly in terms of loudness summation, though the sources of these differences are as yet unclear. The effects of summation on AM detection and frequency discrimination may also differ across patients. Given that clinical fitting must accommodate multi-channel summation, and given that single-channel temporal envelope perception has correlated with multi-channel speech perception, multi-channel psychophysics must carefully control for loudness summation to bridge the gap between simple psychophysics and complex speech perception. The present studies also show that the contribution of single channels to the multi-channel percept depends on the listening task. For example, across site-variability was a factor in multi-channel perception for Chapter 5 (where different AM frequencies were delivered to each channel), but not for Chapters 3 and 4 (where coherent AM was delivered across channels).

Further studies are required to better understand the effects of loudness summation on complex multi-channel perception. Multiple factors related to the CI can affect the overall loudness. Adding channels will increase loudness; compensating for multi-channel summation will reduce current levels on each channel, which, depending on the specific measure used, may or may not reduce temporal envelope perception. Adding channels will also increase channel interaction, a major limiting factor in CI performance. Increasing stimulation rates will also increase loudness, as well as channel interaction. Present-day CI signal processing provides many more channels than can be functionally accessed, using stimulation rates seemingly much higher than needed to effectively code speech perception. There is a balance that has yet to be struck among the optimal number of channels, spacing of channels in each stimulation frame, stimulation rates on each channel, and current levels delivered to each channels. Because all these factors will interact in a multi-channel context, CI research and development should psychophysically evaluate the effects of stimulation parameters within a multi-channel context. This is no easy task, requiring specialized hardware (research interfaces), complex experimental designs and careful control of stimuli. These present studies are by no means comprehensive, but represent important steps forward toward understanding mutli-channel temporal envelope perception.

Given the effects of loudness summation and channel interaction observed in these studies, there are some clinical recommendations that might improve multi-channel temporal envelope perception. Loudness summation may result in reduced current levels on each channel, which in turn will reduce AM detection. Channel interaction will increase temporal envelope interference. It may be preferable to stimulate fewer, widely spaced electrodes within each stimulation frame, thereby reducing multi-channel summation and channel interaction. To further protect against temporally induced channel interaction and loudness summation associated with multi-pulse integration, lower stimulation rates should be used. Previous studies have shown no advantage for speech perception with high stimulation rates, and a deficit in single-channel AM detection with high rates. Lower stimulation rates would increase current levels and reduce channel interaction, both of which would benefit temporal envelope perception according to the data from the present studies. Ultimately, such parameter manipulations should also be sensitive the speech information that is to be transmitted. Frequency importance functions for speech (e.g., speech intelligibility index, articulation index, etc.) may be useful in determining the most important channels to select within each stimulation frame. Present-day clinical speech processors stimulate all channels within each frame (variants of the CIS strategy for Advanced Bionics and Med-El devices) or subset of channels with the most energy (ACE strategy in Cochlear Corp. devices). Given the current spread and/or spread of excitation from each channel, such strategies cannot avoid the detrimental effects of channel interaction. By being mindful of the relative importance of the frequency information to be transmitted (e.g., formant information) as well as the potential for channel interactions, electrodes could be selected to better preserve both spectral and temporal envelope information. Again, intelligent selection of fewer channels within each frame would reduce loudness summation and channel interaction and hopefully increase the functional throughput for CI users beyond the seemingly hard limit of 8 channels.

Chapter 9: Samenvatting

Hoofdstuk 1 – Introductie

Cochleaire implantaten (CI) verbeteren het gehoor van ernstig slechthorenden door elektrische stimulatie van de overgebleven auditieve neuronen. De signaalverwerking van de CI omvat elektrische pulsen die aan 12 tot 22 elektroden worden afgeleverd, nadat het signaal is gemoduleerd op basis van de temporele envelop die van ieder akoestische frequentie is geabstraheerd per elektrode. Door deze signaalverwerking geven CI's twee soorten signalen: een spectrale, gebonden aan de locatie van stimulatie in de cochlea, en een temporele, welke de informatie van het signaal onder de 50Hz en de periodiciteit onder de 500Hz representeren. In deze signaalverwerking is een wisselwerking aanwezig: als het aantal spectrale kanalen (zoals in een CI) afneemt, neemt de rol van de temporele informatie voor perceptie toe. Door de interactie tussen de elektrodes kunnen CI-gebruikers slechts gebruik maken van ongeveer 8 spectrale kanalen, hoewel de CI zelf 12 tot 22 elektroden en daarmee mogelijke kanalen bevat.

Als gevolg van de beperkte spectrale resolutie van de CI, is de temporele informatie erg belangrijk voor het spraakverstaan van CI-gebruikers. In tegenstelling tot andere maatstaven voor temporele verwerking (zoals detectie van gaten in een akoestisch signaal of herkenning van pulsen), kunnen temporele metingen van een enkel kanaal – zoals detectie van amplitudemodulaties en frequentiedetectie door amplitudemodulatie – zijn gecorreleerd met meerdere maten van spraakverstaan. In het dagelijks gebruik echter, wordt de CI op meerdere kanalen tegelijkertijd gestimuleerd. Hoewel men de waarneming van de spraakenvelop direct zou kunnen meten, zouden top-down processen gerelateerd aan de waarneming van spraak, de psychofysische beperkingen van de temporele envelop kunnen beïnvloeden. Het is daarom belangrijk om de psychofysische beperkingen te kennen om de signaalverwerking van de CI te kunnen optimaliseren of verbeteren. Hoewel de temporele envelop van een enkel kanaal is gecorreleerd aan het spraakverstaan van een CI-gebruiker, is het onduidelijk hoe dit wordt beïnvloed door de stimulatie van meerdere kanalen in de CI. Bijvoorbeeld, de detectie en discriminatie van de modulaties van de amplitude zijn afhankelijk van de ingestelde stroomlevels van de CI. Doordat de stimulatie van meerdere kanalen de luidheid kan beïnvloeden, worden de stroomlevels dusdanig ingesteld dat er een functionele range ontstaat voor de stimulatie van meerdere kanalen. Deze bijgestelde, meestal verlaagde levels kunnen de amplitudemodulatie detectie (AM) beïnvloeden. In voorgaande studies zijn significante verschillen voor de AM detectie gemeten tussen verschillende elektroden. Het is echter onduidelijk hoe deze individuele verschillen bijdragen aan de algehele waarneming bij de stimulatie van meerdere elektroden (bijvoorbeeld: luistert de CI-gebruiker met name naar het beste kanaal; of hoe beïnvloedt het slechte kanaal de waarneming).

Terwijl de temporele perceptie voor een enkel kanaal uitvoerig is onderzocht in CI-gebruikers, is er weinig bekend over de AM detectie en discriminatie bij het stimuleren van meerdere kanalen. Twee factoren die het onderzoek hiernaar bemoeilijken, zijn de expertise en toegang tot CI-onderzoeksprocessoren, die de eigen klinische processor van de CI-gebruiker omzeilen. Met de onderzoeksprocessoren kan controle worden verkregen over de precieze stimulatie van elk kanaal, zoals de energielevels, de selectie van elektrodes, etc. Het is belangrijk om directe controle over deze parameters te hebben, om zo de microfoon, de stimulatiecontrole, de filters en de geïnstalleerde map van de CI te omzeilen, teneinde de echte psychofysische waarde van een elektrode te kunnen meten zonder de limitaties van de klinische processor. Het is van belang om de beperkingen van de echte perceptie te kunnen begrijpen en te weten, om met die kennis de klinische instellingen van de CI te kunnen optimaliseren. De huidige experimenten zijn met een onderzoeksinterface verricht om de stimultie-instelling direct te kunnen controleren, daarnaast bevatten de studies nieuwe, experimentele methodes in verhouding tot voorgaande studies naar AM detectie en discriminatie.

In dit proefschrift onderzoeken we de temporele perceptie met stimulatie van enkele en multipele kanalen. In de eerste en tweede studie, meten we de AM detectie van zowel een enkele, als multipele kanalen. In de derde en vierde studie, meten we de AM discriminatie van zowel een enkele, als multipele kanalen. Hoewel AM detectie en discriminatie aan elkaar gerelateerde uitkomsten zijn, zijn ze een weergave van andere aspecten van de temporele perceptie. Bijvoorbeeld, men kan enkel AM frequenties discrimineren als men ze kan detecteren; evenzo is AM detectie niet voldoende om veranderingen in AM te discrimineren (Dit is bijvoorbeeld belangrijk voor het waarnemen van de toonhoogte van de stem van een spreker of van de prosodie van spraak).

Hoofdstuk 2 – Een methode om luidheid dynamisch te controleren bij het meten van amplitude modulatie detectie in cochleair-implantaat gebruikers
(Galvin et al., 2014; J Neurosci Methods. 222:207-212).

Bij het meten van AM detectie, kunnen luisteraars enveloppe- (gerelateerd aan AM snelheid) of luidheidsaanwijzingen (gerelateerd aan de piek in het stroomlevel van de AM-stimulus) gebruiken. Het is belangrijk de piek van de luidheid te controleren, omdat we geïnteresseerd zijn in de gevoeligheid van de luisteraar voor de enveloppe. Wij hebben een methode ontworpen en geëvalueerd waarmee we dynamisch de piek in luidheid konden controleren, terwijl via een adaptieve methode de AM van een enkel kanaal werd gemeten.

Deze methode meet modulatie detectie drempels (MDD's) (zoals in vele voorgaande CI studies), terwijl voor potentiële luidheidsaanwijzingen wordt gecontroleerd in de AM-stimulus. Bij het adaptief meten van de MDD, wordt de diepte van de AM aangepast per taak naar aanleiding van het al dan niet correcte antwoord. Hierbij kan de potentiële luidheid die geassocieerd is met een hogere AM ook verschillen, met grotere luidheidsaanwijzingen bij een hogere AM. In deze studie, hebben we een methode ontwikkeld en geëvalueerd om de stroomlevels van de niet-AM-stimulus te matchen aan de luidheid van de AM-stimulus voor elke trial van een MDD meting (ook wel een dynamische controle voor de AM luidheidsaanwijzingen tijdens een adaptieve MDD test). Eerst werden AM stimuli met hoge en lage stimulatie snelheden (500 en 2000 pulsen per seconde (pps)), AM frequentie (10 of 100 Hz) en AM diepte (5, 10, 20 en 30% van de referentie amplitude) gebalanceerd voor de luidheid van de niet-AM-stimulus. Tijdens het proces van het uitbalanceren van de luidheid, werd het level van de aangeboden stroom van de niet-AM-stimulus aangepast op basis van de reactie van de deelnemer om uiteindelijk samen te komen met de luidheid van de AM-stimulus. Op deze wijze werden luidheidscurves voor iedere individuele deelnemer verkregen. Deze curves werden vervolgens toegepast om op dynamische wijze het stroomniveau van de niet-AM-stimulus te matchen met de luidheid van de AM-stimulus tijdens de daaropvolgende AM detectie taak. De stroomlevels tussen de stimuli werden tot het maximum gebracht om mogelijke luidheidsaanwijzingen te detecteren. De MMD's werden vervolgens adaptief gemeten met en zonder de curve voor de compensatie van de luidheid bij verschillende stimulatie snelheden, AM frequenties en de luidheid.

De resultaten toonden dat MDD's over het algemeen slechter werden wanneer de luidheid van de AM werd gecontroleerd. Het effect van de AM snelheid en de luidheid van de MDD's was vergelijkbaar met en zonder de luidheidscompensatie. Echter, het effect van de stimulatiesnelheid op MDD's was alleen significant wanneer er niet werd gecontroleerd voor luidheid. Voorgaande studies hebben significant slechtere MDD's getoond met hogere stimulatiesnelheden, hoewel in deze studies niet voor het effect van luidheid op AM is gecontroleerd. De huidige data suggereren dat AM luidheidsaanwijzingen mogelijk hebben bijgedragen aan deze bevindingen in de voorgaande studies.

Concluderend, is het van belang om bij een AM detectie taak te controleren voor potentiële luidheidsaanwijzingen om de sensitiviteit van een individu voor de temporele enveloppe te kunnen bepalen (i.e. veranderingen in de amplitude door de tijd meer dan de piek in AM luidheid). Hierbij moet echter worden aangemerkt, dat de gevonden variabiliteit in MDD's mogelijk kan zijn veroorzaakt door de verschillende aanpassingen van de stroomlevels door de compensatie voor de AM luidheid en het aanpassen van het elektriciteitsniveau van trial tot trial. Met name wanneer de AM zeer lage waardes had en dus de stroomlevels tussen de verschillende stimuli groter waren dan wanneer ze alleen geassocieerd waren met de AM waarde, kan dit de MDD hebben beïnvloed.

Hoofdstuk 3 – Enkele en multipele kanaal modulatie detectie door cochleair-implantaat gebruikers
(Galvin et al., 2014; PLoS One, 9(6):e99338)

Hoewel AM van een enkel kanaal uitgebreid onderzocht is in CI-gebruikers, is AM-detectie van meerdere kanalen waarschijnlijk meer relevant voor de dagelijks luisterervaring van de CI-gebruikers, met name omdat de temporele waarneming gebaseerd is op de stimulatie van multipele kanalen. Tevens wordt tijdens de klinische aanpassing van de CI, de luidheid van meerdere kanalen tegelijkertijd bepaald om comfortabele luisterluidheid te bereiken. Eerdere studies hebben aangetoond dat in een enkel kanaal de MDD's sterk afhankelijk zijn van het niveau van de stimulatie. Het is echter onduidelijk hoe het afstellen van de luidheid op basis van meerdere kanalen tegelijkertijd de modulatie detectie van meerdere kanalen beïnvloedt. Tevens is er sprake van verschillen in de MDD's van de kanalen over de gehele electrode; hierdoor is onduidelijk hoe deze verschillen in AM detectie van een enkel kanaal interfereren met de perceptie bij multi-kanaal stimulatie. Bijvoorbeeld, luistert de CI-gebruiker naar het kanaal met de beste of de slechtste temporele resolutie?

Om een antwoord te vinden op deze vragen, hebben we AM detectie in CI-gebruikers per kanaal en met meerdere kanalen getest met een range van AM frequenties en luidheidsniveau's. Multi-kanaal stimuli werden samengesteld met behulp van 4 elektrodes die zo ver mogelijk van elkaar vandaan gelegen waren. Deze elektrodes werden gebalanceerd voor luidheid. Vervolgens werd de multi-kanaal stimulus voor luidheid gebalanceerd, gerefereerd aan een enkel-kanaal; de stroomlevels van ieder kanaal werden bijgesteld om de luidheid van multi-kanaal stimulatie mogelijk te maken. MDD's werden vervolgens op adaptieve wijze gemeten voor enkel en multi-kanaal stimuli, met en zonder bijstelling van de luidheid voor de multi-kanaal instelling. Met de bijstelling waren de stimulaties van een enkel kanaal en meerdere kanalen even luid. MDD's werden gemeten met de dynamische controle voor de AM luidheid, zoals beschreven in Hoofdstuk 2.

De resultaten toonden dat, zowel de enkele modulatie detectie, als de modulatie detectie van meerdere kanalen significant werden beïnvloed door het aanbiedingsniveau en de AM snelheid, zoals beschreven in voorgaande studies waarin alleen enkele kanalen werden getest. Een significant verschil tussen de verschillende plekken van stimulatie over de electrode bij eenkanaals MDD metingen werd gevonden, eveneens in overeenkomst met voorgaande studies. Bij gelijke luidheid, waren de MDD's significant slechter bij stimulatie van meerdere kanalen, dan bij de stimulatie van een enkel kanaal, op basis van de afgenomen stroomniveaus die nodig zijn voor de aanpassing van de luidheid bij stimulatie van meerdere kanalen. Zonder de compensatie voor de luidheid bij de multikanaal stimulatie, waren de MDD's significant beter bij de stimulatie van meerdere kanalen in vergelijking tot de stimulatie van een enkel kanaal. Waarschijnlijk berust dit op de toegenomen luidheid en/of de weergave van temporele informatie tussen de kanalen. Het patroon van de resultaten laat zien dat CI-gebruikers de enveloppe aanwijzingen van alle kanalen optimaal combineren, in plaats van zich op het beste of slechtste kanaal te richten of op het gemiddelde MDD tussen de kanalen. De resultaten laten verder zien dat metingen van een enkel kanaal wellicht een overschatting geven van de AM detectie met meerdere kanalen in CI-gebruikers wanneer men de resultaten met de compensatie voor de luidheid meeneemt. Hierbij moet wel worden meegenomen dat de compensatie van de luidheid tussen de verschillende CI-gebruikers erg verschillend was. Hoewel de mate van aanpassing van de luidheid niet gecorreleerd was aan de MDD's bij stimulatie van meerdere kanalen, hadden de deelnemers met de minste aanpassing van de luidheid over het algemeen een betere MDD, waarschijnlijk door de hogere stimulatie levels.

Hoofdstuk 4 – Modulatie frequentie discriminatie met enkel en multi kanaal stimulatie in CI-gebruikers
(Galvin et al., 2015; Hear Res, 324: 7-18)

Detectie van AM en discriminatie van de frequentie van de AM is een weergave van twee aspecten van de waarneming van de temporele enveloppe in CI-gebruikers, die gecorreleerd zijn aan verschillende onderdelen van spraakverstaan (bijvoorbeeld: foneemherkenning, prosodie waarneming, toontaalherkenning). Om de frequentie van AM te kunnen onderscheiden, moet men in staat zijn om AM waar te nemen (i.e. de diepte van de AM meer dan de MDD). Zoals in hoofdstuk 3 is uitgelegd, kunnen voor het betrouwbaar meten van de AM detectie, de piek van de AM en de luidheid hiervan invloed hebben op MDD metingen, waardoor het moeilijk is om detectie van de enveloppe te onderscheiden van de waarneming van luidheid. Maatregelen om voor de luidheid van de AM te controleren kunnen ongewild variatie in MDD metingen geven. Bij een toereikende AM diepte is AM frequentie discriminatie (AMFD) minder gevoelig voor de luidheid van AM en daarmee geeft AM diepte waarschijnlijk beter weer wat de daadwerkelijke temporele enveloppe van de CI-gebruiker is.
Wanneer discriminatie van het tempo van stimuleren wordt gemeten, een andere manier van temporele dataverwerking die erg lijkt op de AMFD, is AMFD bovendien niet vatbaar voor

luidheidsaanwijzingen met multi-puls integratie. Hoogstwaarschijnlijk ervaren CI-gebruikers dynamische veranderingen in AM frequentie gerelateerd aan de toonhoogte van de stem, de overgangen van klinkers naar medeklinkers, etc. Het is aangetoond dat AMFD, net als AM detectie, significant beïnvloed wordt door de referentie van de AM en de stroomlevels. Voorheen was al aangetoond dat de AMFD drempelwaarden beter zijn bij multikanaal stimulatie dan bij enkelkanaal stimulatie. Dit is waarschijnlijk te relateren aan het feit dat er meerdere representaties zijn van de enveloppe informatie, verspreid over meerdere kanalen. Echter, zoals uitgelegd is in hoofdstuk 3, kan de luidheidssummatie bij stimulatie van meerdere kanalen een aanzienlijk effect hebben op multikanaal AM detectie. Hiervoor is het dus belangrijk om controle te hebben over de luidheidssummatie om te bepalen waar betere perceptie kan worden bereikt (bij stimulatie van meerdere kanalen) met betrekking tot het waarnemen van de temporele enveloppe. Bovendien is het belangrijk om goed te begrijpen hoe kanaalinteractie bij kan dragen aan het waarnemen van de enveloppe over meerdere kanalen. Zodra dezelfde temporele enveloppe wordt aangeboden aan meerdere kanalen, krijgen namelijk ver uit elkaar staande kanalen relatief onafhankelijke representaties, terwijl dicht bij elkaar staande kanalen deze niet krijgen. In temporele signaalverwerking kunnen de verschillen binnen de elektrode ook afhankelijk zijn van de ruimte tussen de elektroden bij de perceptie van meerdere kanalen. Verschillende waarnemingen van de temporele enveloppe kunnen bijvoorbeeld meer op elkaar lijken in het geval van dicht bij elkaar staande elektroden (die hetzelfde neurale gebied stimuleren) ten opzichte van ver uit elkaar staande elektroden (die verschillende gebieden stimuleren). Het is tot dusver onduidelijk hoe AMFD daadwerkelijk afhankelijk is van luidheidssummatie en de afstand tussen de elektroden.

Om deze problemen te onderzoeken, hebben we enkel- en multikanaal AMFD in CI-gebruikers gemeten, zowel met als zonder aanpassingen aan de stroomlevels om te compenseren voor luidheidsummatie bij stimulatie van meerdere kanalen. Er is gevarieerd met de afstand tussen de elektroden, ver uit elkaar en dicht bij elkaar, met als doel verschillende neurale gebieden te stimuleren.

Enkelkanaal stimuli werden eerst genormaliseerd in luidheidsniveau aan elkaar. Vervolgens werden de multikanaal stimuli genormaliseerd (in luidheidsniveau) aan de enkelkanaal stimuli door de stroomlevels in elk afzonderlijk kanaal te reduceren. AMFD werd gemeten aan de hand van de maximale modulatiediepte (d.w.z., de modulatie zat tussen de detectiedrempels in en onder de maximaal aanvaarde luidheid), zodat de enveloppe informatie goed duidelijk gemaakt kon worden.

De referentie AM frequentie was 100 Hz; de target AM frequentie was 101 Hz, 102, 104, 108, 116, 132, 164, 228 of 356 Hz. Voor de stimuli van meerdere kanalen werd een samenhangende AM, dus eenzelfde temporele enveloppe, toegepast op elk van de drie kanalen. AMFD werd vervolgens gemeten voor enkel- en multikanaal stimuli met dezelfde luidheid. Enkelkanaal AMFD werd tevens gemeten met de op luidheid aangepaste stroomlevels, die gebruikt werden voor de multikanaal stimuli. In plaats van een adaptieve procedure (zoals in hoofdstuk 2 en 3) is ditmaal een non-adaptieve procedure gebruikt. In deze non-adaptieve procedure zijn constante stimuli gebruikt om AMFD te meten. Hierdoor was het mogelijk om psychometrische functies aan de data te fitten, die vervolgens gebruikt konden worden om de eigenschappen van de temporele enveloppe te bepalen buiten de drempelwaarden (vooral als deze drempelwaarden moeilijk te verkrijgen waren door b.v. lage stroomlevels). De resultaten van het experiment demonstreerden dat zonder compensatie voor de luidheidssummatie de AMFD drempelwaarden <u>significant beter</u> waren dan de waarden bij stimulatie van een enkel kanaal. In deze conditie waren de multikanaal stimuli echter wel veel luider dan enkelkanaal stimuli. Zodra enkel- en multikanaal stimuli genormaliseerd werden voor luidheid, waren er <u>geen significante verschillen</u> in AM drempelwaarden. Deze vondst was daardoor niet consistent met voorgaande onderzoeken. Die demonstreerden dat het hebben van multikanaal stimulatie juist wel een voordeel gaf bij AMFD. Het is van belang om op te merken dat de huidige resultaten van AMFD opmerkelijk verschillend waren van de data van hoofdstuk 3, waar de AM detectie significant slechter was met meerdere kanalen ten opzichte van de stimulatie van een kanaal bij gelijke luidheidsniveaus. Verder werden er geen significante verschillen gevonden met betrekking tot de afstand tussen de elektroden (ver uit elkaar en dicht bij elkaar). Het is belangrijk om te noemen dat met de voor luidheid aangepaste_stroomlevels, de AMFD van een enkel kanaal vrij slecht was. Zodra vervolgens meerdere kanalen werden toegevoegd, was er een sterke verbetering te zien in de prestaties van de proefpersonen. Het is niet duidelijk of dit veroorzaakt wordt door het verhoogde luidheidsniveau, ofwel door de meerdere representaties van enveloppe informatie over de gehele cochlea. De resultaten suggereren dat het luidheidsniveau waarschijnlijk de sterkste factor is als men kijkt naar de AMFD in CI-gebruikers, ten opzichte van de stroomlevels en/of de hoeveelheid kanalen die gestimuleerd worden.

Hoofdstuk 5 – Interacties van de enveloppe in multikanaal amplitude modulatie frequentie discriminatie door cochleair-implantaat gebruikers
(Galvin et al., 2015; PLoS One, 10(10):e0139546)

In hoofdstuk 4 werd geen voordeel van het stimuleren van meerdere kanalen voor AMFD gevonden wanneer de luidheidsniveaus van de enkel- en multikanaal stimuli even luid waren. De resultaten lieten bovendien zien dat AMFD significant verbeterd kan worden zodra men meerdere kanalen toe gaat voegen met samenhangende AM aan een enkel kanaal dat slecht presteert. Het is nog onduidelijk of het voordeel van de stimulatie van meerdere kanalen veroorzaakt werd door een toenemend luidheidsniveau, door de meerdere representaties van de enveloppe informatie, of door het kanaal in de serie met betere temporele signaalverwerking. Het was in hoofdstuk 4 dan ook lastig om waar te nemen wat de bijdrage van de kanalen op zich was aan de perceptie bij de stimulatie van meerdere kanalen. Bij de gereduceerde stroomlevels was de prestatie per kanaal over het algemeen slecht. Bij verhoogde stroomlevels was de prestatie zeer goed. Met zowel de aangepaste als de verhoogde stroomlevels was er weinig verschil tussen enkelkanaal AMFD tussen de verschillende plekken van stimulatie. Een andere aanpak zou zijn om de target AM aan te bieden aan slechts 1 kanaal (in tegenstelling tot alle 3 kanalen zoals in Hoofdstuk 4). In dit geval zal de stimulus van meerdere kanalen toereikend luid zijn, waardoor kanaal-specifieke contributies aan AM frequentie detectie bij meerdere kanalen goed kunnen worden waargenomen. Hoewel AMFD verbeterd kan worden door samenhangende AM aan te bieden aan meerdere kanalen, zullen CI-gebruikers vaak verschillende enveloppen aan meerdere kanalen aangeleverd krijgen. Daardoor zullen ze ervaren dat het moeilijk is om onderscheid te maken tussen deze enveloppen. Voor het onderscheiden van de enveloppen, is het van belang samenhangende AM (als in Hoofdstuk 4) en concurrerende enveloppe informatie (zoals in dit onderzoek) te onderzoeken, aangezien beide een rol spelen in het combineren van enveloppe-informatie over de verschillende kanalen.

In dit onderzoek werd de AMFD van verschillende kanalen gemeten met dezelfde CI-gebruikers, soortgelijke stimuli, en dezelfde procedures als in Hoofdstuk 4. Voor de referentiestimulus werd dit keer 100 Hz AM gebruikt, over alle drie de kanalen. Het verschil van deze studie ten opzichte van de vorige studie, waarin samenhangende AM werd gebruikt over alle drie de kanalen, was dat de target AM (101-356 Hz) werd aangeboden aan 1 van de 3 kanalen en de referentie AM (100Hz) aan de overige 2 kanalen. Net als in de vorige studie is er variatie in de afstand tussen de elektroden. Ze werden ver uit elkaar en dicht bij elkaar geplaatst om te zien of er interactie plaatsvond tussen de kanalen.

De resultaten demonstreerden dat CI-gebruikers zeer gevoelig zijn voor interacties tussen de target en de referentie enveloppe. AMFD was gelijk aan en in sommige gevallen zelfs beter dan in de voorgaande studie (Hoofdstuk 4). Echter, anders dan in het vorige onderzoek, was de AMFD bij stimulatie van meerdere kanalen niet een een-op-een functie van de target AM frequentie. Bij ver uit elkaar gelegen elektroden was er significant minder enveloppe interactie wanneer de target AM aangeboden werd aan het basale kanaal (ten opzichte van het apicale of het middelste kanaal). Deze vondst suggereert dat er sprake is van locatie-specifieke effecten wanneer meerdere kanalen worden gecombineerd. In het geval van de dicht bij elkaar gelegen elektroden was er geen effect van het target AM kanaal. Dit is te wijten aan de stimulatie van een enkel neuraal gebied.

De huidige AM-frequentie data werd ook vergeleken met de data uit de vorige studie in hoofdstuk 4 (waar de target AM werd aangeboden aan alle 3 de kanalen). Voor de kleine verschillen in AM-frequentie (2 – 4Hz) was er vaker een verhoogde gevoeligheid wanneer de target AM werd aangeboden aan 1 van de 3 kanalen dan aan alle 3 de kanalen. Dit effect was duidelijker voor de elektroden die dicht bij elkaar gelegen waren. Omdat de kanalen min of meer tegelijkertijd werden gestimuleerd, doet dit vermoeden dat het effect gebaseerd is op interactie op neuraal niveau. Voor de relatief grote verschillen in AM frequenties (> 32 Hz) werd een klein (maar niet significant) voordeel gevonden indien de AM target aan alle 3 de kanalen in plaats van aan 1 van de 3 kanalen werd aangeboden. In de voorgaande studie was AMFD van een enkel kanaal slecht bij de aangepaste stroomlevels. De prestaties werden beter zodra 2 AM kanalen werden toegevoegd aan het enkele AM kanaal. Dit gebeurde zowel wanneer de andere kanalen samenhangende AM bevatten (hoofdstuk 4) als bij niet-samenhangende AM (dit hoofdstuk), relatief gezien ten opzichte van het target AM kanaal.

Zowel deze, als de voorgaande studie maakten gebruik van een discriminatie taak (“welke van de 3 intervallen is verschillend?”)

om de AMFD te bepalen. De target AM van de te onderzoeken stimulus werd gevarieerd in beide studies. Echter, in hoofdstuk 4 richtten de deelnemers zich hoogstwaarschijnlijk op de verschillen in toonhoogte tussen de referentie en de te onderzoeken stimulus. In deze studie richtten de deelnemers zich waarschijnlijk op de enveloppe-interacties die plaatsvonden tussen de referentie en te onderzoeken stimulus (en dus niet de verschillen in toonhoogte). De gevoeligheid voor de verschillend (verschillen in toonhoogte versus enveloppe-interacties) wordt verklaard door verschillende psychometrische functies tussen de vorige en de huidige studie. Doordat het effect van de target AM frequentie niet een-op-een was in deze studie, zijn adaptieve procedures wellicht niet geschikt om AMFD te meten met interfererende enveloppen. Om deze reden is de non-adaptieve procedure gebruikt in deze studie en bieden de gerelateerde psychometrische functies de mogelijkheid om inzicht te verkrijgen over enveloppe-interacties die wellicht niet mogelijk waren in het geval van een adaptieve procedure (zoals gebruikt in veel AMFD studies).

De resultaten suggereren dat de representatie van meerdere enveloppen (als aanvulling op het algehele luidheidsniveau) wellicht een rol speelt in het voordeel van waarneming bij stimulatie van meerdere kanalen, zoals aangetoond in voorgaande studies over AMFD. De uiteenlopende resultaten bij verschillende afstanden tussen de elektroden suggereren dat er een bijdrage is van het perifere auditieve systeem aan temporele signaalverwerking bij stimulatie van meerdere kanalen.

Over het algemeen zijn de enveloppe-interacties over meerdere kanalen erg complex, afhankelijk van de enveloppe-informatie die aangeboden wordt aan elk kanaal en de relatieve onafhankelijkheid van de gestimuleerde kanalen.

Hoofdstuk 6 – Algemene discussie

Deze serie experimenten geeft inzicht in de verschillen tussen de temporele enveloppe-perceptie van CI-gebruikers bij het stimuleren van één of meerdere kanalen, alsmede in de verschillen tussen AM-detectie en discriminatie. Oorspronkelijk waren metingen van slechts één kanaal de norm binnen het CI-onderzoek, voornamelijk door de moeilijkheden die worden geassocieerd met experimenten waarbij meerdere kanalen worden gestimuleerd. Veel CI-onderzoekers hebben geen toegang tot, of ervaring met onderzoeksinterfaces die de stimulatieparameters direct kunnen controleren. CI-onderzoeksinterfaces zouden, in tegenstelling tot klinische processoren, moeten worden gebruikt om de psychofysische verschillen van CI-gebruikers te meten, omdat juist in de klinische processoren de signalen verstoord worden door de in de klinische processor geïnstalleerde map, de frequentietoewijzing etc. Dit is met name relevant wanneer de perceptie van de temporele envelop wordt gemeten, omdat veranderingen in de diepte van de AM van een signaal waarschijnlijk het meest worden verstoord door de instellingen van de klinische processor. Het is tevens moeilijk om specifieke elektroden te selecteren met een klinische processor, omdat de bandbreedtes van het analysefilter kunnen resulteren in het stimuleren van meerdere elektroden, terwijl de stimulatie maar op één elektrode was gericht. Hoewel sommige CI-onderzoekers toegang hebben tot onderzoeksinterfaces zijn er maar enkelen die multi-kanaalpsychofysica bestuderen. Dit heeft geresulteerd in een discrepantie tussen uitkomsten van veel enkel-kanaalsmetingen (bijvoorbeeld het ordenen van toonhoogtes per elektrode, het discrimineren van stimuli of detectie van drempels per elektrode) en het daadwerkelijke spraakverstaan bij gebruik van meerdere kanalen. Een deel van deze verschillen kan worden uitgelegd op basis van top-downprocessen die het verstaan van spraak in CI's robuust maken, ondanks verstoringen in het aangeboden signaal en/of psychofysische gebreken. We weten echter relatief weinig over psychofysische perceptie met drie of meer kanalen. Het is belangrijk om de psychofysische limieten van CI-gebruikers te weten te komen in zowel een enkel- als multikanaalcontext om de verwerking van het CI-signaal te verbeteren en te optimaliseren. Daarom zijn de hier gepresenteerde studies een eerste stap in de richting van het beter begrijpen van de perceptuele limieten van CI-gebruikers in een brede context.

De studies van dit proefschrift laten zien dat luidheidssummatie bij gebruik van meerdere kanalen van invloed kan zijn wanneer de temporele enveloppe-perceptie wordt geëxtrapoleerd van een meting aan een enkel kanaal naar meerdere kanalen. De luidheidssummatie verschilt sterk tussen CI-gebruikers, maar de oorzaak van deze verschillen is onbekend. De effecten van luidheidssummatie op AM-detectie en -frequentie kunnen verschillen tussen patiënten. Omdat de waarneming van de temporele enveloppe van een enkel kanaal met de spraakperceptie bij het gebruik van meerdere kanalen correleert en, omdat bij het afregelen van een CI de luidheid moet worden aangepast op het waarnemen van geluid met meerdere kanalen, moeten psychofysische studies controleren op luidheidssummatie bij stimulatie van meerdere kanalen om het gat tussen de simpele psychofysica en het spraakverstaan met een CI in real life te dichten. De studies van dit proefschrift laten zien dat ieder kanaal een verschillende bijdrage kan leveren aan de waarneming met meerdere kanalen, iets wat onder andere afhankelijk is van de aangeboden taak. Bijvoorbeeld, de verschillen per plaats van stimulatie beïnvloeden de perceptie van meerdere kanalen in Hoofdstuk 5 (waarbij verschillende AM-frequenties worden aangeboden aan elk kanaal), maar niet in Hoofdstuk 3 en 4 (waarbij een samenhangende AM werd aangeboden aan meerdere kanalen).

Er is meer onderzoek nodig om de effecten van luidheidssummatie op de perceptie bij stimulatie van meerdere kanalen te begrijpen. Meerdere aspecten van een CI kunnen de luidheid beïnvloeden. Het toevoegen van meerdere kanalen zal de luidheid versterken. Compenseren voor de versterking van luidheid door het gebruik van meerdere kanalen zal de hoeveelheid stroom naar elk apart kanaal verminderen en kan, afhankelijk van de gebruikte meetmethode, misschien de perceptie van de temporele enveloppe verminderen. Het toevoegen van kanalen zal ook de interactie tussen kanalen versterken, wat de prestaties van de CI sterk vermindert. Het verhogen van het tempo van de stimulatie versterkt tevens de luidheid en de interactie tussen kanalen. Moderne CI-processoren creëren meer kanalen dan de CI functioneel heeft, waarbij ze ook sneller stimuleren dan nodig is om effectief spraak te verstaan te coderen. Er is balans nodig tussen het optimale aantal kanalen, de afstand tussen de kanalen, de hoeveelheid stimulatie voor elk kanaal, en de hoeveelheid stroom die moet worden geleverd aan elk kanaal. Omdat al deze factoren interacteren wanneer meerdere kanalen worden gebruikt, zal CI-onderzoek en -ontwikkeling de effecten van deze stimulatieparameters op een psychofysische manier moeten meten in een context waarin meerdere kanalen worden gebruikt. Dit is niet eenvoudig en het vereist gespecialiseerde hardware (onderzoeksinterfaces), complexe experimentele designs en een goede controle van de stimuli. De gepresenteerde studies zijn bij lange na niet eenduidig, maar laten zien dat er belangrijke stappen zijn genomen in het begrijpen van de perceptie van de temporele enveloppe bij het gebruik van meerdere kanalen.

Gezien de effecten van luidheidssummatie en de interactie tussen de kanalen, zijn er enkele klinische aanbevelingen die de perceptie van de temporele enveloppe bij gebruik van meerdere kanalen kan verbeteren. Luidheidssummatie kan resulteren in verminderde stroomtoevoer naar elk kanaal, wat weer zal zorgen voor een verminderde AM-detectie. De interactie tussen kanalen zal de temporele enveloppe interferentie versterken. Het is daarom wellicht wenselijk om minder elektrodes te stimuleren en de elektroden relatief ver uit elkaar te stimuleren. Dit reduceert de summatie en interactie. Om verdere interactie tussen de kanalen en luidheidssummatie tegen te gaan welke door de temporele enveloppe worden veroorzaakt bij het integreren van meerdere pulsen, zouden lagere stimulatie snelheden moeten worden gebruikt. Eerdere studies hebben laten zien dat meer stimulatie de perceptie van spraak niet verbetert en dat het de AM-detectie van een enkel kanaal belemmert. Lagere stimulatie snelheden verhogen de hoeveelheid stroom en verlagen de interactie tussen kanalen, wat samen de perceptie van de temporele enveloppe verbetert. Uiteindelijk zullen de spraakprocessoren het signaal op een dergelijke manier moeten bewerken dat de spraakinformatie wordt doorgegeven. Belangrijke frequentie-gerelateerde spraakeigenschappen (zoals de begrijpelijkheid en articulatie) kunnen nuttig zijn in het vaststellen van de belangrijkste kanalen die moeten worden geselecteerd binnen een stimulatiekader. Tegenwoorden stimuleren klinische spraakprocessoren alle kanalen binnen een stimulatiekader (varianten van de CIS-strategy voor Advanced Bionics en Med-El-apparaten) of een subset van de kanalen met het meeste vermogen (ACE-strategie in Cochlear Corp. apparaten). Doordat de stroom zich verspreid (of de elektrische reactie van elk kanaal zich verspreid), kunnen zulke strategieën de nadelige effecten van interactie tussen kanalen niet vermijden. Door het relatieve belang van zowel de frequentie-informatie (zoals formantinformatie) als de mogelijkheid van kanaalinteractie in het achterhoofd te houden, kunnen elektroden beter worden geselecteerd om zowel de spectrale als temporele enveloppe-informatie te behouden. Slimme selectie van minder kanalen binnen een bepaalde instelling zou de luidheidssummatie en kanaalinteractie verminderen en hopelijk de functionele output voor CI-gebruikers verder verbeteren dan de tot nu toe aangehouden harde limiet van acht kanalen.

Chapter 10: Curriculum vitae

Name: John J. Galvin III
Education: Hampshire College, Amherst MA (BA: Comparative literature, music), 1990

Statement

I have been a cochlear implant (CI) researcher for nearly twenty years. My background (comparative literature, music), though not typical for this field, has served me well in that I have authored and co-authored many CI research papers regarding CIs, with an emphasis on music perception. I have benefitted greatly from training and collaborative opportunities while working at the House Ear Institute, and have in turn trained many students and visiting researchers. Besides my interest in improving CI users' music perception, I have long been actively involved in improving auditory rehabilitation for CI users. I have also extensively studied speech perception in CI users. At House Ear Institute, I contributed to the drafting of many funded NIH grants and contracts, as well as collected and analyzed the data from these studies. At UCLA, I have been an integral part of drafting, submitting, and now collecting and analyzing data for an FDA IDE trial regarding cochlear implants and single-sided deafness. The present proposal builds on previous speech, music, and auditory training experience to explore cochlear implant users' perception of "atypical" speech (e.g., emotional speech, sung speech, speech from non-native talkers, speech from hearing-impaired talkers, etc.), which may better represent speech encountered outside the clinic. My experience, record of productivity, and research interests uniquely position me to purse the proposed studies.

Positions and employment

1994-1995: Research Technician, Dept. of Auditory Implants and Perception, House Ear Institute, CA

1996-1998: Research Assistant, Dept. of Auditory Implants and Perception, House Ear Institute, CA

1998-2001: Research Associate, Dept. of Auditory Implants and Perception, House Ear Institute, CA
2001-2007: Advanced Research Associate, Dept. of Auditory Implants and Perception, House Ear Institute

2007-2011: Senior Research Associate, Dept. of Auditory Implants and Perception, House Research Institute

2013-present: Senior Systems Analyst, Department of Head and Neck Surgery, David Geffen School of Medicine, University of California, Los Angeles, CA

Other Experience and Professional Memberships

2008-2011: Publications Committee for the Association for Research in Otolaryngology

2002-present: Member of the Association for Research in Otolaryngology

2009-present: Coordinator for sound and visual services for the bi-annual Conference on Implants and Auditory Prostheses

Contribution to Science

Music perception in cochlear implant users

Under ideal listening conditions, cochlear implants (CIs) provide many deaf people with good speech understanding. However, as listening conditions become more challenging (e.g., noise, reverberation, unfamiliar talkers, etc.), CI users typically have great difficulty due to the coarse spectral resolution provided by the device. Music perception and appreciation is perhaps one of the most challenging conditions for CI users, primarily due to poor pitch perception. The CI does not provide the fine-structure cues needed for harmonic pitch, and CI users must rely on coarse spectral envelope information and temporal envelope cues to perceive pitch. Historically, CI users' pitch perception had been measured using simple pitch discrimination, familiar melody recognition, and subjective evaluations, none of which provided very good quantification of CI users' pitch resolution or their ability to use pitch information. To address this issue, we developed a melodic contour identification (MCI) task, which measured listeners' ability to track changes in pitch using simple 5-note melodies. The MCI has proved to be very useful in our research group and others, and has been modified to include different instruments, pitch ranges, spacing between notes, etc. Our most recent version uses "sung speech," in which words are used as "instruments." With sung speech, we can measure sentence recognition in the presence of pitch changes, and music perception in the presence of timbre changes, bringing perceptual measures to be more in line with "real-world" listening conditions. The MCI task has also been used to successfully train CI users' music perception; short training exercises performed at home on personal computers was shown to improve CI users" MCI and familiar melody recognition performance.

Auditory training in cochlear implant users

After receiving their implant, post-lingually deafened CI users must adapt to both the new mode of electrical stimulation and to the distortions to speech patterns learned during previous acoustic hearing experience. Much of this adaption will occur "automatically" during the first 6-12 months of experience with the CI. However, this adaption is often incomplete. Experienced CI users also have to adapt to changes in CI signal processing (e.g., stimulation rate, frequency allocation, etc.). Active training, rather than passive learning, may be necessary to maximize patient performance. However, auditory rehabilitation resources are few and far between for adult CI users. From a research perspective, it is difficult to characterize patient performance if deficits are due to incomplete adaptation. To address these issues, we developed a training platform that would automatically adapt to individual CI users' performance levels. We created a large multi-talker database of monosyllable words that could be used to target contrasts specific to patients' needs. We expanded the training stimuli and exercises to include recognition of common words, speech recognition in noise, and music perception. The custom software designed by Qian-jie Fu also allowed patients to train at home at their convenience on their home computers, making it very time- and cost-effective. Of course, none of this would matter if the training didn't work. However, the many studies in our group have shown the training to be very successful. Mean improvements were often 10-15 percentage points, with some subjects improving by 40 points or more in some tasks. Our research also showed that the training improved auditory perception, rather than task-specific or procedural learning, and that the improvements often generalized to untrained tasks and were retained log after the training had stopped. The training software has been licensed by two major implant manufacturers and has been developed for different languages. This, of course, is the ultimate goal of research – to benefit people in need as effectively and quickly as possible. We have long realized that auditory training is key to both understanding the limits of CI performance and to maximizing the benefit of implantation.

Perception of temporal modulation in cochlear implant users

Most CIs work by using the temporal envelopes extracted from frequency analysis bands to modulate pulse trains of current delivered to appropriate intra-cochlear electrodes. Perception of temporal envelope information is critical, especially when the spectral resolution is degraded, as in the CI case. Temporal modulation detection and modulation frequency discrimination are psychophysical measures that be used to characterize CI users' perception of temporal envelopes.

Perception of temporal modulation has also been correlated with CI users' speech performance. We have extensively studied single- and multi-channel perception of temporal modulation using custom research interfaces, as perception via CI users' clinical processors is limited by parameter settings that may or may not relate to patients' psychophysical limits. We found that, for equally loud stimuli, single-channel modulation detection worsens with high carrier stimulation rates, most likely due to the lower current levels needed to accommodate multi-pulse summation. Similarly, we found that, for equally loud stimuli, modulation detection is poorer with multiple than with single channels, again due to the lower current levels needed to accommodate multi-channel loudness summation. Interestingly, we found that modulation frequency discrimination was similar for equally loud single- and multi-channel stimuli. Overall, our research has pointed out the need to consider the effects of loudness summation and amplitude mapping when optimizing CI signal processing to preserve temporal envelope information.

Perception of indexical cues and other "atypical" speech measures by cochlear implant users

Most CI speech research has focused on perception of lexical content. However, prosodic and indexical cues contribute greatly to overall communication, and depend strongly on fundamental frequency (F0) cues which are not well represented by CIs. For tonal languages such as Mandarin Chinese, F0 cues are lexically meaningful. We have researched many aspects of speech perception beyond understanding of lexical information. We have also collaborated with many Chinese researchers to better understand the limits of tonal language perception with CIs. We found that voice gender identification depends strongly on temporal envelope perception when the spectral resolution is reduced. We found that CI users' vocal emotion recognition is poorer than for normal-hearing listeners, due poor access to F0 information. We found that Mandarin tone recognition improves greatly when a hearing aid is used in combination with a CI. We found that CI users are sensitive to differences between synthetic and natural speech, and are susceptible to fast speaking rates. We found that musical experience and training can improve Chinese CI users' speech perception. We found that native English-speaking CI users' understanding of English produced by non-native talkers is poorer than when produced by native talkers. Taken together, our research indicates there is much to understand and much to improve in CI users' perception of prosodic and indexical cues.

Chapter 11: Publications

Peer-reviewed publications:

Crew JD, **Galvin JJ 3rd,** Fu QJ. Perception of sung speech in bimodal cochlear implant users. Trends Hear. 2016 [in press]

Li Y, Wang S, Su Q, **Galvin JJ,** Fu QJ. Validation of list equivalency for Mandarin speech materials to use with cochlear implant listeners. Int J Audiol. 2016 Jul 14:1-10.

Zhou X, Li H, **Galvin JJ 3rd,** Fu QJ, Yuan W. Effects of insertion depth on spatial speech perception in noise for simulations of cochlear implants and single-sided deafness. Int J Audiol. 2016 Jul 1:1-8

Su Q, **Galvin JJ**, Zhang G, Li Y, Fu QJ. Effects of Within-Talker Variability on Speech Intelligibility in Mandarin-Speaking Adult and Pediatric Cochlear Implant Patients. Trends Hear. 2016 Jun 30;20.

Fang Y, Wang B, **Galvin JJ 3rd,** Tao D, Deng R, Ou X, Liu Y, Dai P, Sha Y, Zhang T, Chen B. MPR-CT imaging for stapes prosthesis: Accuracy and clinical significance. Otol Neurotol. 2016 Apr;37(4):321-23.

Galvin JJ 3rd, Oba SI, Başkent D, Chatterjee M, Fu QJ. Envelope interactions in multi-channel amplitude modulation frequency discrimination by cochlear implant users. PLoS One. 2015 Oct 2;10(10):e0139546.

Zhou X, Yuan W, **Galvin JJ,** Fu QJ, Zhang Y. Influence of language experience on digit recognition by English and Chinese listeners. J Acoust Soc Am. 2015 Sep;138(3):EL324-8.

Bakhos D, Vitaux H, Villeneuve A, Kim S, Lescanne E, Pigeon V, Aoustin JM, Bordure P, **Galvin J.** The effect of the transducers on paediatric thresholds estimated with auditory steady-state responses. Eur Arch Otorhinolaryngol. 2016 Aug;273(8):2019-26.

Crew JD, **Galvin JJ 3rd,** Fu QJ. Melodic contour identification and sentence recognition using sung speech. J Acoust Soc Am. 2015 Sep;138(3):EL347.

Galvin JJ 3rd, Oba S, Başkent D, Fu QJ. Modulation frequency discrimination with single and multiple channels in cochlear implant users. Hear Res. 2015 Jun;324:7-18.

Fu QJ, **Galvin JJ 3rd,** Wang X, Wu JL. Benefits of music training in mandarin-speaking pediatric cochlear implant users. J Speech Lang Hear Res. 2015 Feb 1;58(1):163-9.

Tao D, Deng R, Jiang Y, **Galvin JJ 3rd,** Fu QJ, Chen B. Melodic pitch perception and lexical tone perception in Mandarin-speaking cochlear implant users. Ear Hear. 2015 Jan;36(1):102-10.

Galvin JJ 3rd, Oba SI, Başkent D, Chatterjee M, Fu QJ. Envelope Interactions in Multi-Channel Amplitude Modulation Frequency Discrimination by Cochlear Implant Users. PLoS One. 2015 Oct 2;10(10):e0139546.

Crew JD, **Galvin JJ 3rd,** Landsberger DM, Fu QJ. Contributions of electric and acoustic hearing to bimodal speech and music perception.
PLoS One. 2015 Mar 19;10(3):e0120279.

Fuller CD, Gaudrain E, Clarke JN, **Galvin JJ,** Fu QJ, Free RH, Başkent D. Gender categorization is abnormal in cochlear implant users. J Assoc Res Otolaryngol. 2014 Dec;15(6):1037-48.

Ji C, **Galvin JJ,** Chang YP, Xu A, Fu QJ. Perception of speech produced by native and nonnative talkers by listeners with normal hearing and listeners with cochlear implants. J Speech Lang Hear Res. 2014 Apr 1;57(2):532-54.

Fuller CD, **Galvin JJ 3rd**, Free RH, Başkent D. Musician effect in cochlear implant simulated gender categorization. J Acoust Soc Am. 2014 Mar;135(3):EL159-65.

Galvin JJ 3rd, Fu QJ, Oba S, Başkent D. A method to dynamically control unwanted loudness cues when measuring amplitude modulation detection in cochlear implant users. J Neurosci Methods. 2014 Jan 30;222:207-12.

Bakhos D, Delage H, **Galvin J,** Lescanne E, Roux S, Bonnet-Brilhault F, Bruneau N. Cortical electrophysiological markers of language abilities in children with hearing aids: a pilot study. Biomed Res Int. 2014;2014:198153.

Fuller CD, **Galvin JJ 3rd,** Maat B, Free RH, Başkent D. The musician effect: does it persist under degraded pitch conditions of cochlear implant simulations? Front Neurosci. 2014 Jun 30;8:179.

Li Y, Zhang G, **Galvin JJ 3rd,** Fu QJ. Mandarin speech perception in combined electric and acoustic stimulation. PLoS One. 2014 Nov 11;9(11):e112471.

Tao D, Deng R, Jiang Y, **Galvin JJ 3rd,** Fu QJ, Chen B. Contribution of auditory working memory to speech understanding in mandarin-speaking cochlear implant users. PLoS One. 2014 Jun 12;9(6):e99096.

Galvin JJ 3rd, Oba S, Fu QJ, Başkent D. Single- and multi-channel modulation detection in cochlear implant users. PLoS One. 2014 Jun 11;9(6):e99338.

Ji C, **Galvin JJ 3rd,** Xu A, Fu QJ. Effect of speaking rate on recognition of synthetic and natural speech by normal-hearing and cochlear implant listeners. Ear Hear. 2013 May-Jun;34(3):313-23.

Wilkinson EP, Abdel-Hamid O, **Galvin JJ 3rd,** Jiang H, Fu QJ. Voice conversion in cochlear implantation. Laryngoscope. 2013 Apr;123 Suppl 3:S29-43.

Oba SI, **Galvin JJ 3rd,** Fu QJ. Minimal effects of visual memory training on auditory performance of adult cochlear implant users. J Rehabil Res Dev. 2013;50(1):99-110.

Crew JD, **Galvin JJ 3rd,** Fu QJ. Channel interaction limits melodic pitch perception in simulated cochlear implants. J Acoust Soc Am. 2012 Nov;132(5):EL429-35.

Galvin JJ III, Eskridge E, Oba S, Fu QJ. Melodic contour identification training in cochlear implant users with and without a competing instrument. Semin Hear. 2012; 33(04): 399-409.

Eskridge EN, **Galvin JJ 3rd,** Aronoff JM, Li T, Fu QJ. Speech perception with music maskers by cochlear implant users and normal-hearing listeners. J Speech Lang Hear Res. 2012 Jun;55(3):800-10.

Zhu M, Chen B, **Galvin JJ 3rd,** Fu QJ. Influence of pitch, timbre and timing cues on melodic contour identification with a competing masker (L). J Acoust Soc Am. 2011 Dec;130(6):3562-5.

Oba SI, Fu QJ, **Galvin JJ 3rd.** Digit training in noise can improve cochlear implant users' speech understanding in noise. Ear Hear. 2011 Sep-Oct;32(5):573-81.

Landsberger D, **Galvin JJ 3rd.** Discrimination between sequential and simultaneous virtual channels with electrical hearing. J Acoust Soc Am. 2011 Sep;130(3):1559-66.

Zhu M, Fu QJ, **Galvin JJ 3rd,** Jiang Y, Xu J, Xu C, Tao D, Chen B. Mandarin Chinese speech recognition by pediatric cochlear implant users. Int J Pediatr Otorhinolaryngol. 2011 Jun;75(6):793-800.

Galvin JJ 3rd, Fu QJ. Effect of bandpass filtering on melodic contour identification by cochlear implant users. J Acoust Soc Am. 2011 Feb;129(2):EL39-44.

Shannon RV, Cruz RJ, **Galvin JJ 3rd.** Effect of stimulation rate on cochlear implant users' phoneme, word and sentence recognition in quiet and in noise. Audiol Neurootol. 2011;16(2):113-23.

Sagi E, Fu QJ, **Galvin JJ 3rd,** Svirsky MA. A model of incomplete adaptation to a severely shifted frequency-to-electrode mapping by cochlear implant users. J Assoc Res Otolaryngol. 2010 Mar;11(1):69-78.

Luo X, **Galvin JJ,** Fu QJ. Effects of stimulus duration on amplitude modulation processing with cochlear implants. J Acoust Soc Am. 2010 Feb;127(2):EL23-9.

Galvin JJ 3rd, Fu QJ, Shannon RV. Melodic contour identification and music perception by cochlear implant users. Ann N Y Acad Sci. 2009 Jul;1169:518-33.

Li T, **Galvin JJ 3rd,** Fu QJ. Interactions between unsupervised learning and the degree of spectral mismatch on short-term perceptual adaptation to spectrally shifted speech. Ear Hear. 2009 Apr;30(2):238-49.

Galvin JJ 3rd, Fu QJ. Influence of stimulation rate and loudness growth on modulation detection and intensity discrimination in cochlear implant users. Hear Res. 2009 Apr;250(1-2):46-54.

Galvin JJ 3rd, Fu QJ, Oba SI. Effect of a competeing instrument on melodic contour identification by cochlear implant users. J Acoust Soc Am. 2009 Mar;125(3):EL98-103.

Galvin JJ 3rd, Fu QJ, Oba S. Effect of instrument timbre on melodic contour identification by cochlear implant users. J Acoust Soc Am. 2008 Oct;124(4):EL189-95.

Fu QJ, **Galvin JJ 3rd.** Maximizing cochlear implant patients' performance with advanced speech training procedures. Hear Res. 2008 Aug;242(1-2):198-208.

Liu C, **Galvin J 3rd,** Fu QJ, Narayanan SS. Effect of spectral normalization on different talker speech recognition by cochlear implant users. J Acoust Soc Am. 2008 May;123(5):2836-47.

Xin Luo, Fu QJ, **Galvin JJ 3rd.** Vocal emotion recognition by normal-hearing listeners and cochlear implant users. Trends Amplif. 2007 Dec;11(4):301-15. Trends Amplif. 2007 Sep;11(3):e1.

Fu QJ, **Galvin JJ 3rd.** Perceptual learning and auditory training in cochlear implant recipients. Trends Amplif. 2007 Sep;11(3):193-205.

Galvin JJ 3rd, Fu QJ, Nogaki G. Melodic contour identification by cochlear implant listeners. Ear Hear. 2007 Jun;28(3):302-19.

Fu QJ, **Galvin JJ 3rd.** Computer-Assisted Speech Training for Cochlear Implant Patients: Feasibility, Outcomes, and Future Directions. Semin Hear. 2007 May 1;28(2).

Nogaki G, Fu QJ, **Galvin JJ 3rd.** Effect of training rate on recognition of spectrally shifted speech. Ear Hear. 2007 Apr;28(2):132-40.

Fu QJ, **Galvin JJ 3rd.** Recognition of simulated telephone speech by cochlear implant users. Am J Audiol. 2006 Dec;15(2):127-32.

Chatterjee M, **Galvin JJ 3rd,** Fu QJ, Shannon RV. Effects of stimulation mode, level and location on forward-masked excitation patterns in cochlear implant patients. J Assoc Res Otolaryngol. 2006 Mar;7(1):15-25.

Fu QJ, Chinchilla S, Nogaki G, **Galvin JJ 3rd.** Voice gender identification by cochlear implant users: the role of spectral and temporal resolution. J Acoust Soc Am. 2005 Sep;118(3 Pt 1):1711-8.

Galvin JJ 3rd, Fu QJ. Effects of stimulation rate, mode and level on modulation detection by cochlear implant users. J Assoc Res Otolaryngol. 2005 Sep;6(3):269-79.

Fu QJ, Nogaki G, **Galvin JJ 3rd.** Auditory training with spectrally shifted speech: implications for cochlear implant patient auditory rehabilitation. J Assoc Res Otolaryngol. 2005 Jun;6(2):180-9.

Fu QJ, **Galvin J,** Wang X, Nogaki G. Effects of auditory training on adult cochlear implant patients: a preliminary report. Cochlear Implants Int. 2004 Sep;5 Suppl 1:84-90.

Fu QJ, Chinchilla S, **Galvin JJ.** The role of spectral and temporal cues in voice gender discrimination by normal-hearing listeners and cochlear implant users. J Assoc Res Otolaryngol. 2004 Sep;5(3):253-60.

Shannon RV, Fu QJ, **Galvin J 3rd.** The number of spectral channels required for speech recognition depends on the difficulty of the listening situation. Acta Otolaryngol Suppl. 2004 May;(552):50-4.

Fu QJ, Shannon RV, **Galvin JJ 3rd**. Perceptual learning following changes in the frequency-to-electrode assignment with the Nucleus-22 cochlear implant. J Acoust Soc Am. 2002 Oct;112(4):1664-74.

Shannon RV, **Galvin JJ 3rd,** Baskent D. Holes in hearing. J Assoc Res Otolaryngol. 2002 Jun;3(2):185-99.

Zeng FG, Grant G, Niparko J, **Galvin J,** Shannon R, Opie J, Segel P. Speech dynamic range and its effect on cochlear implant performance. J Acoust Soc Am. 2002 Jan;111(1 Pt 1):377-86.

Fu QJ, **Galvin JJ 3rd.** Recognition of spectrally asynchronous speech by normal-hearing listeners and Nucleus-22 cochlear implant users. J Acoust Soc Am. 2001 Mar;109(3):1166-72.

Fu QJ, **Galvin JJ 3rd,** Wang X. Recognition of time-distorted sentences by normal-hearing and cochlear-implant listeners. J Acoust Soc Am. 2001 Jan;109(1):379-84.

Zeng FG, **Galvin JJ 3rd.** Amplitude mapping and phoneme recognition in cochlear implant listeners. Ear Hear. 1999 Feb;20(1):60-74.

Zeng FG, **Galvin JJ 3rd,** Zhang C. Encoding loudness by electric stimulation of the auditory nerve. Neuroreport. 1998 Jun 1;9(8):1845-8.

Other publications:

Shannon RV, Fu QJ, **Galvin J,** Friesen L. Speech perception with cochlear implants. In *Cochlear Implants: Auditory Prostheses and Electric Hearing*. Springer Handbook of Auditory Research. Springer New York. 2004;Vol 20:334-76.

Fu QJ, **Galvin J III.** Auditory training for cochlear implant patients. In Auditory Prostheses. . Springer Handbook of Auditory Research. Springer New York. 2011; Vol 39: 257-78

Chapter 12: Gratitude

There are so many people to thank...

Deniz: so lovely and fitting that we've made a circle of friendship that now includes my PhD.

Qian-jie: my friend for more than 20 years, thanks for all you have shown and shared with me.

Astrid, Bart, and Christian: thanks for your careful and kind reading of my work.

Monita: you are truly my sister from another mister.

Bob: father, grandfather, and friend to all, and to me from Day 1.

Christina: thanks for all your help along the way, especially translating all of this into Dutch (also, thanks to Rolien).

Special thanks to Prof. Dr. Eelco Huizinga and the Huizinga Foundation for support of this thesis.

Friends, family and colleagues: Alena, Bob, Monita, Manager, David L, Joe, Qian-jie, Xiao-song, Monica, Tim, Arthi, Lendra, Xin, Sandy, Bomjun, Fan-gang, Chao, Johan, Tanya, David O'G, Elizabeth, Dana, Geri, Sherol, Chuping, Yiping, Tian-hao, Justin, Mark, Eric W, Eric L, Dawna, Suzzanah, Jian, Duoduo, Tassos, Etienne, Ria, Davide, Mathieu, Jean-Marie, Andy F, Andy O, Chris, Josh, Zach, Oliver, Dan, Stuart, Debbie, Bas, Ning, Danny, Dan, Tim A, Tim W, Jon, Giddle, Jack, John, Earl, Ron, Carl, Dianne, Matt F, Sanjay, Ami, Daniel, Maggie, Barbara, Bette-Ann, Christian, Frank, Armen, Sean, Neal, Jack, Maria, Matthew, Leah, Nadia, Lucille, Eva, Jeff, Nancy, Dean, Jon, Kristin, Aiden

And never last, never least: Lisa, you've been with me longer than without, in all the ways that one can be with another. Thank you for sharing, caring, and living with my little life, made larger by the lens of your love.