CORTICAL RESPONSES TO SPEECH AND COMPLEX TONAL STIMULI IN ADULTS WITH NORMAL HEARING AND SENSORINEURAL HEARING LOSS

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A THESIS SUBMITTED IN FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE OF DOCTOR OF PHILOSOPHY IN SPEECH SCIENCE

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Abstract

Aims: Aims of this thesis were to: 1) investigate speech stimuli and background-noise-dependent changes in cortical event related potentials (ERPs) in unaided and aided conditions, and determine amplification effects on ERPs, 2) examine behavioural and neural processing of pitch cues in adults with normal hearing (NH) and adults with sensorineural hearing loss (SNHL), and 3) investigate the effects of auditory training on pitch processing using behavioural and electrophysiological approaches in adults with SNHL.

Method: In Study 1 P1, N1, and P2 responses to naturally produced syllables in quiet and in multi-talker babble were recorded, with and without a hearing aid in the right ear. Acoustic characteristics of the hearing-aid-transduced stimuli were measured using in-the-canal probe microphone measurements. In Study 2 behavioural pitch discrimination abilities were tested using the monaural TFS1 test (Moore & Sek, 2009a). Cortical potentials (N1, P2 and acoustic change complex, ACC) were recorded in response to frequency shifted (deltaF) tone complexes in an 'ABA' pattern in adults with mild and high frequency SNHL. In Study 3 N1, P2 and P3 ERPs and their related behavioural measures of discrimination (d-prime sensitivity and reaction time) were recorded using an active oddball paradigm. Behavioural pitch discrimination abilities were tested using the monaural (right ear) TFS1 (Temporal Fine Structure 1) test. All tests were conducted during pre-training and post-training sessions. Training consisted of discrimination of complex tones varying in pitch using custom software (Vandali et al., 2015)

Results: The first study revealed that CAEP latencies and amplitudes showed significant effects of speech contrast, background noise and amplification. N1 and P2 components varied differently across conditions. Hearing-aid induced spectral and temporal changes to the speech stimuli affected P1-N1-P2 components. The second study showed that the SNHL group performed more poorly than the NH group for the TFS1 test and hence had poorer discrimination of fine structure cues, despite having normal or mild hearing loss in the frequency region of the stimulus. P2 (latency and amplitude) was more reflective of pitch differences between the complexes than N1. The presence of the acoustic change complex in response to the TFS transitions in the ABA stimulus varied with deltaF (and hence with pitch salience). Acoustic change complex amplitudes were reduced for the group with SNHL compared to controls. The third study demonstrated stimulus-specific ERP changes after training with no significant improvement in behavioural discrimination performance. In Study 3 P2 amplitude was more sensitive to training mastery (progress on the auditory training task) than behavioural discrimination abilities.

Conclusion: Cortical ERPs reflect spectral and temporal characteristics of speech and complex-tonal stimuli and changes induced by background noise, amplification and training.

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To the life experiences that taught me about decision-making and, my family & friends who handles the consequences.

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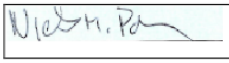
Niels H. Pontoppidan		23/06/2015
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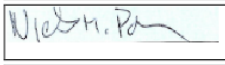
Niels H. Pontoppidan		23/06/2015
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Chapter 1

Introduction

Sensorineural hearing loss (SNHL) is associated with deficits in signal 'audibility' (elevated thresholds for sound detection) and 'distortions' that exist even when audibility is achieved (Plomp, 1986). Audibility depends on the signal presentation level relative to hearing thresholds or background noise levels whereas supra-threshold deficits due to distortion can be present even when the signal level ensures audibility. Distortion can occur as a result of the interaction between the signal and the noise. Signal audibility in listeners with SNHL is easily estimated or verified clinically using an audiogram and tools such as speech mapping (Moore, 2006) that are used to determine if a hearing aid effectively amplifies the speech spectrum. These measures are recognised as imprecise predictors of speech in noise abilities, however (Ching, Dillon, & Byrne, 1998; Smoorenburg, 1992). Speech understanding in the elderly in particular is poorly predicted by measures of signal audibility alone (Humes, 1996). The pure tone audiogram alone is not an ideal method for prediction of the effects of a hearing loss especially when the impact on hearing thresholds is minimal. For all severities of SNHL, the impact is most noticeable when listening to speech in adverse listening conditions (Killion, 1997). Recently, there has been increased attention directed towards early identification of 'subclinical' or 'hidden' hearing loss as a result of aging and noise exposure that has minimal effects on the audiogram but is associated with listening difficulties (Plack, Barker, & Prendergast, 2014; Stone & Moore, 2014). In listeners with minimal hearing

loss, audibility is intact and hence it is assumed that suprathreshold distortion in the peripheral and central auditory system accounts for these listening difficulties.

The 'distortional' component described by Plomp (1986) has been attributed to generalised deficits in the ability to use spectral and temporal cues associated with SNHL. Spectral cues in speech are important for consonant identification based on the frequency characteristics of onset release burst of voiced stop consonants and frication noise of obstruents (Dorman, Studdert-Kennedy, & Raphael, 1977; LaRiviere, Winitz, & Herriman, 1975). Spectral distortions associated with SNHL can arise from reduced frequency selectivity as evidenced by increased auditory filter bandwidths estimated using notched-noise masking (Glasberg & Moore, 1986). This results in spectral smearing of cues that relay consonant and vowel information in speech. Consistent with this are experiments showing: 1) reduced consonant recognition when frequency content of the consonants falls into a region of increased bandwidth (Dubno, Dirks, & Ellison, 1989), and 2) reduced speech intelligibility when listeners with normal hearing identified spectrally smeared signals that simulates the effects broadened auditory filters (Moore & Baer, 1993).

Temporal cues in speech correspond to the slow variations of amplitude over time, temporal envelope (ENV) that is imposed over a rapidly varying temporal fine structure (TFS) (Rosen, 1992). In a normal auditory system, broadband signals like speech are essentially decomposed into narrowband signals by filtering (depending on the centre frequency and bandwidth of the channel) that takes place within the cochlea. For each filter centre frequency, the cochlear/neural signal represents a sinusoid that is varying in peak amplitude and frequency from cycle to cycle. This can represent a slowly varying

envelope imposed on a more rapid TFS. Information about speech sounds is carried by both ENV and TFS cues (Moore, 2014). Rosen (1992) showed that ENV information mainly conveys manner of articulation cues derived from stimulus rise time and duration and TFS information conveys place of articulation cues derived from rapid frequency modulations such as F2 transitions. Previous studies have also shown that while ENV cues are sufficient for speech perception in quiet, TFS cues primarily convey pitch cues which enhances discrimination of the speech signal from background noise (Lorenzi et al., 2006; Smith et al., 2002). There is strong evidence for individuals with SNHL having limited ability to use TFS information to understand masked speech (Lorenzi et al., 2006) and to perform pitch discrimination tasks (Hopkins & Moore, 2007). Thus, intact processing of spectral and temporal cues is required for accurate perception of speech in quiet and in noise. Bernstein et al. (2013) showed a strong relationship between spectral and temporal modulation (STM) sensitivity and speech intelligibility in older adults with high frequency SNHL. They used a two-alternative forced choice adaptive method whereby listeners detected changes in modulations of broadband signals that varied in spectral density and temporal rate. Variation in STM sensitivity across participants was attributed to the combined variation in frequency selectivity at 4 kHz (notched-noise method) and TFS processing at 500 Hz (frequency modulation detection) and this was unrelated to their audiometric thresholds.

Signal to noise ratios (SNRs) required to achieve 50% correct speech recognition depend on the speech material and listening environment (Neuman et al., 2010), but there are consistent differences in performance between people with normal hearing and those with SNHL (Killon et al., 2004). Compared to listeners with normal hearing thresholds, people with SNHL require higher SNRs to achieve the same speech perception scores when

speech is presented in competing noise. For example, Phatak et al. (2009) reported that a subject with high frequency SNHL required a higher SNR of 14 dB, to achieve the same 50% correct consonant recognition scores compared to subjects with normal hearing (whose SNR was -16 dB on average). While the consequences of SNHL for speech perception in noise are well established, overcoming this problem for people with SNHL remains a challenge for aural rehabilitation.

Hearing aids are designed to restore speech to an audible level and are the usual recommended rehabilitation option for individuals with SNHL. Advances in hearing aid signal processing have addressed issues such as poor audibility and susceptibility to background noise in individuals with SNHL. Improved audibility has been addressed using multiple frequency bands to allow flexibility in gain provision to match the audiometric profile, multi-channel compression in many frequency channels to allow amplification of soft sounds and feedback cancellation to allow more gain without feedback (Dillon, 2001). The effectiveness of these technological developments for enhancing signal audibility has been established (Pittman, Pederson, & Rash, 2014). The problem of poor speech perception in noise has been addressed by providing adaptive directional microphones to pick up sounds from a specific direction/target in the listening field and binaural beam forming that further improves directionality and listening in noisy conditions (Kreikemeier et al., 2013). There is strong evidence for enhanced speech perception in noise using directional microphone technologies (Bentler, Palmer, & Mueller, 2006). Despite advances in technology, there still remain limitations in hearing aids. Moore (2013) listed some of the known limitations to include issues with imprecise gain and compression ratios achieved on real ears compared to manufacturer targets, limited hearing aid bandwidth and gain above 3 kHz and variability in compression

speed. Most research has explored hearing aid technological advances one at a time (e.g. directional versus omnidirectional microphones with other hearing aid settings fixed) but, when combined, different hearing aid features may affect speech quality and intelligibility. In addition, hearing aid digital signal processing can alter the intensity, frequency and timing of the signal in complex ways (Bor, Souza, & Wright, 2008; Jenstad, & Souza, 2007; Souza, Jenstad, & Boike, 2006) and the effects of these changes in the speech signal on speech processing for people with SNHL is not well understood.

Because of the limitations of hearing aids, and the relatively low uptake of hearing aids (Lin, 2011) there has been focus on auditory training strategies to partially rehabilitate sensory and cognitive processing skills in adults with SNHL (Humes, Burk, Strauser, & Kinney, 2009; Woods & Yund, 2007; Stecker et al., 2006; Sweetow & Henderson-Sabes, 2004). Auditory training is based on the brain's ability to change its structure and function with auditory stimulation, referred to as brain plasticity (Lövdén et al., 2010). Most of the recent developments in this area involve training exercises delivered using computer-based programmes that people engage with either at home or in the clinic (e.g. LACE, Sweetow et al., 2004; aTune, Vandali et al., 2014). Training exercises usually include speech or non-speech signals in a wide range of listening tasks. The ideal amount of training required per day to facilitate learning appears to depend on the complexity of the auditory stimulus and difficulty of the task (Watson, 1991). Auditory training studies generally measure outcomes using speech identification scores (e.g. nonsense syllable test, Dubno & Levitt, 1981) or psychophysical tasks such as F0DL (fundamental frequency difference limen) measures (Amitay et al., 2006). Many training studies have reported small changes in performance between baseline and follow-up sessions (e.g. Henderson-Sabes & Sweetow, 2007), however, these changes vary across participants

and the significance of these changes in terms of improved everyday auditory function is not yet established.

Regardless of the choice of rehabilitation (amplification, auditory training), difficulty listening in noise continues to be a concern for hearing aid users (Kochkin, 2007). Whilst hearing aid fitting is an important first step for restoring signal audibility, and engagement in repeated and varied listening tasks during auditory training may be beneficial, results continue to be variable. Jin and Nelson (2006) noted considerable variability in performance on speech-in-noise tasks among hearing aid users. Factors contributing to variable performance with hearing aid technology include differences in psychoacoustic abilities (Glasberg & Moore, 1989), degree of hearing loss and length of hearing aid experience (Hickson, Clutterbuck, & Khan, 2010), and also cognitive skills such as auditory working memory capacity (Pichora-Fuller & Singh, 2006). It is important to understand how hearing aids affect the acoustics of speech and how the auditory signal is processed and integrated along the central auditory pathway in order to make progress in more reliably improving rehabilitation outcomes for all listeners with SNHL.

There is evidence from animal and human studies for changes in the pattern of central auditory processing after acquired hearing loss (Harrison, Stanton, Ibrahim, Nagasawa, & Mount, 1993; Campbell & Sharma, 2013; Oates, Kurtzberg, & Stapells, 2002; Peelle, Troiani, Grossman, & Wingfield, 2011; Schwaber, Garraghty, & Kaas, 1993). Using magnetoencephalography (MEG), Dietrich et al. (2001) showed that tonotopic representation in the auditory cortex (frequency mapping) in adults with sudden hearing loss differs from adults with normal hearing. Peelle et al. (2011) monitored brain function

and structure in older adults with SNHL using functional magnetic resonance imaging (fMRI) and voxel-based morphometry (VBM) during a sentence comprehension task of grammatically complex sentences and correlated the findings to the audiometric thresholds of the participants. Participants with poorer hearing showed a reduction in neural activity for tasks requiring higher linguistic demands and had reduced cortical grey matter, indicating an association between peripheral hearing loss, speech processing and cortical function and structure. This link between brain activity/morphology and listening behaviour is consistent with other studies showing correlations between speech perception and auditory evoked brain activity (e.g. Kelly et al., 2005). Restoration of hearing using hearing aids or cochlear implants is associated with increased brain activation demonstrated by brain metabolic activity using positron emission tomography imaging studies (Naito et al., 2000).

Changes in auditory processing within the central auditory system can be determined objectively using electrophysiology (Tremblay et al., 2001; Tremblay & Kraus, 2002; Warrier et al., 2004). Electrophysiological recording equipment is widely available in audiology clinics and is inexpensive compared to neuroimaging techniques and hence electrophysiological measures should be considered as a clinical method for probing auditory processing and rehabilitation effectiveness in individuals with SNHL. Many studies have shown changes in electrophysiological responses as a result of improved processing of auditory signals. For example, Warrier et al. (2004) showed improved cortical representation (N2) of speech stimuli and corresponding improvement in speech perceptual scores in children with learning disability after training using the Earobics program.

Auditory evoked potentials (AEPs) measured using electroencephalography (EEG) or MEG result from electrical activity in the auditory system that is time-locked to an auditory stimulus (Stapells, 2002). In humans AEPs can be measured from the cochlea (cochlear microphonic, summating potential), the brainstem (auditory brainstem response), thalamo-cortical projections (middle latency response) and the auditory cortex (cortical auditory event related potentials (ERPs) (Hall, 2007). Research into ERPs is of particular interest for the current thesis as ERPs are affected by hearing loss (Oates et al., 2002), auditory training (Tremblay et al., 2001) and amplification (Tremblay et al., 2006), and correlations between ERP characteristics and speech perception have been demonstrated (Kelly et al., 2005).

ERPs reflect hierarchical levels of pre-attentive and attentive sound processing at the auditory cortex. ERPs have traditionally been used for objective hearing assessment in adults (Hyde, 1997) but could also be used clinically as a non-invasive objective measure of the adequacy of neural processing (Hyde, 1997; Martin et al., 2008; Munro, Purdy, Ahmen, Begum, & Dillon, 2011). The P1-N1-P2 complex, also known as cortical auditory evoked potentials (CAEPs), can be recorded using a simple obligatory passive listening paradigm. Other cortical potentials include the acoustic change complex (ACC), mismatch negativity (MMN) and P3, which are recorded using a discriminative listening paradigm in which the change in stimulus acoustic properties evokes a response (ACC, MMN) or the listener's active attention and response to the stimulus change evokes a response (P3) (Stapells, 2002).

The presence of a P1-N1-P2 complex in adults reflects detection of a stimulus (Davis, 1939) whereas the ACC indicates a response to change in an ongoing stimulus (Ostroff et al., 1998). ACCs contain multiple P1-N1-P2 complexes in response to sound onset, change and offset. These pre-attentive responses reflect stimulus acoustic properties, e.g. rise time, frequency transition, pitch (Crowley et al., 2004; Martin & Boothroyd, 2000; Onishi et al., 1968). Although these responses are passively recorded, they can be modulated by memory and attention in certain conditions (Hillyard et al., 1973; Ross et al., 2013). The MMN and P3 are negative and positive deflections elicited using a passive or an active oddball paradigm, respectively (Näätänen, Gaillard, & Mäntysalo, 1978; Sutton, Braren, Zubin, & John, 1965). These potentials are indices of stimulus discrimination influenced by sensory memory and attention (Näätänen, 1992; Overtoom et al., 1998).

CAEPs are generally described by their latency (timing information), amplitude (reflecting stimulus salience and strength of neural response) and topography of scalp distribution (Martin et al., 2008). These CAEP characteristics have been compared to behavioural auditory measures to establish whether CAEPs are sensitive to the stimulus and other experimental manipulations and to the effects of auditory pathology. Cortical auditory responses vary depending on the complexity of the evoking stimulus, e.g. speech vs. tones and the physical characteristics of the evoking stimulus, e.g. duration, rise time, level (Beukes et al., 2009; Purdy, Sharma, Munro & Morgan, 2013; Swink & Stuart, 2012). These responses have been used to show differences in encoding of complex acoustic cues in quiet and in noise (Agung et al., 2006; Kaplan-Neeman et al., 2006), as an outcome measure to validate benefits from hearing aids (Golding et al., 2007; Martinez, Eisenberg, & Boothroyd, 2013) and auditory training (Tremblay et al., 2001). A

number of studies show good correspondence between CAEP responses and behavioural performance (e.g. Anderson, Chandrasekaran, Yi, & Kraus, 2010; Chang, Dillon, Carter, Van Dun, & Young, 2012). Anderson et al. showed that CAEP N2 amplitude correlated with speech in noise performance of young children with normal hearing. Poor performers showed greater N2 amplitude change between quiet and noise conditions. Not all studies show this link between brain activity and behaviour, however. For example, CAEPs can be inconsistent with behavioural discrimination abilities (e.g. Kraus et al.. 1993), and in auditory training studies CAEPs can show change in the absence of behavioural changes (Tremblay et al., 1998). Thus, although there is interest in using evoked potentials to probe auditory discrimination and plasticity at the level of the auditory cortex, it is possible that far-field recordings from scalp electrodes may lack sensitivity to stimulus differences and changes in cortical activity after training or hearing loss. ERP protocols are needed that are reliably sensitive to differences between individuals and stimulus parameters and to changes in cortical auditory processing over time. Recently, research has focused on improving methodological variables such as the choice of stimuli, the stimulus presentation paradigm to enhance cortical response detection (Bardy, Dillon, & Van Dun, 2015) and testing multiple parameters at once (Näätänen, Pakarinen, Rinne, & Takegata, 2004).

Studies investigating encoding of signals in noise using a wide range of stimulus and recording paradigm have concluded that P1, N1, and P2 are differentially affected by the level, SNR and monaural versus binaural presentation (Whiting et al., 1998; Billings et al., 2011; Papesh, Billings, & Baltzell, 2014; Sharma, Purdy, Munro, Sawaya, & Peter, 2014). It is possible to measure CAEPs in people with hearing loss wearing hearing aids or cochlear implants (Korczac et al., 2005; McNeil et al., 2005). Studies in which CAEPs

have been recorded from people wearing hearing aids show that the response is largely influenced by the sensation level of the stimulus, type of stimulus and the SNR (Billings et al., 2012; Chang et al., 2012; Easwar et al., 2012; Papesh, Billings, & Baltzell). Papesh and colleagues showed that N1 wave was the best CAEP predictor of behavioural speech in noise abilities in adults with normal hearing. Correlations between P2 latencies and speech perception have been found in adults with hearing loss using cochlear implants (Kelly et al., 2005; Makhdoum et al., 1998). Thus, evoked cortical responses to speech and non-speech sounds potentially have a wide range of applications to supplement behavioural measures in hearing research and clinical audiology. The presence of a CAEP onset response indicates that sound is detected at the auditory cortex. Further work is needed, however, in adults using hearing instruments to establish whether aided CAEPs can provide additional information beyond simply indicating the detection of an amplified signal at the level of the auditory cortex. Another consideration in the clinical application of evoked potentials is the variation in cortical auditory responses across target populations; inter-subject variability and hearing loss and maturational/age effects on CAEPs can have significant effects on the generalization of data (Rufener, Liem, & Meyer, 2014).

Mechanisms for changes in neural activity underlying auditory evoked potentials include a range of short-term and long-term processes such as excitation (increased probability of neuronal firing), inhibition (restricted neuronal firing/regulation of excitatory neurons) (Sarro et al., 2015), adaptation (changes in neural activity in response to constant sensory stimulation) (Pérez-González & Malmierca, 2014), and long term potentiation (long-lasting increase in synaptic efficacy following high-frequency stimulation of afferent fibres) (Clapp et al., 2005). In addition to changes in sensory processing, top-down

processes such as enhanced auditory attention could also lead to changes in cortical evoked potentials (Hyde, 1997; Naatanen & Picton, 1987). Intervention studies that have used CAEPs to measure training effects have highlighted the possibility that altered neural activity results from repeated stimulus exposure during recording rather than reflecting training effects (Sheehan et al., 2005). Sheehan et al. proposed that the amplitude of P2 could be an index of inhibitory processes that are strengthened when an irrelevant sound is repeated (Crowley & Colrain, 2004). The auditory brain responds robustly to the onset of novel sounds. This attentive response is non-adaptive for repeated sound that has no personal significance, such as a ticking clock. Prolonged exposure to repeated sounds during evoked-potential recording may invoke inhibitory mechanisms so that the sounds no longer elicit attention. Changes in pre-training baseline CAEP recordings have been reported for children with auditory processing disorder, prior to their participation in a training study (Sharma, Purdy, & Kelly, 2014). Studies of adult listeners with normal hearing have also shown CAEP changes associated with repeated testing within and between sessions (e.g. Sheehan et al., 2005; Tremblay et al., 2010).

We need to better understand the physiology of CAEPs and the impact of factors such as signal acoustics, testing regime, hearing aid processing and auditory training. This information would facilitate the inclusion of CAEPs in auditory assessments to support more targeted and efficacious rehabilitation for people with SNHL. If CAEP characteristics in people with SNHL and the impact of rehabilitation on CAEPs are better understood, it may be possible to use CAEPs to predict or objectively describe the benefits of hearing rehabilitation.

Aims of research

Speech perception in noise in adults with SNHL is not only attributed to inaudibility but also to the inability of the listener to use important cues when sound is made audible. This thesis aimed to investigate the role of electrophysiology in understanding cortical processing of signals in noise in adults with normal hearing and in adults with mild to moderate SNHL.

This doctoral thesis includes three studies with the following aims:

Study 1) determined the effects of: (a) noise and aiding across speech stimuli, (b) different speech contrasts and (c) amplification (unaided versus aided) on speech-evoked CAEPs in adults with normal hearing. In this study also determined the effect of amplification on speech stimulus onset characteristics using in-the-ear acoustic measurements.

Study 2) aimed to increase understanding of processing of pitch cues in adults with SNHL and adults with normal hearing, using a combined behavioural and electrophysiological approach.

Study 3) investigated the effects of auditory training on pitch processing using behavioural and electrophysiological approaches in adults with SNHL.

Overview of the thesis

The studies in this thesis emphasize the contribution of ERPs to understanding suprathreshold processing within the central auditory system in people with SNHL. Three

studies were conducted as part of this doctoral thesis and are presented in Chapters 2-4. All three studies used electrophysiological and behavioural measures to better understand links between the characteristics of the auditory signal and neural processing.

In Chapter 2 ERPs were recorded using brief naturally produced speech signals in adults with normal hearing. These signals were tested to determine whether differential neural encoding was evident in various conditions that are frequently reported in literature (quiet, noise; unaided, aided). The findings highlighted the effects of stimulus acoustics on CAEPs. This study examined the effect of complex spectral and temporal speech stimulus differences on CAEPs. The following study used synthetic stimuli to better control for changes in stimulus acoustic characteristics and to focus on spectral and temporal fine structure changes.

In Chapter 3 ERPs were recorded to synthesized stimuli that varied in specific acoustic cues contributing to pitch perception. Processing of these cues has been previously described to play a crucial role in perception in noise (Moore 2008). A combined behaviour-electrophysiology approach was used to understand processing of and sensitivity to these cues in two groups of participants: normal-hearing listeners and listeners with high frequency SNHL. The previous study recorded obligatory CAEPs to a repeated stimulus. In the study described in Chapter 3 discriminative evoked potentials (ACC) were recorded to a change in the stimulus. ACC findings were compared to behavioural discrimination results.

In the two studies described in Chapters 2 and 3 a passive listening paradigm was used. The study described in Chapter 4 investigated whether non-linguistic auditory training improved neural processing and behavioural sensitivity to these cues. Training effects were investigated electrophysiologically using two types of test stimuli: training-related (complex tones) and non-related (speech). ERPs were recorded using an active listening paradigm. N1-P2 and P3 responses were evaluated to determine how listeners with SNHL process complex auditory signals using sensory and top-down processes.

The final chapter of the thesis summarises this research and discusses future research needs. This research shows effects of stimulus characteristics, hearing loss, amplification, auditory training and attention on auditory evoked potentials. More research will be needed before ERPs can be used in clinical practice to evaluate processing of complex auditory stimuli in people with hearing loss.

Chapter 2

Cortical Encoding of Speech Acoustics: Effects of Noise and Amplification

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This chapter includes content from the article "**Cortical Encoding of Speech Acoustics: Effects of Noise and Amplification**" published ahead of print in the International Journal of Audiology, 2015, doi: 10.3109/14992027.2015.1055838.

Introduction

Cochlear hearing loss leads to rapid and wide ranging changes in the auditory cortex which are yet to be understood completely (Lomber & Eggermont, 2006). These can be investigated using event related potentials, particularly the obligatory cortical auditory evoked potentials (CAEPs), which reflect the audibility and physical properties of a stimulus (Hyde, 1997; Martin et al., 2008). Historically CAEPs have been recorded using brief stimuli such as clicks and tone bursts. More recently studies have used synthetic and natural speech sounds, including natural vowels, consonants, consonant-vowel (CV) syllables and synthetic speech stimuli (Sharma et al., 2000; Tremblay et al., 2003; Agung et al., 2006; Korczak & Stapells, 2005, 2010). Auditory cortex neurons are more sensitive to transient changes at the onset of a stimulus than the presence of an on-going stimulus (Philips & Hall, 2002). CAEPs are primarily onset responses but are also produced by the offset of a stimulus (Pratt et al., 2008). Hence, CAEPs evoked by consonant-vowel (CV) syllables have overlapping onset responses evoked by the consonant and change responses evoked by the consonant-vowel transition and offset (Ostroff et al., 1998; Sharma et al., 2000; Martin et al., 2008; Digeser et al., 2009).

Previous studies using speech stimuli have demonstrated the ability of CAEPs to show encoding of speech features (Tremblay et al., 2003; Purdy et al., 2006; Korczak et al., 2010; Doellinger et al., 2011). For example, Digeser et al. (2009) found significant differences in CAEP waveforms in response to the spectro-temporally different consonant-vowel (CV) syllables, /da/ and /ta/. Group waveform differences in CAEPs to different speech stimuli are also evident in individual subject waveforms recorded in quiet for unaided and aided conditions (Tremblay et al., 2003, 2006).

It has been proposed that onset CAEPs reflect stimulus level relative to the level of background noise (SNR) rather than the absolute stimulus level (Billings et al., 2009; Baltzell & Billings, 2013). In general, as the SNR becomes unfavourable, the morphology of the CAEPs becomes poorer. Changes in tone and speech evoked CAEP morphology occurs for different masking stimuli, including white noise and speech (Whiting et al., 1998; Kaplan-Neeman et al., 2006; Billings et al., 2011). Obligatory CAEPs are sensitive to the acoustic characteristics of speech signals in noise (Kaplan-Neeman et al., 2006). Kaplan-Neeman compared CAEPs evoked by /da/ and /ga/; N1 latencies for both speech stimuli increased with the addition of background noise. N1 latencies were longer for /ga/ than for /da/ in quiet and in noise. The authors proposed that this was because of the long /ga/ burst duration. Acoustic cues specific to the speech stimuli were encoded even in the presence of background noise. Studies such as this could help understanding of difficulties discriminating specific acoustic cues in the presence of noise in people with hearing loss.

More recently CAEPs have been used as a clinical tool for validation of hearing aid fittings in children (Dillon, 2005; Purdy et al., 2005). This may be useful in infants and difficult to test populations where behavioural information regarding hearing aid benefit for speech perception is limited and unreliable (Carter et al., 2010). However, several studies have highlighted uncertainty in the clinical use of aided CAEPs and there is variability in the methodology and results across individual studies and participants (Korczak et al., 2005; Tremblay et al., 2006; Billings et al., 2007; Marynewich et al., 2012; Billings et al., 2011; Munro et al., 2011; Easwar et al., 2012b). Recent studies (Munro et al., 2011; Billings et al., 2012) investigating the clinical utility of aided CAEPs

suggest that it is valid to use CAEPs as an indicator of speech stimulus detection but not discrimination.

Hearing aids may use several nonlinear processing algorithms, such as compression and noise suppression, and may produce unwanted nonlinear distortions (Kim & Loizou, 2011). Prior studies indicate that SNR and stimulus onset modification (e.g., altered rise time) are two stimulus characteristics that have significant effects on aided CAEPs (Billings et al., 2007, 2009, 2013; Jenstad et al., 2012). Hearing aid output also varies with different speech contexts (phonemes in running speech vs. isolation), which may influence interpretation of aided CAEPs (Easwar et al., 2012a). Only a few studies have measured hearing aid output in the ear and correlated this to CAEP findings (Tremblay et al., 2006; Billings et al., 2007, 2009). Natural speech tokens are preferred to non-speech sounds for unaided and aided CAEP testing if the goal is to better understand speech processing in people with hearing loss (Aiken & Picton, 2008). Aided CAEPs could reflect changes in the short-term envelope of the stimulus triggered by changes in acoustic spectro-temporal features, level, and SNR due to the hearing aid. To our knowledge no study has examined the effects of speech stimulus onset characteristics on aided CAEPs.

Poor performance in noise is one of the frequent complaints reported by hearing aid users (Kochkin, 2007). It may be possible to use aided speech evoked CAEPs to better understand suprathreshold speech perception difficulties of people with sensorineural hearing loss using hearing aids. The present study investigated the effects of natural speech stimuli on CAEP components across a range of conditions, unaided and aided in quiet and in noise, in adults with normal hearing. When combined with acoustic analysis of the speech stimuli, CAEPs elicited by a number of naturally spoken speech sound

contrasts may help to better understand neural processing of place, manner, and voicing cues (Tremblay et al., 2003; Digeser et al., 2009; Korczak et al., 2010).The selection of speech contrasts in the current study was based on several factors, but was primarily based on speech features (place, manner, voicing). In addition, speech stimuli were selected that encompassed commonly observed speech errors (e.g. Warner-Czyz et al., 2010) and common consonant confusion errors in background noise of people with hearing loss (Doyle & Edgerton, 1981; Phatak et al., 2009; Woods et al., 2010). Studying these effects first in adults with normal hearing wearing a hearing aid allows us to separate the effects of amplification from hearing loss, since hearing loss is accompanied by frequency and temporal deficits and reduced audibility (Moore, 1996).

The current study investigated speech stimuli and background noise dependent changes in CAEP components in unaided and aided conditions and determined amplification effects on CAEPs in young adults with normal hearing. Specifically this study investigated effects of: a) noise (multi-talker babble) and aiding across speech stimuli b) speech contrasts, c) amplification (unaided versus aided) on speech CAEPs, and d) speech stimulus onset characteristics. In-the-ear probe-microphone measurements were used to determine effects of hearing aid on SNR, since previous research by Billings and colleagues (Billings et al., 2007, 2009) highlighted the impact of SNR on aided CAEPs.

Materials and Methods

Testing was conducted in two sessions of three hours each. Unaided and aided CAEPs were recorded in adults with normal hearing using seven speech stimuli (Figure 1). Participants were monaurally fitted in the right ear when testing the aided condition. For

all conditions, the left ear was plugged using a foam ear plug. Conditions (aided/unaided, quiet/noise) were counterbalanced and stimulus presentation order was randomized across participants and test sessions. In-the-canal signal acoustic measurements were done with and without the hearing aid to determine stimulus levels, amplification effects (unaided versus aided) on the SNR and to compare stimulus acoustic characteristics to CAEP findings.

Participants

Ten young adults with normal hearing were recruited (seven females, three males) aged 19 to 35 years (M 24 years, SD 4.6). All were right handed, English speakers, with pure tone audiometric thresholds of 20 dB HL or better at 250 to 8000 Hz and normal Type A tympanograms. CAEPs were recorded in quiet and in noise, with and without a hearing aid in place. For all conditions, the left ear was plugged using a foam ear plug. All participants gave informed consent before testing.

Stimuli

Stimuli consisted of naturally produced speech syllables (/di, ti, gi, mi, pi, si, ʃi/) and multi-talker babble as the masker. Speech stimuli and contrasts were selected based on differences in: 1) single speech features such as place (/di-gi/, /ʃi-si/, /ti-pi/) and voicing (/di-ti/), 2) multiple speech features including place, manner, and voicing (/mi-pi/, /mi-ti/,/gi-ti/), and 3) consonants dominant in low (<3 kHz) (/di, gi, mi/) versus high frequency (>3 kHz) (/pi, ti, ʃi, si/) energy. Speech syllables were recorded using a native New Zealand female speaker in a soundproof room via a AKG HC 577 L omnidirectional headset microphone placed 3 cm from the lips of the speaker attached to an M-Audio

MobilePre, using Adobe Audition version CS6 sound editing software, with a sampling rate of 44.1 kHz and 16 bit quantization rate.

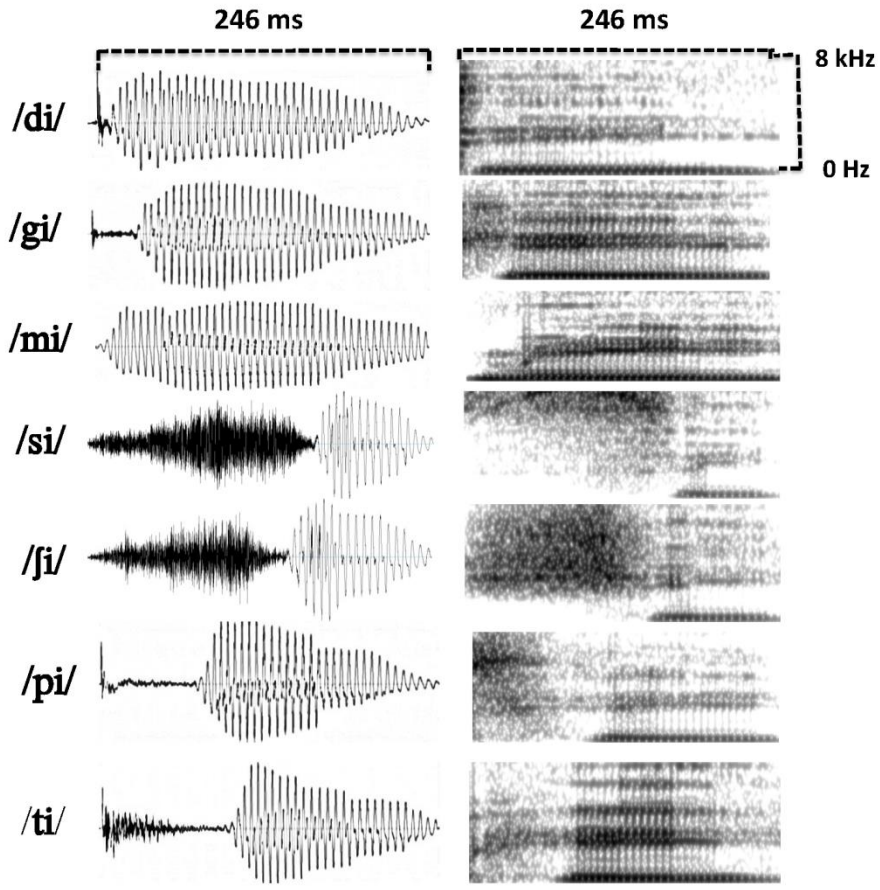

Figure 1. Consonant-vowel (CVs) acoustic stimuli used to elicit CAEPs. Time-domain waveforms (left column) and respective spectrograms (right column), derived using Praat 5.3.53 software, are shown. All stimuli were shortened from their original length to 246 ms.

Total duration of each syllable was 246 ms after editing; before editing each syllable was approximately 400-500 ms in duration. Segments were removed during the steady-state part of the vowel, starting and ending at zero crossings to prevent audible clicks. Stimulus

onsets were not changed in an effort to minimise the effect on CAEP waveforms, however the reduction in stimulus duration does reduce ecological validity of the stimuli. Stimuli were individually root mean square (RMS) normalized using Adobe Audition and presented via an Impact 50 Turbosound loudspeaker at 0-degrees azimuth at 1 m distance at 65 dB SPL (overall RMS). Spectrograms and time-domain waveforms for the seven stimuli derived using Praat version 5.3.53 are shown in Figure 1.

The competing noise signal was eight-talker babble presented via a DELL laptop. The babble noise comes from an anechoic recording of four males and four females reading out loud from different materials at the same time (NAL CD Speech and Noise for Hearing Aid Evaluation; Keidser et al., 2002). The spectra of the speech stimuli and multi-talker babble, showing the substantial differences in the region of dominant spectral energy across stimuli are shown in the Figure 2. Spectra were derived for the unaided and aided stimuli using in-the canal measurements and Adobe C S6 software. For each stimulus, babble noise was continuously presented for the entire duration of the noise condition, beginning a few seconds before the speech stimuli, at 55 dB SPL (+10 dB SNR). A favourable +10 dB SNR was used to investigate CAEP responses to hearing aid processed signals with all advanced features switched off. Kaplan-Neeman et al. (2006) found that noise affected CAEP latencies and amplitudes even at +15 dB SNR. The choice of SNR in the current study allowed robust responses to be recorded to a range of speech stimuli in noise.

Hearing aid

For aided cortical recordings a digital nonlinear Oticon Alta Pro behind-the-ear (BTE 13) hearing aid was coupled to a plastic closed temporary tip that occluded the right ear canal for all participants. Alta Pro uses a 10 channel sound processor. The processing delay

time of the hearing aid is 5 – 6 ms (Schum & Beck, 2006), with little variation across frequencies. The frequency range of this hearing aid extends from 100 to 7700 Hz; equivalent input noise is 18 dB SPL in a 2cc coupler. The hearing aid was programmed using Oticon's voice aligned compression (VAC) fitting rationale (Flynn, 2004), assuming a N4 audiogram, which has thresholds ranging between 55 and 80 dB HL and a pure tone average across .5, 1, and 2 kHz of 55 dB HL (Bisgaard et al., 2010). The VAC fitting algorithm provides curvilinear compression with both low-level compression, as well as increased linearity for high-level signals. The hearing aid was set to omnidirectional mode with all other auto-listening support features (noise reduction, feedback cancellation) disabled. In the VAC algorithm the hearing aid is programmed to provide more gain for soft-moderate speech and in the moderate to loud range, the compression ratio is unchanging and low even in the presence of background noise. Manufacturer specific ''first fits'' to a particular audiogram are significantly different and thus hearing aid output levels can be highly variable (Keidser et al., 2003).

In-the-canal acoustic measurements

The level of the hearing aid processed speech stimuli should affect amplitudes and latencies of CAEP components. Therefore, stimulus levels were measured with and without the hearing aid in place.

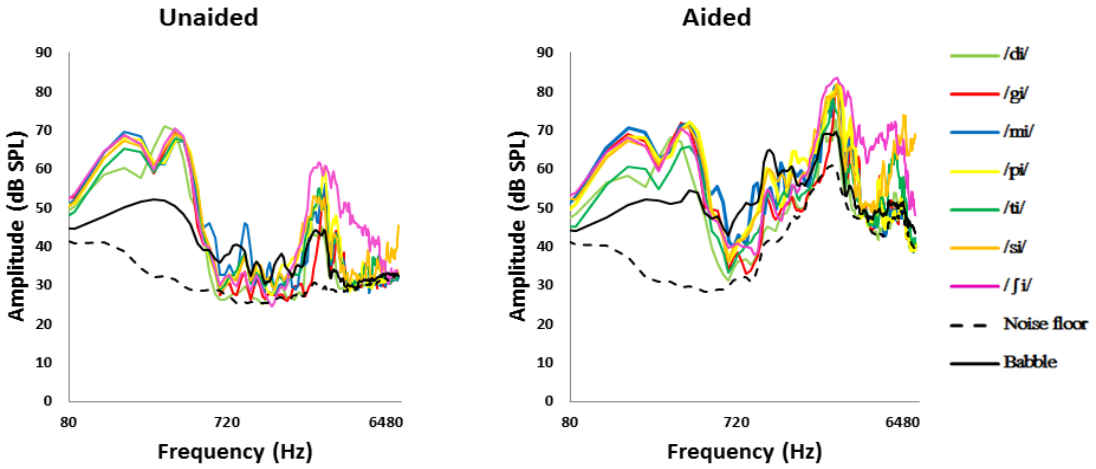

Figure 2. Frequency spectra of the speech stimuli, multi-talker babble, and hearing aid noise floor for unaided and aided conditions. Recordings were made in the ear canal of participants using an ER-7C probe microphone with speech stimuli presented at 65 dB SPL and multi-talker babble at 55 dB SPL**.**

Output levels of the hearing aid transduced stimuli for all conditions were measured in the ear canal of participants using an Etymotic Research Inc ER-7C probe microphone and preamplifier set to -20 gain (see Figure 2) connected to a DELL laptop. Stimuli were recorded and analysed using Adobe C S6 software. Measurements were performed by placing a probe tube 28 mm past the intertragal notch. To maintain correct positioning throughout testing, the probe tube was taped to the earlobe. For these measurements CAEP stimuli were delivered via the equipment used for evoked potential recordings. Figure 2 indicates that in the unaided condition, energy was most intense in frequencies below 500 Hz, presumably showing contributions from the vowel. The onset consonants were generally less intense than the vowel (by about 10 dB) as observed in running speech (Easwar et al., 2012a). In the aided condition, the high frequency emphasis onset

consonants were more intense than the lower frequency sounds (see Figure 2), as expected as the hearing aid was programmed to fit a sloping N4 audiogram configuration. Mean output levels of the speech stimuli in the presence of background noise were slightly more intense (by 6 dB on average) than for the quiet condition, presumably due to the VAC algorithm, which provides a more linear response when background noise is present. The peak output level of speech in the low and high frequency range was 69 and 58 dB SPL in the unaided condition and 72 and 82 dB SPL in the aided condition, respectively. Participants reported that the aided speech stimuli were ‘‘loud but comfortable’’.

SNRs for the unaided and aided conditions (in quiet) were measured for nine of the 10 participants using the in-the-canal recordings (Adobe C S6 software). SNRs for the unaided and aided conditions were obtained from one repetition of the stimulus presentation. Signal level was computed based on the average RMS levels of each speech stimulus. The level of the noise floor was measured from the interstimulus interval. The noise floor of the hearing aid was 40 to 58 dB SPL between 129 Hz and 7019 Hz with predominant spectral energy extending from about 990 to 7019 Hz. As gain was greater at higher frequencies, the noise floor of the hearing aid would also be greater at higher frequencies. SNRs were calculated individually for the initial 50 ms and the entire 246 ms of the speech stimuli. The 50 ms SNRs were computed because of the important influence of stimulus onset on CAEPs.

Electrophysiology

The Neuroscan SCAN^{TM} (version 4.3) was used for recording CAEPs. CAEPs were obtained using four EEG channels with 10 mm silver silver-choride disc electrodes placed

at Cz and Fz, referenced to M1 and M2. Data presented here are for the contralateral (M1) reference electrode (sound was presented frontally but the left ear was plugged). The ground electrode was located on the forehead and eye blink activity was monitored using electrodes placed above and below the right eye. Electrode impedances were under 3 kΩ. The Neuroscan STIM^{TM} system was used to present the speech stimuli. Two different randomized sequences of the seven stimuli were presented, with two blocks of 150 sweeps for each stimulus and each condition. Interstimulus interval was 920 ms. EEG was amplified with a gain of 1000 and sampled at the rate of 1000 Hz. EEG data were pre-processed using Neuroscan's built-in functions. Recordings with eye blink artifacts were corrected using a regression procedure, the ocular artifact rejection function in Neuroscan software. First, the vertical electro-oculogram (VEOG) channel was scanned for the maximum eye movement potential. EOG deviations of more than 10% from the maximum were used as indicators of blinks. A minimum of 20 blinks was required to estimate an average blink. The procedure discarded artifacts starting <400 ms before a previous artifact, to avoid double detection. From the average VEOG ocular artifact, transmission coefficients were computed for each EEG channel by estimating the covariance of the averaged potentials of the VEOG channel with the EEG channels. The contribution of the average blink from the VEOG channel was then subtracted from all other channels on a point-by-point basis. EEG epochs with -100 ms pre-stimulus to 600 ms post-stimulus time windows were extracted post hoc from the continuous file. Before averaging, responses were digitally filtered between .1 and 30 Hz. All recordings were baseline corrected before averaging. The artifact rejection threshold was set to ± 50μV.

Testing was performed in a double-walled sound attenuating booth. Sounds were calibrated using a Bruel and Kjaer 2215 sound level meter measured at 1 m distance from

the loudspeaker located in the participant's midline. Short breaks were given between testing conditions. Participants were seated comfortably on a reclining chair while watching a close captioned DVD of their choice (Lavoie et al., 2008). The study was approved by the University of Auckland Human Participants Ethics Committee.

Data analysis

Grand average CAEP waveforms were created for each participant by averaging two blocks of 150 runs for each condition. CAEP peak amplitudes and latencies were identified for each subject by two independent observers. Waves P1, N1, and P2 were analysed at Cz, as this electrode site gave the largest response waveforms. Repeated measures analyses of variance were performed separately on latencies and amplitudes of each component (P1, N1, and P2). Analyses included the 2 × 2 × 7 factors of noise condition (quiet and noise), aiding condition (unaided and aided) and speech stimuli. Post-hoc analyses of repeated measures ANOVA *interaction effects* were performed using paired t-tests. A Bonferroni correction (< .007) was used to adjust the alpha level of .05 to correct for the seven post-hoc stimulus contrast comparisons. The amplitude of P1 was defined as the largest positive deflection occurring between 50–125 ms after stimulus onset (see Appendix 1). The amplitude of N1 was identified as the largest negative deflection between 80–190 ms after stimulus onset. P2 amplitude was defined as the largest peak occurring between 170 and 290 ms. Peak latency was measured at the centre of the peak. When the waveform contained a double peak of equal amplitude or a peak with a plateau, latency was measured at the midpoint of the peak. Using these criteria it was possible to pick peaks for all participants and all speech stimuli. Peaks appeared to be present but were difficult to distinguish from the noise floor for responses to /si/ in babble for four individuals. For these four measurements the residual noise level present in the

averaged response within the latency region of interest (determined from the grand averaged waveform) was used as the estimate of peak amplitude and the latency was picked at the point of peak amplitude within the latency region of interest. To evaluate variations in SNR for the quiet condition a repeated measures analysis of variance was performed for the 2 × 2 × 7 factors of amplification (unaided versus aided SNRs), stimulus time window (50 ms and 246 ms), and speech stimuli.

Results

CAEP latencies and amplitudes

CAEPs with characteristic morphology were elicited for all participants across all conditions. The repeated measures ANOVA (Table 1) showed significant main and interaction effects on N1 and P2 latencies and amplitudes for the three independent variables (noise, speech stimuli, and aiding). Means and SDs of latencies and amplitudes (Cz electrode) across all testing conditions are listed in Appendix 1. Effects on P1 latencies and amplitudes were only seen for noise and speech stimuli.

Table 1. ANOVA Results. Repeated measures analyses of variance (ANOVA) results for data collected at electrode Cz. Results for latency and amplitude across components P1, N1, P2 are included.

	Noise		**Stimulus**		**Aiding**		**Noise*Stimulus**		**Noise*Aiding**		**Aiding*Stimulus**		**Aiding* Noise* Stimulus**	
	***F*-stat (df)**	***p*-value**	***F*-stat (df)**	***p*-value**	***F*-stat (df)**	***p*-value**	***F*-stat (df)**	***p*-value**	***F*-stat (df)**	***p*-value**	***F*-stat (df)**	***p*-value**	***F*-stat (df)**	***p*-value**
Latency														
P1	55.6 (1,9)	0.001	8.5 (6,54)	0.001	4.0 (1,9)	0.070	7.9 (6,54)	0.001	0.1 (1,9)	0.754	1.8 (6,54)	0.112	1.5 (6,54)	0.182
N1	235.1 (1,9)	0.001	23.1 (6,54)	0.001	2.8 (1,9)	0.125	29.2 (6,54)	0.001	1.5 (1,9)	0.332	2.7 (6,54)	0.019	2.8 (6,54)	0.016
P2	249.1 (1,9)	0.001	42.9 (6,54)	0.001	10.5 (1,9)	.010	29.7 (6,54)	0.001	0.7 (1,9)	0.778	3.4 (6,54)	0.007	1.1 (6,54)	0.358
Amplitude														
P1	7.3 (1,9)	0.024	0.8 (6,54)	0.539	0.7 (1,9)	0.785	0.9 (6,54)	0.469	4.6 (1,9)	0.059	2.6 (6,54)	0.025	0.3 (6,54)	0.926
N1	0.1 (1,9)	0.706	11.5 (6,54)	0.001	2.5 (1,9)	0.145	4.8 (6,54)	0.001	8.3 (1,9)	0.018	0.1 (6,54)	0.173	0.4 (6,54)	0.827
P2	3.6 (1,9)	0.089	5.3 (6,54)	0.001	4.0 (1,9)	.075	5.5 (6,54)	0.001	0.4 (1,9)	0.584	0.9 (6,54)	0.473	0.7 (6,54)	0.580

Effects of noise across speech stimuli

A general trend evident in Figure 3 (a) & (b) was an increase in latency and decrease in amplitude of all CAEP components in noise for the unaided and aided conditions across all speech stimuli. A significant main effect of noise was seen for latencies of P1, N1, and P2 (P1: $F[1,9] = 55.61$, $p < 0.001$; N1: $F[1,9] = 235.13$, $p < 0.001$; and P2:$F[1,9] = 249.13$, $p < 0.001$). Significant main effects of noise on amplitudes was seen only for P1 (P1: $F[1,9] = 7.36$, $p = 0.024$) and there was also a trend for smaller P2 amplitudes in noise (Table 1). There were significant Noise × Stimuli interaction effects for all latencies (P1: $F[6,54] = 7.94$, $p < 0.001$; N1: $F[6,54] = 29.24$, $p < 0.001$; and P2: $F[6,54] = 29.72$, $p < 0.001$). Post-hoc paired t-tests to determine the influence of noise for different speech stimuli (averaged across aiding conditions) showed no significant latency differences ($p \geq 0.007$) in quiet versus noise for P1 for two speech sounds: (/ʃi/: $t[9] = -0.006$, $p = 0.995$); /si/: $t[9] = 0.67$, $p = .519$) and for P2 for three speech sounds (/ti/: $t[9] = -0.65$, $p = 0.595$); /ʃi/: $t[9] = -1.11$, $p = .296$); /si/: $t[9] = .34$, $p = 0.742$). All other speech stimuli had significantly longer latencies in noise than in quiet ($p \leq 0.001$). A three-way interaction between Aiding × Noise × Speech stimuli was seen for N1 latency ($F[6,54] = 2.85$, $p = 0.016$) and hence post-hoc paired t-tests were conducted to determine the effect of noise on N1 latencies for each stimulus separately for aided and unaided conditions. Two speech stimuli failed to show significant latency differences for quiet versus noise in the unaided condition: /si/: $t[9] = -1.78$, $p = 0.108$); /ʃi/: $t[9] = -2.01$, $p = 0.075$). Three speech stimuli did not show a difference in the aided condition: /ti/: $t[9] = -2.06$, $p = 0.013$); /si/: $t[9] = .96$, $p = 0.362$); /ʃi/: $t[9] = -2.62$, $p = 0.028$). In general, the noise did not affect latencies for high frequency emphasis speech stimuli.

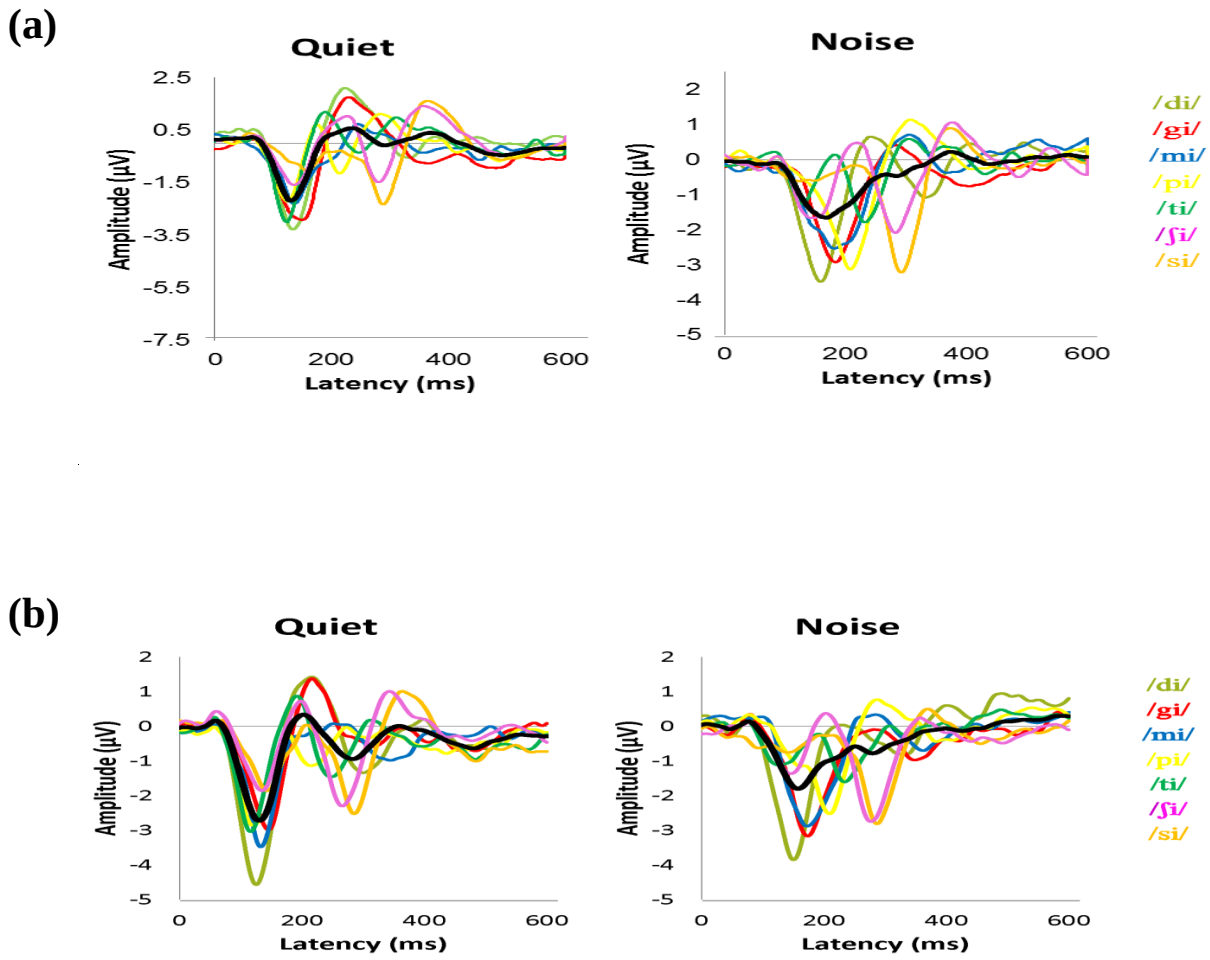

Figure 3. (a, upper) Unaided grand mean CAEP waveforms (N = 10) recorded at Cz for each speech stimulus in quiet and in noise. The waveform in black is the grand average of all stimuli for each condition (quiet and noise). (b, lower) Aided grand mean waveforms (N = 10) recorded at Cz for each speech stimulus in quiet and in noise. The waveform in black is the grand average of all stimuli for each condition (quiet and noise).

There were significant two-way interactions for Noise × Stimuli for N1 and P2 amplitudes (N1: $F[6,54] = 4.84$, $p < 0.001$; and P2: $F[6,54] = 5.52$, $p < 0.001$). As was observed for latencies, the effect of noise on N1 amplitude was inconsistent across stimuli. Post-hoc comparisons of N1 amplitudes in quiet versus noise showed no significant effects of noise, $p > 0.007$. Although a general trend of reduced P2 amplitude in noise was seen across all speech stimuli, post-hoc comparisons also showed no significant effect of noise on P2 amplitude ($p > 0.007$). A Noise × Aiding interaction was seen only for N1 amplitude (N1: $F[1,9] = 8.3$, $p = 0.018$); N1 amplitude was slightly increased by noise for the aided condition, by 12% on average, and reduced on average by noise in the unaided condition by 15%. (see Appendix 1)

Effect of speech contrasts

A significant main effect of speech stimulus was seen for latencies of P1, N1, and P2 (P1: $F[6,54] = 8.53$, $p < 0.001$; N1: $F[6,54] = 23.16$, $p < 0.001$; and P2:$F[6,54] = 42.94$, $p < 0.001$). Significant main effects of stimulus on amplitudes were seen for N1 and P2 (N1: $F[6,54] = 11.50$, $p < 0.001$; P2: $F[6,54] = 5.30$, $p < 0.001$). CAEP latencies and amplitudes were compared for speech contrasts in Table 2. Results of paired t-tests for unaided speech contrasts along with comparable findings from previous unaided CAEP studies are summarized. Post-hoc comparisons for many of the unaided speech contrasts showed amplitude and latency differences in quiet and/or in noise. Overall, Table 2 shows there were more CAEP latency than amplitude differences between speech contrasts. In general, significant differences in CAEPs were observed within a stimulus pair when the onset characteristics of the speech stimuli differed substantially (e.g., /mi/ versus /ti/).

Table 2. Comparisons of published speech-CAEP studies findings to unaided results from the current study at Cz. NS=not significant (p > 0 .007)
Voice onset time (VOT) Place of Articulation (POA) Manner of Articulation (MOA)

Speech contrasts	**Main feature Contrast**	**Summary of findings from present study**		**Previous Studies**	**Summary of findings from previous studies in quiet**
		Quiet	*Noise*		
/di – ti/	VOT	/ti/ significantly shorter latency for P2*	/ti/ significantly shorter latencies & smaller amplitude for N1* & P2*	Digeser et al. (2009) /da –ta/	/ta/ significantly shorter latencies for N1 and P2 & larger N1-P2 amplitudes
/mi – pi/	MOA/VOT	/pi/ significantly shorter P2*	NS	N/A	N/A
/di – gi/	POA	/di/ significantly shorter latency for N1*	/di/ significantly shorter latency for N1**	Kaplan-Neeman et al. (2006)	/da/ significantly shorter latency for N1 compared to /ga/
/ʃi/ – /si/	POA	NS	NS	Tremblay et al. (2003)	/si/ larger onset N1 amplitude compared to /ʃi/
/mi – ti/	POA/MOA/VOT	/ti/ significantly shorter latencies for N1* & P2*	/ti/ significantly shorter latencies for N1** & P2**	Purdy et al. (2006)	/t/ significantly shorter latency & smaller amplitude for N1 in infants compared to /m/
/ti – gi/	POA/VOT	/ti/ significantly shorter latencies for N1* & P2*	/ti/ significantly shorter latencies for N1** & P2**	Purdy et al. (2006)	/t/ significantly shorter latency & smaller amplitude for N1 in infants compared to /g/

Note: * = p ≤ 0.007 unaided, ** = p ≤ 0.007 aided

Effects of speech contrast were similar for aided and unaided conditions (Table 3), however, a few contrasts had significant latency and amplitude differences in the aided, but not the unaided condition. Enhanced CAEP differences between speech contrasts for the aided condition were mainly seen for latency measures.

Table 3. Statistically significant CAEP latency (L) & amplitude (A) measures for post hoc comparisons of speech contrasts where the main effect of stimuli was significant.

Speech Contrast	Quiet						Noise					
	P1 L	P1 A	N1 L	N1 A	P2 L	P2 A	P1 L	P1 A	N1 L	N1 A	P2 L	P2 A
/di/ vs /ti/					*#				*#	*#	*#	
/di/ vs /gi/		#	*#				#		*#		#	
/ti/ vs /mi/			*#		*#		#		*#		*#	
/ti/ vs /gi/			*#		*#				*#		*#	
/si/ vs /ʃi/												

Note: * = p ≤ 0.007unaided, # = p ≤ 0.007 aided

Effects of amplification (aided versus unaided)

Although a main effect of hearing aid amplification was seen only for P2 latency (F[1,9] = 10.56, p = 0.010), a statistical trend was observed for P2 amplitude (Table 1). In general, compared to the unaided condition, P2 was later and larger for the aided condition. Two-way interactions between Aiding × Stimuli were found for N1 (F[6,54] = 2.17, p = 0.019) and P2 latencies (F[6,54] = 3.49, p = 0.007), however post-hoc comparisons (averaged across noise condition) showed that latencies, after correcting for hearing aid delay, were not significantly different between unaided and aided conditions across speech stimuli, $p \leq 0.007$.

Effects of amplification on SNR

The effects of amplification (unaided vs. aided) on the In-the-canal measurements of SNR for each stimulus were investigated using repeated measures ANOVA (Figure 4). The main effect of higher SNRs in the aided compared to the unaided condition was significant ($F[1,8] = 78.13$, $p < 0.001$), averaged across speech stimuli and time window. Apart from some minor variations, unaided SNRs were similar (within 2 dB) across stimuli when computed across the entire speech stimulus (246 ms), as expected because the stimuli were amplitude normalized. There was a significant two-way interaction between Aiding x Speech stimuli ($F[6,48] = 106.17$, $p < 0.001$); the effect of aiding on SNR varied across stimuli. In general, aided SNRs were better than unaided and reflected the frequency response of the hearing aid, with higher SNRs for high frequency emphasis speech stimuli. The hearing aid was programmed for a sloping moderate-severe hearing loss, with more gain at high frequencies. The drop in SNR for /si/ may be due to the spectral peak for /si/ at >8000 Hz being outside the hearing aid's effective amplification range (100 – 7700 Hz). Three high frequency emphasis speech sounds (/pi, ti, ʃi/) showed very high SNR values, particularly for the aided condition. SNR calculations based on comparison of pre-stimulus noise floor to the stimulus level could overestimate SNR as this calculation does not consider the effect of input gain on the frequency spectrum of the hearing aid noise floor. However, a comparison of real ear measures of output (Figure 2) to noise floor values also indicates better aided SNRs for the high frequency emphasis speech stimuli.

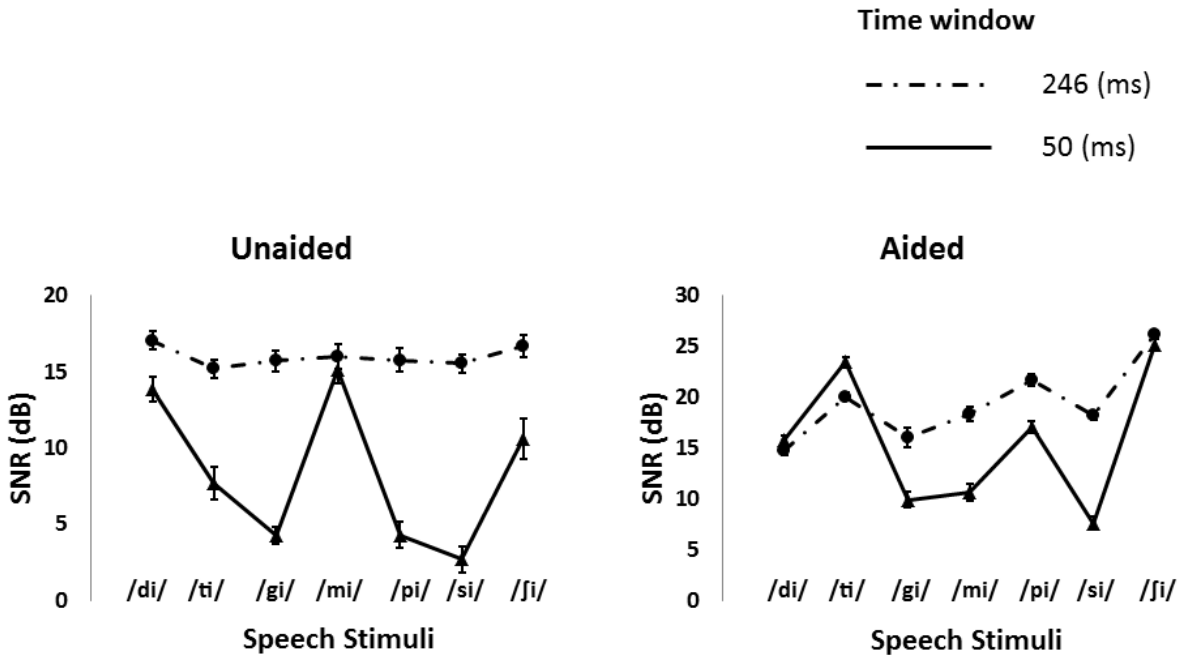

***Figure 4*:** Unaided (left column) and aided (right column) acoustic SNRs across speech stimuli for the initial 50 ms and the entire 246 ms of the stimuli. Measurements were made with and without the hearing aid in place and the speech stimuli presented at 65 dB SPL. The hearing aid was programmed assuming an N4 audiogram. Aided SNRs were better than unaided, with higher SNRs for high frequency emphasis speech stimuli. Error bars show standard errors.

There were significant differences between 50 and 246 ms speech stimulus time windows for SNR measurements ($F[1,8] = 1120.98$, $p < 0.001$). There was also a significant three-way interaction between Stimulus time window × Aiding × Speech stimuli ($F[6,48] = 90.50$, $p < 0.001$). SNR was lower for the onset for five of the speech stimuli (/gi, pi, ti, si, ʃi/) (Figure 4). These speech stimuli had long onset burst durations that did not include the vowel. The stimulus /di/ has a short burst duration; for this stimulus SNR values did not differ between onset and the total stimulus.

Time window did not influence SNR for /mi/ for the unaided condition, but the onset SNR was reduced in the aided condition because the hearing aid changed the onset amplitude (Figure 5). The onset SNR is increased for /ti/ for the aided condition due the effect of the hearing aid gain on the onset burst (Figure 5).

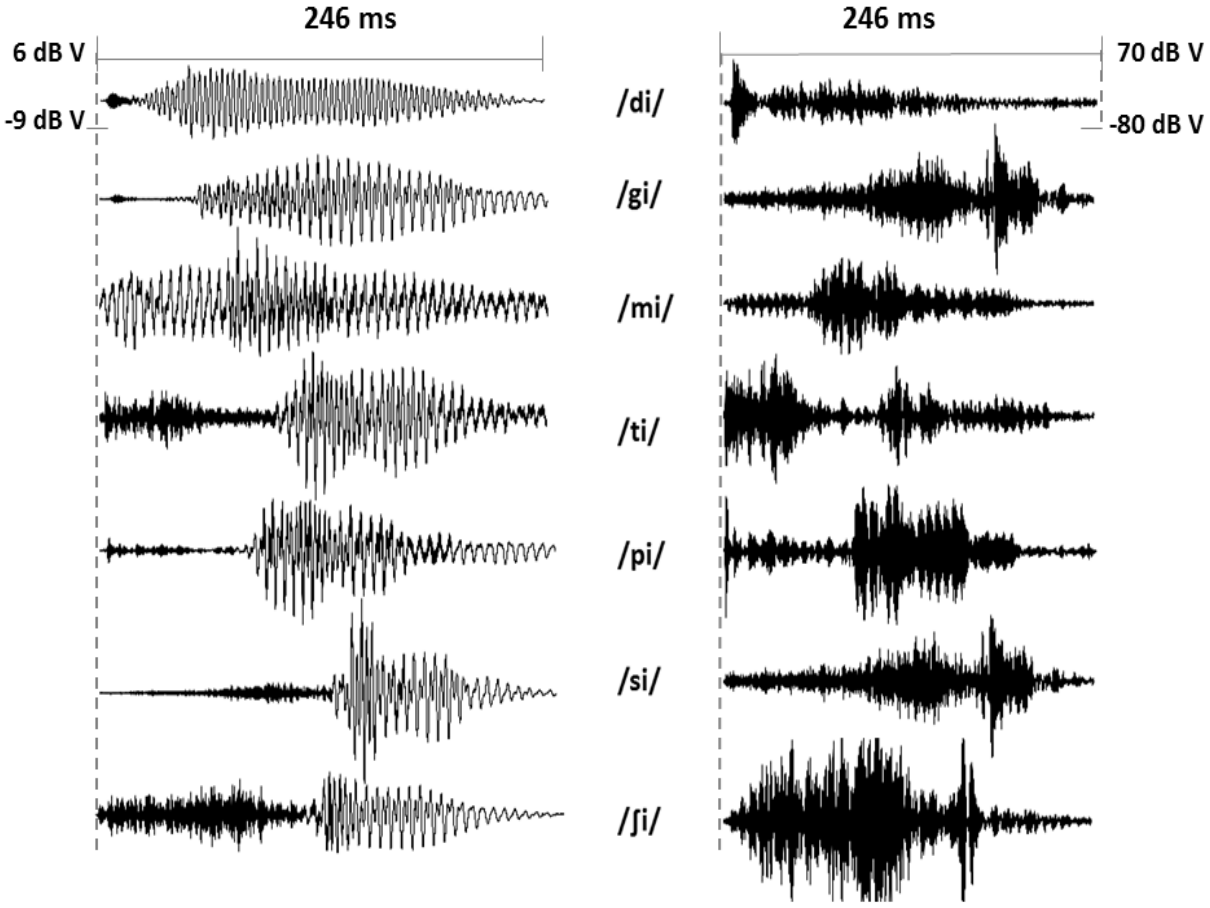

Figure 5. Acoustic waveforms of the speech stimuli recorded in the ear canal with (right column) and without (left column) the hearing aid in place. Aiding alters the onset envelope for sounds, e.g. /mi/. The hearing aid introduces a 5-6 ms stimulus delay that is not evident from this figure due to the noise floor in the acoustic recordings.

Discussion

Summary of key findings

Our results demonstrate that speech stimuli and background noise can affect CAEP components for both unaided and aided conditions. The addition of background noise resulted in increased N1 and P2 latencies and decreased P2 amplitudes for most speech stimuli. Furthermore, CAEPs showed neural encoding of different speech stimuli and acoustic features. The data indicate that stimulus onset characteristics influence CAEPs. Acoustic measures of hearing aid output (SNR and real-ear output level) were quantified. Overall, aiding resulted in later and larger P2 responses.

Effects of noise across speech stimuli

We investigated the neural representation of unaided and aided CV syllables in quiet and in noise (10 dB SNR). Noise and aiding are known to affect CAEPs (Whiting et al., 1998; Korczak et al., 2010) but previous studies have not investigated a wide range of natural speech stimuli. Both amplitudes and latencies were affected by background noise although, overall, noise had more effect on latencies. N1 and P2 latencies increased and P2 amplitudes decreased in noise. Increased CAEP latencies in noise is a consistent pattern seen across studies (Kaplan-Neeman et al., 2006; McCullagh et al., 2012; Billings et al., 2013). This can be attributed to disruption of synchronised neural discharges to stimulus onset. Although significant increases for N1 and P2 latencies and a trend for reduction of P2 amplitudes were evident, N1 amplitude was not consistently affected by background noise. This could be because cortical neurons adaptively adjust their thresholds relative to the background noise (Philips, 1990). Therefore, a proportional decrease in N1 amplitude with background noise level may be evident only once the

masker level is above a certain threshold for effecting a reduction in N1 amplitude. This is consistent with Whiting et al's finding that N1 amplitude significantly reduced once SNR reduced to < 5 dB. N1 amplitude increase in noise for the aided condition may be the result of slightly intense hearing aid output when background noise was present (by 6 dB on average).

Effects of multi-talker babble on CAEP morphology varied across stimuli, which may reflect the spectral relationship between the noise and speech stimuli. Latencies for the three high-frequency emphasis sounds (/ti, si, ʃi/) were not affected by noise, presumably because the multi-talker babble had little energy at high frequencies (Figure 2), and hence had minimal impact on the onset envelope. Gordon-Salant (1985) reported similar spectrum dependent energetic speech babble masking of CV syllables based on behavioural measures of speech perception. This frequency-specific masking effect was less apparent for amplitudes. Although P2 amplitudes consistently reduced in noise this did not reach statistical significance. N1 amplitude also did not show a consistent reduction in amplitude across stimuli in noise. The +10 dB SNR used in the present study was not low enough for the noise to substantially decrease CAEP amplitudes consistently across all speech stimuli (Whiting et al., 1998).

Effects of speech contrasts on CAEPs

While CAEPs may be a suitable approach for studying speech processing in people with hearing loss, age-related hearing and cognitive deficits, and auditory processing disorder (Billings et al., 2013; Sharma et al., 2014; Wilson, 2013), the ability of CAEPs to show encoding of different speech features in people with hearing loss wearing a hearing aid is not yet established. The different natural speech stimuli investigated in the current study

evoked distinct neural response patterns based on group data. These distinct patterns were also reliably recorded in individuals (see Figure 6). Feature differences between speech contrasts, CAEP differences and findings from previous studies are summarized in Table 2. Overall results of indicate that CAEPs were sensitive to the various speech features investigated here. There were significant latency and/or amplitude differences for each contrast, with the exception of /ʃi /-/si/.

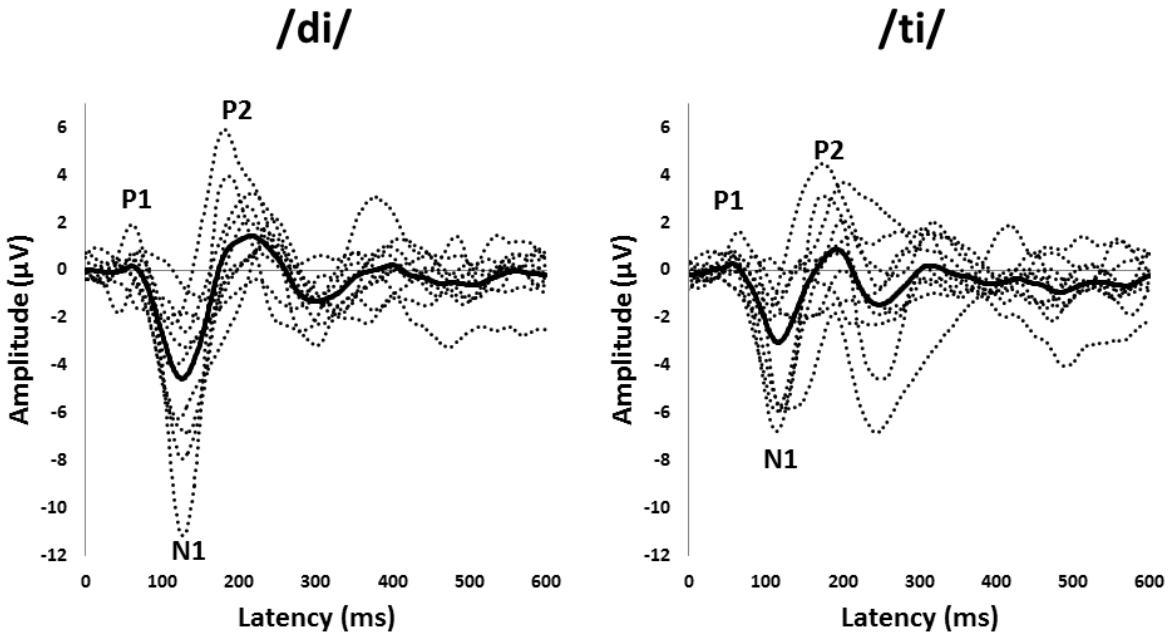

Figure 6. Grand average CAEP waveforms (solid line) along with superimposed individual waveforms of all participants (dotted lines) recorded at Cz for /di/ and /ti/.

In general, CAEP responses to stop consonants (/di, ti, pi, gi/) were earlier and larger compared to steady consonants (/mi, si, ʃi/) (Figure 7). A possible explanation for the stop consonant effect on CAEPs might be the sharp onset and higher burst energy of the stop consonants compared to the steady/non-stop consonants (Gage et al., 1998; Young, 2008). Similar findings were reported by Golding et al (2006) for the unvoiced stop consonant, /t/, which produced greater CAEP amplitude and shorter latency than a voiced nasal consonant, /m/. Low frequency stop consonants /di/ and /gi/ produced larger amplitude responses compared to high-frequency emphasis stops, /ti/ and /pi/. This is consistent with previous studies showing that high-frequency sounds produce smaller CAEPs than low-frequency sounds of the same intensity (Picton et al., 1978). Antinoro et al. (1969) also observed a decrease in peak-to-peak amplitude with increase in tone burst frequency. Also, aperiodic speech sounds have been shown to evoke smaller N1m responses than periodic sounds (Yrttiaho et al., 2008). CAEPs to /si/ and /ʃi/ were generally smaller compared to other speech stimuli in the current study. This difference in CAEP morphology could be attributed to the high frequency content of the onset consonant or the slower rise time and lack of burst energy at the onsets of these stimuli. Responses to fricative-vowel stimuli are usually dominated by prominent later peaks associated with the transition from the consonant to the vowel (Tremblay et al., 2003).

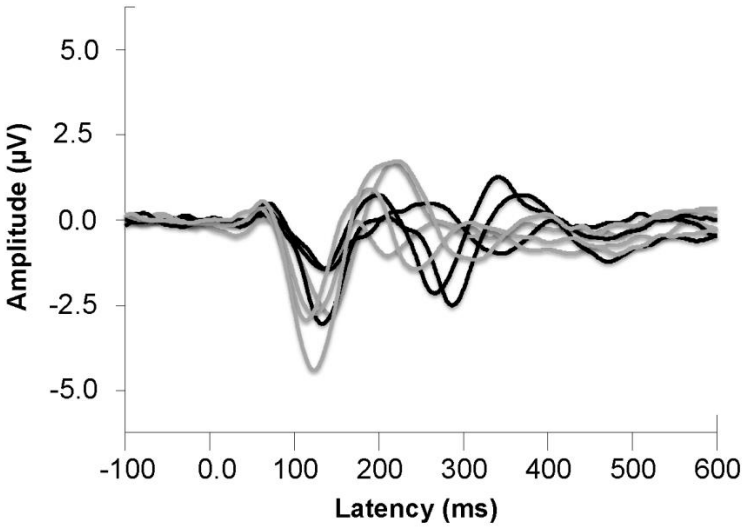

Figure 7. Grand average unaided CAEP waveforms in response to steady consonants in black (/mi, ʃi, si/) and stop consonants in grey (/di, ti, gi, pi/). CAEPs evoked by stop consonants were earlier and larger compared to the steady consonants.

A previous study by Sharma et al. (1999) proposed that N1 morphology reflected changes in VOT. A later paper, however, concluded that N1 morphology is not the cortical correlate of VOT differences between voiced and unvoiced speech stimuli (Sharma et al., 2000). An interesting finding in the current study was the difference in morphology between CAEPs evoked by voiced and voiceless consonants (Figure 8). P2 and the following negativity were broader and significantly later for voiced consonants. This distinction could be attributed to the continuity in the spectral and temporal characteristics of voiced CVs compared to voiceless CVs. Voiced consonants have a short VOT and an onset burst with spectral energy similar to the following vowel's formant frequencies. In

contrast voiceless consonants have longer VOT and little energy in the first formant (F1) frequency region before the onset of the vowel (Figure 1). The pause in the F1 frequency region for voiceless consonants could be contributing to the negativity/positivity in the 200-400 ms region seen for voiceless but not voiced CVs. For example, in Figure 8, latencies of the negative peaks between 200 and 300 ms for the voiceless speech stimuli are arranged in time as one would expect given the increasing delays of voicing onset (/pi < ti < ʃi < si/).

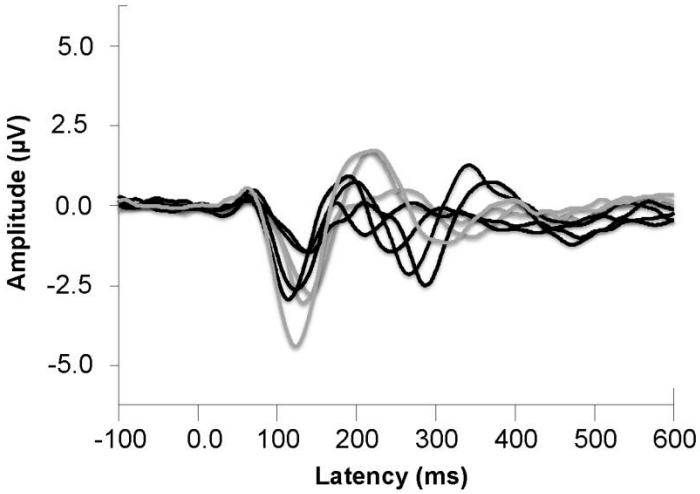

Figure 8. Grand average unaided CAEP waveforms in response to voiceless stimuli in black (/pi, ti, ʃi, si/) and voiced stimuli in grey (/di, gi, mi/). P2 was broader and later for CAEPs evoked by voiced stimuli.

This pattern was also reported by Ostroff et al. (1998) for /sei/ and by Tremblay et al. (2006) for /si/ and /ʃi/. Digeser et al. (2009) also found that short duration CV syllables evoked CAEP components that interfered with each other. The overlap in CAEPs to

successive portions of speech was noted by Aiken et al. (2008) for responses to sentences. The current study suggests that CAEPs to CV stimuli are composite responses to short-term spectro-temporal characteristics such as rise time, primarily influencing the onset response at N1, and a P2/acoustic change complex dependent on the consonant vowel transition (Ostroff et al., 1998; Martin et al., 2008).

Effects of amplification on onset CAEPs

N1's sensitivity to the stimulus envelope/rise time at stimulus onset is well documented (Onishi & Davis, 1968; Easwar et al., 2012c). Sensitivity of CAEPs to stimulus onset characteristics is especially important when considering hearing aid complex automatic signal processing, hearing aid processing delay, and expansion and compression characteristics where the time constants vary, but can be as short as a few milliseconds.

Jenstad et al. (2012) measured the effect of hearing aid processing on the onset time of their 1000 Hz stimulus and linked this to aided CAEP amplitudes. Depending on the hearing aid gain and frequency response the prominence of different frequency regions in the stimulus will change with amplification. In the present study the hearing aid gain characteristics typically changed the stimulus envelope at the onset, introducing short-term amplitude contrasts between /mi/ and /ti/, for example (Figure 5). The contrast /mi/ vs. /ti/ produced a 23 ms greater N1 latency difference in the aided condition compared to the unaided condition. N1 latencies for /mi/ were further increased in the aided condition, which may be due to the effect of the hearing aid altering stimulus specific rise times or the effect of frequency specific gain characteristics. The latter seems more likely as the

gain varied between the two speech stimulus (Figure 5) and the hearing aid caused a greater change for /mi/ than /ti/. It is difficult to separate these two factors, however.

Previous research has demonstrated that interpretation of cortical neural processing of amplified speech signals can be problematic because of known confounding factors such as SNR, hearing aid processing delay, and onset modifications induced by the type of hearing aid (Billings et al., 2012; Marynewich et al., 2012). In the current study, aiding was associated with significantly later (and a trend for larger) P2 responses. Ear canal acoustic measurements showed that aided stimulus SNR was higher than unaided for most of the stimuli (Figure 4). Larger P2 amplitudes could be linked to the higher SNR and effects of higher output levels in the aided condition. Previous studies have reported an increase in P2 and N1-P2 amplitude with 20 dB of gain compared to unaided (Billings et al., 2007; Marynewich et al., 2012). N1 amplitude has been shown to asymptote at intensities ~70 dB, which may account for the hearing aid not affecting N1 amplitude in the present study (Adler & Adler, 1989). A few speech contrasts produced significant latency and amplitude differences for the aided, but not the unaided condition (Table 3), which may reflect the hearing aid altering envelope rise time due to the effects of nonlinear gain and higher noise floor. Because of complex effects of amplification on speech stimuli a better understanding of the effects of hearing aid processing on CAEPs may be gained using hearing aid transduced signals controlling for factors such as signal processing delay, SNR, stimulus levels, and aided spectrum (Billings et al., 2012; Easwar et al., 2012b), in individuals with and without hearing loss.

Effects of stimulus parameters on N1 versus P2

Although N1 and P2 components co-vary they can be differentiated using experimental conditions such as those used in the present study. Stimulus conditions (noise and aiding) affected N1 only when the stimulus envelope was altered. For example, in the aided condition N1 was affected only for the speech stimuli that had changed envelope shape as a result of amplification e.g., /mi/ vs. ti/. This is consistent with the view that N1 is a transient response evoked as a response to envelope change (Onishi et al., 1968). A relatively high SNR (+10 dB) was used in the current study for the noise condition, which could explain why N1 amplitude did not reduce consistently in noise. Latencies increased only when the noise interacted with stimulus envelope for stimuli with spectral profiles similar to the background noise (Martin et al., 1999).

P2 is sensitive to attention and stimulus parameters such as intensity and pitch (Crowley et al., 2004). A recent study associated P2 with processing of altered phrase boundaries in music, especially in the context of harmonics (Istók et al., 2013), suggesting important effects of pitch and harmonics on P2. Amplification from a hearing aid introduces modulations or additional harmonics across the entire bandwidth of the signal, which changes the timbre of the aided output signal (Chasin & Russo, 2004). A listener with normal hearing can distinguish these differences in timbre, which may account for morphological changes in the P2 component in the aided compared to the unaided condition.

Although N1 and P2 originate from different sources and are functionally different responses (Ross & Tremblay, 2009), the distances between their sources are small and it is likely that P2 sources in planum temporale overlap with N1 sources with a centre of

activity near Heschl's gyrus (Crowley et al., 2004). Planum temporale has been described as the ''computational engine'' for the segregation and spectro-temporal matching of complex sounds (Griffiths & Warren 2002, p. 348), responsible for pitch processing and melodic perception (Zatorre et al., 1998; Keenan et al., 2001). Activity in Heschl's gyrus and planum temporale is enhanced by stimulus pitch differences (Schadwinkel & Gutschalk, 2010). Differential effects of noise and aiding on N1 and P2 suggest differential effects of processing of stimulus pitch, timbre, and envelope on these components..

Summary and conclusions

Stimulus characteristics and hearing aid model and settings are important factors to consider when investigating aided CAEP responses. Overall the current study showed that CAEPs evoked using natural stimuli are sensitive to adverse effects of background noise, onset characteristics of the stimuli, and spectro-temporal differences between speech stimuli. N1 and P2 components varied differently across noise and aiding conditions. Future studies differentiating N1 and P2 components in relation to pitch, timbre and envelope cues will be useful. Nonlinear hearing aids with adaptive features may process speech differently in babble than they do in stationary noise and hence it would be useful to determine hearing aid effects on speech-CAEPs for different types (and levels) of background noise. Establishing that CAEPs can reflect differences in speech acoustic features is a first step, further studies linking this to hearing aid signal processing strategies and central auditory processing of speech stimuli are needed to further explore the potential use of CAEPs for evaluating hearing aids.

Chapter 3

Electrophysiological and behavioural processing of complex acoustic cues

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This chapter includes content from the article "**Electrophysiological and behavioural processing of complex acoustic cues**" published ahead of print in Clinical Neurophysiology, 2015, doi: 10.1016/j.clinph.2015.04.002.

Introduction

The impact of sensorineural hearing loss (SNHL) is particularly noticeable while listening to speech in noisy backgrounds (Festen & Plomp, 1990; Gordon-Salant, 1985). Even when amplification is provided, a persistent complaint of hearing aid users is difficulty understanding speech in noise (Kochkin, 2007). A listener's ability to extract cues for pitch perception is an important factor for successful communication in background noise. The main acoustic cues contributing to the streaming of signals in noise are the slowly varying temporal envelope (ENV) and the rapidly varying temporal fine structure (TFS) (Moore, 2014). While ENV cues are primarily important for speech perception in quiet, TFS cues are important for speech perception in noise, sound localisation, music perception, and pitch perception (Moore, 2008). Recent studies using psychophysical measures have shown that listeners with SNHL have reduced ability to benefit from TFS information while the perception of ENV information is well preserved (Hopkins et al., 2008; Lorenzi et al., 2006, 2009; Moore et al., 2006b). It is thought that this lack of TFS sensitivity might account for poor speech understanding in noise and music perception in individuals with SNHL. Although most studies report group differences in the ability to make use of TFS cues between people with normal hearing (NH) and those with SNHL, performance varies greatly within each group, despite similar audiometric configurations (Hopkins et al., 2008; Hopkins and Moore, 2010; Strelcyk and Dau, 2009).

The processing of pitch-related acoustic cues can be investigated using objective cortical auditory evoked potentials (CAEPs) that reflect differential neural encoding of stimulus acoustic cues. CAEPs elicited using brief stimuli (clicks, tone bursts) consist of three peaks (P1-N1-P2) that occur within 300 milliseconds (ms) after stimulus onset (Martin et al., 2008). N1 is a transient response evoked by short-term envelope change (Onishi & Davis, 1968). P2 is sensitive to attention and stimulus parameters such as intensity and pitch (Crowley and Colrain, 2004), as well as musical experience (Seppänen et al., 2012). CAEPs elicited using complex long-duration stimuli with acoustic changes within the stimulus have multiple N1-P2 complexes evoked by the stimulus onset, the acoustic change, and the stimulus offset (Digeser et al., 2009; Martin et al., 2008; Ostroff et al., 1998; Sharma et al., 2000). Cortical responses encoding the change in an ongoing stimulus have been described as the acoustic change complex (Martin & Boothroyd, 1999). Acoustic change complexes have been recorded in response to both speech and non-speech sounds (Martin et al., 1999; Ostroff et al., 1998), as well as to acoustic changes within a speech sound such as formant frequency transition within a vowel (Martin & Boothroyd, 2000). The acoustic change complex shows distinct neural patterns in response to changing speech syllables in adults using hearing aids and cochlear implants (Friesen & Tremblay, 2006; Tremblay et al., 2006). The acoustic change complex was used in the current study to show differential neural encoding of complex acoustic cues important for pitch processing. Establishing a link between electrophysiological and behavioural TFS measures may help future research determine optimal hearing aid settings for robust speech perception in noise. Moreover, it would be useful to determine pitch-related enhancements in cortical responses corresponding to specific stimulus acoustic cues.

Sensitivity to changes in pitch cues has been extensively studied using complex tones (Hopkins & Moore, 2007; Moore & Moore, 2003; Schouten et al., 1962). Complex tones resemble the sounds of vowels in normal speech and sounds produced by many musical instruments. Pitch extraction of a complex tone primarily depends on the harmonic resolvability and this in turn depends on the number in the harmonic sequence, *N*, rather than the absolute F0 (Houtsma & Smurzinski, 1990; Plack et al., 2005). Pitch discrimination is usually good when filtered complex tones contain only low-numbered harmonics, which may be resolved at the level of the cochlea, i.e. $N < 8$, due to access to both place (spectral) and TFS (temporal) cues. Complexes with only high-numbered harmonics (partially resolved), with *N* between 8 and 12 harmonics produce a weaker pitch percept which might be conveyed solely based on TFS information (Bernstein & Oxenham, 2003; Moore et al., 2006a). Hence, pitch perception depends on the salience of pitch cues. Most cochlear implants have only a small number of channels and thus TFS cues important for pitch perception are typically not successfully encoded by these instruments (Wilson & Dorman, 2008). On the other hand, although hearing aids restore audibility (ENV cues) and convey TFS cues, HI listeners cannot utilize TFS cues for pitch and music perception (Chasin & Russo, 2004). The current study aimed to increase understanding of behavioural pitch discrimination abilities in NH adults and adults with either mild or high frequency SNHL, using low- and high-numbered harmonic complex tones. Behavioural results were compared to the neural encoding of pitch cues measured using the acoustic change complex. This combined approach using behavioural and electrophysiological measures will help determine stimulus acoustic cues dominant for pitch processing at the level of cortex.

Materials and Methods

Participants

Ten young adults with NH aged 21 to 36 years (mean: 29 years, SD 4.6) and 9 adults with either mild or high frequency SNHL aged 20 – 55 years (mean: 37 years, SD 11.8) were recruited. Although there is a considerable variation in the age of participants, age effects on CAEPs are commonly reported when results are compared between young adults and people aged 60+ (Harris et al., 2009; Kim et al., 2012; Tremblay et al., 2004). Picton et al. (1984) who studied CAEPs across a broad age range from 20 to 79 years found no age effects for P1, N1, and P2 latencies and amplitudes. All NH adults were right handed, English speakers, with normal Type A tympanograms with present acoustic reflexes. Audiometric thresholds of the listeners with SNHL are shown in Table 4. All participants in the SNHL group were right handed, English speakers and had air-bone gaps of less than 15 dB and normal tympanograms. Audiograms for the NH and SNHL participants are shown in Figure 9. Written informed consent was obtained from all participants before testing. The study was approved by the University of Auckland Human Participants Ethics Committee.

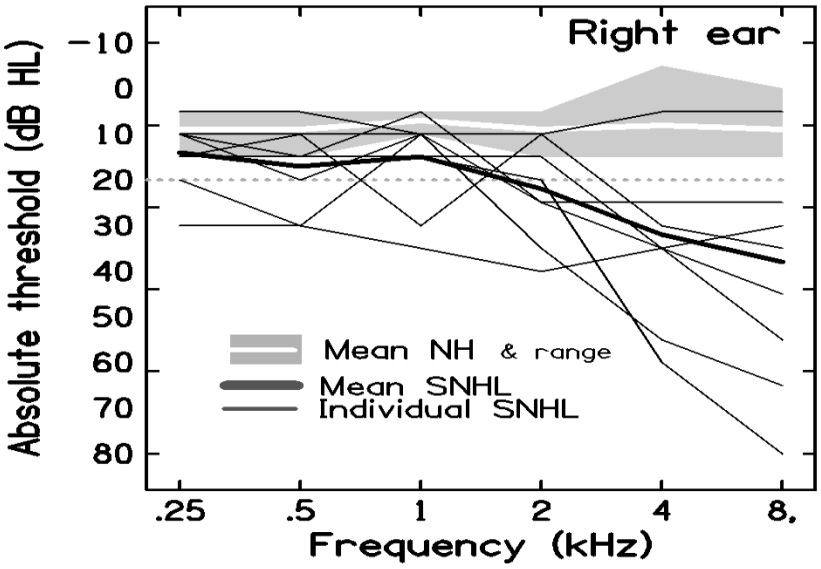


Figure 9. Audiogram for the right ears of the 10 NH and nine SNHL participants**.** The thin and thick black lines represent the individual and mean audiograms of the SNHL participants. The thick white lines and associated light-grey shaded areas represent the mean audiograms and ranges of audiometric thresholds for the NH participants, respectively. The dashed grey line indicates the audiometric inclusion criteria used in the present study.

Table 4. *Audiometric thresholds measured for the right ear for each SNHL participant*

	Frequency (kHz)					
Listener	0.25	0.5	1	2	4	8
SNHL1	15	10	10	35	55	65
SNHL 2	5	5	10	10	30	35
SNHL 3	10	15	15	20	60	80
SNHL 4	10	15	10	25	35	45
SNHL 5	15	15	15	15	35	30
SNHL 6	20	30	10	10	5	5
SNHL 7	10	15	5	25	25	25

SNHL 8	10	10	30	10	5	5
SNHL 9	30	30	35	45	35	55

Stimulus conditions

Processing of pitch differences were tested for two stimulus conditions with strong (N6) and weak pitch salience (N12). Stimuli consisted of bandpass filtered harmonic and frequency shifted (deltaF) complex tones. Pitch processing was separately investigated using both spectral excitation and TFS cues (N6 condition) and TFS cues alone (N12 condition). Here *N* is used to refer to the harmonic number corresponding to the centre of the bandpass filter through which all tones were passed. Spectrograms of the stimuli are shown in Figure 10. Values of the fundamental frequency (F0) and number of components in the passbands were 200 Hz and 3 for the N6 stimulus condition and 100 Hz and 5 for the N12 stimulus condition, respectively. The filter centre frequency was 1200 Hz for all stimuli. The lowest harmonic component within the passbands for the two conditions tested here was 1000 Hz. These stimulus parameters resulted in tone complexes with resolved (N6) and mostly unresolved (N12) components.

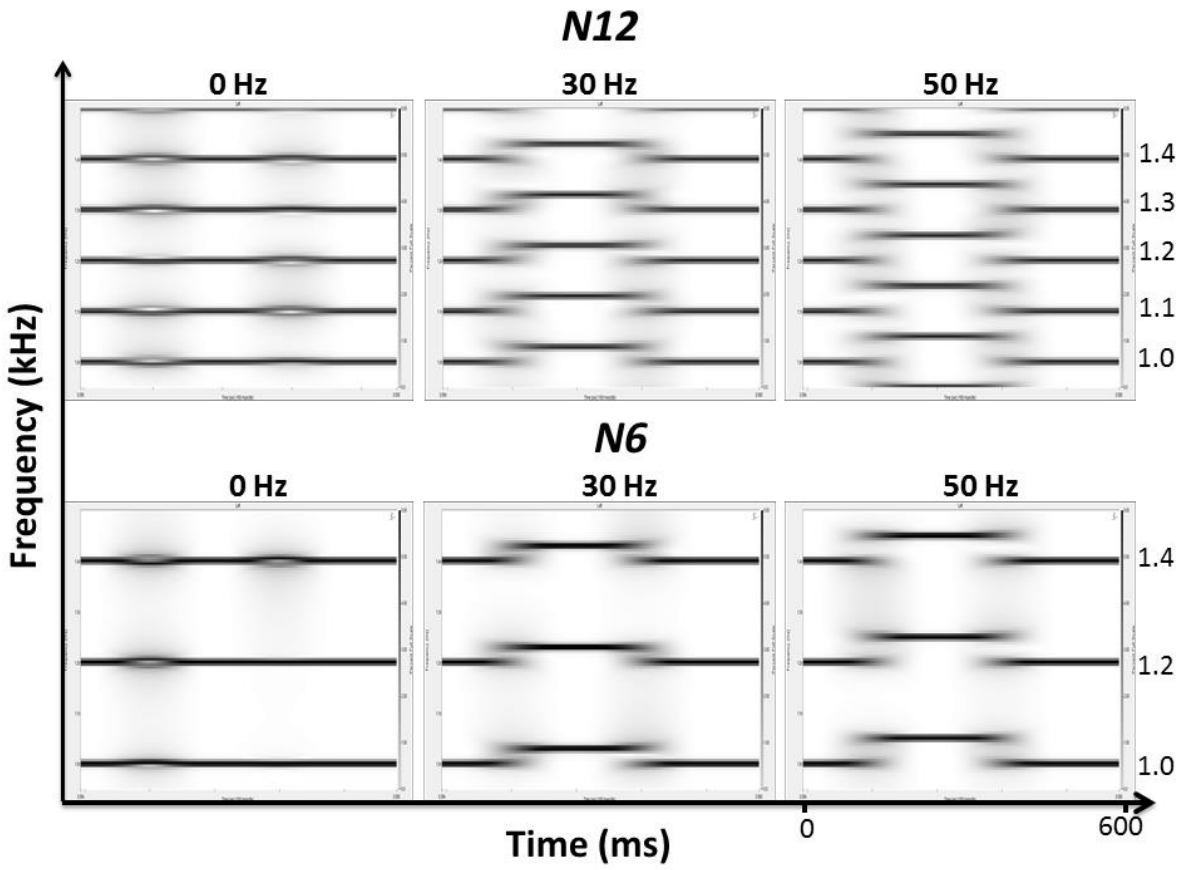


Figure 10. Spectrograms of the presented stimuli. Top portion shows the deltaF values for the N12 stimulus condition and bottom portion shows the deltaF values for the N6 stimulus condition.

Behavioural sensitivity to pitch cues

Pitch discrimination was assessed behaviourally using the TFS1 test downloaded from http://hearing.psychol.cam.ac.uk (Moore et al., 2009a). This test involves discrimination of a harmonic complex tone (A) from an inharmonic complex tone (B) in which the harmonics are shifted upwards by the same amount in Hz, deltaF. The TFS1 test was a two-interval forced-choice task with feedback. Each interval contained four bursts of sound in either AAAA or ABAB sequences. On each trial, two consecutive intervals were presented, separated by 300 ms. Each interval contained four consecutive 200 ms tones,

separated by 100 ms. Both ‘A’ and ‘B’ tone complexes had an envelope repetition rate with equal F0, but differed in their TFS due to the deltaF. The starting phases of the components in each and every tone were random and a new random selection was used for every presentation. This prevented envelope shape from being used as a discrimination cue. A background threshold-equalising-noise (TEN), extending from 200 to 16,000 Hz, was used to mask combination tones (Moore et al., 2000). DeltaF was the manipulated variable, initially set to 0.5F0. DeltaF varied from trial to trial according to a 2-down 1-up procedure, to estimate the value of deltaF producing 70.7% correct responses (Levitt, 1971). The value of deltaF was changed by a factor of 1.953 until the first reversal, then by a factor of 1.5625 until the second reversal, and by a factor of 1.25 thereafter (Moore et al., 2009a). After eight reversals, the run was terminated and the threshold was estimated as the geometric mean of deltaF values at the last six reversals. The maximum possible shift is 0.5F0 and if this was reached three times during a run, the shift was fixed at 0.5F0 and 40 more trials were presented; in this case the procedure changed to a non-adaptive procedure and a score was given as the proportion correct. A score of 25 or below was regarded as chance (Sek & Moore, 2012). The TFS1 test was installed on a DELL Latitude 6420 laptop and stimuli were presented via Sennheiser HD 25 1-ii headphones in a double-walled sound-attenuating booth. The stimulus was presented monaurally to the right ear at 65 dB SPL for the NH listeners. For listeners with SNHL the presentation level was 65 dB SPL or greater to ensure that the level was at least 30 dB SL (sensation level re: 1000 Hz pure tone threshold). The developers of the TFS1 test note that results do not depend critically on level, provided that the SL is at least 30 dB, at the centre frequency being tested (Moore et al., 2009a). The background TEN was presented at + 15 dB SNR, started 300 ms before the first tone in the interval and ended 300 ms after the last tone in the second interval. All participants completed a

practice run to ensure they understood the task. This involved discrimination of simple sine waves in a two-interval forced-choice task, with each interval containing four bursts of sound. Following the practice run, one run was completed for each stimulus condition, with the N6 condition completed first.

Electrophysiology

Stimuli

N6 and N12 evoked potential stimuli were generated using MATLAB 2012b (Mathworks, Inc.) replicating the stimulus parameters described in section 2.2. In order to elicit an acoustic change complex an 'ABA' stimulus sequence was used in contrast to the AAAA/ABAB sequences used in the behavioural TFS1 test.

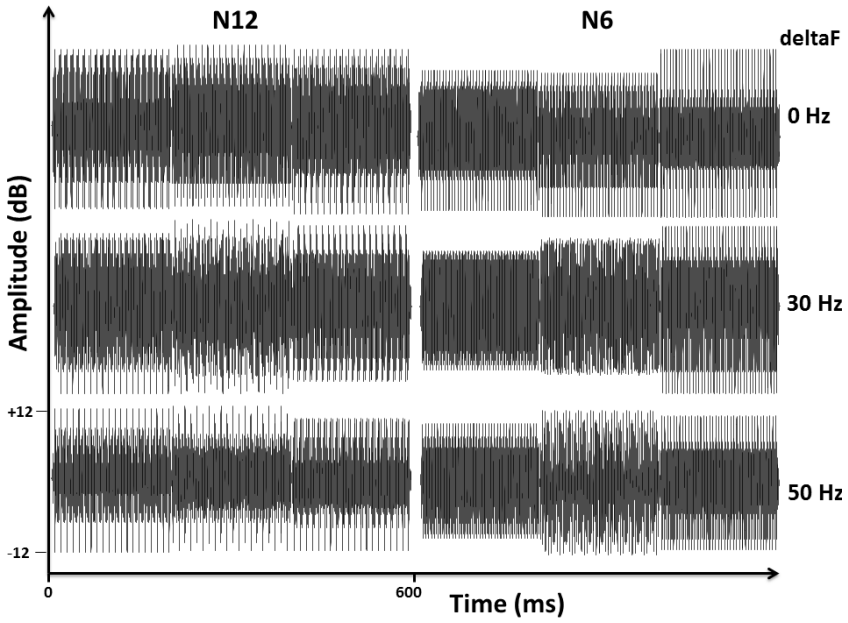


Figure 11. Time waveforms of the 600 ms ABA stimulus triplets for the two stimulus conditions are shown: N12 and its deltaF values (left), and N6 and its deltaF values (right)

As shown in Figure 11, stimuli generated for evoked potential recordings were a sequence of three bandpass filtered complex tones ('ABA' stimulus triplets). A total of six stimulus

triplets, consisting of three deltaF values (0, 30, & 50 Hz), for each stimulus condition (N6, N12), were generated. Within each stimulus condition, the A-tone complex was always harmonic and the B-tone complex varied depending on the deltaF value. The ABA sequence with deltaF = 0 Hz served as a control stimulus with no pitch-based acoustic change. Both 'A' and 'B' tone complexes had an envelope repetition rate with equal F0, but they differed in their TFS due to the deltaF. Onset phases of the components in each tone were randomised from 0 to 360^0. A new randomization was selected for every stimulus presentation. Each stimulus triplet was 600 ms long with no gaps between the A-B-A stimuli; each tone complex (A or B) within the stimulus triplet was 200 ms long. This allowed the stimulus characteristics to be similar to the behavioural TFS1 test. However, the brief duration (200 ms) before the transition may have resulted in overlapping cortical responses and could have affected the overall morphology of the ACC in this study. A Hanning window of 2 ms was used to shape the onset and offset of each complex tone to ensure a smooth transition and avoid audible clicks at A-B-A transitions. There were phase discontinuities and a small amount of spectral splatter at the transitions within the stimulus triplets and hence the deltaF = 0 Hz condition was used as a comparison condition to control for these effects. The magnitude of the splatter was computed in Praat software (www.praat.org), using a 10 ms Gaussian window. The splatter at the transition of each stimulus triplet was 6 dBV (0 Hz), 8 dBV (30 Hz), and 13 dBV (50 Hz) for the N6 stimulus condition and 10 dBV (0 Hz), 15 dBV (30 Hz), and 16 dBV (50 Hz) for the N12 stimulus condition. TEN was used to mask combination tones. All sound stimuli were presented to the right ear using an ER-3A 10 Ω insert earphone. Stimulus triplets were presented at 65 dB SPL for the NH participants and at a level that was at least 30 dB SL for the SNHL participants, with background noise at +15 dB SNR. Calibration was based on overall RMS level. Two blocks of 150 trials for each stimulus

triplet were presented to the participant. The sequence of the six stimulus triplets was randomized across each block and participants. Interstimulus interval (ISI) was set to 1150 ms.

Recording and pre-processing of electrophysiological data

All testing was performed in a double-walled sound-attenuating booth. Participants were seated comfortably on a reclining chair while watching a close captioned DVD of their choice (Lavoie et al., 2008).The Neuroscan SCAN^{TM} (version 4.5) software and Synamps2 was used for recording electrophysiological data. Cortical responses elicited by the stimuli were obtained using three EEG channels with 10 mm silver-silver chloride disc electrodes placed at Cz and Fz, referenced to the ipsilateral mastoid (M2). These electrode sites were selected because of the robust nature of auditory evoked potentials at the midline location. The ground electrode was located on the forehead and eye blink activity was monitored using electrodes placed above and below the right eye. Electrode impedances were kept under 3 kΩ. EEG was amplified with a gain of 1000 and sampled at the rate of 1000 Hz. EEG data were pre-processed using Neuroscan's built-in functions. Trials with eye blink artefacts were corrected offline using the ocular artefact rejection function in Neuroscan software (Neuroscan, Inc. 2007): the vertical electro-oculogram (VEOG) channel was scanned for the maximum eye movement potential. EOG deviations of more than 10% from the maximum were used as indicators of blinks. A minimum of 20 blinks was required to estimate an average blink. The procedure discarded artefacts starting <400 ms before a previous artefact, to avoid double detection. From the average VEOG ocular artefact, transmission coefficients were computed for each EEG channel by estimating the covariance of the averaged potentials of the VEOG channel with the EEG channels. The contribution of the average blink from the VEOG

channel was then subtracted from all other channels on a point-by-point basis. EEG epochs with -100 ms pre-stimulus to 1500 ms post-stimulus time windows were extracted post hoc from the continuous file. Before averaging, responses were digitally filtered at 0.1 to 30 Hz. All recordings were baseline corrected (-100 to 0 ms) before averaging. Trials containing artefacts exceeding ± 75μV were rejected from averaging. The remaining sweeps were averaged for each stimulus triplet.

Data analysis and interpretation

Waves N1, P2, and acoustic change complexes were analysed at electrode sites Cz and Fz. N1 and P2 peak latencies were computed relative to the stimulus onset (0 ms) and peak amplitude relative to the baseline. Acceptable latency ranges were between 90 and 150 ms, and between 180 and 250 ms post stimulus onset for N1 and P2, respectively. Peak amplitudes were computed by locating the largest amplitude that is surrounded on both sides by smaller amplitudes within the latency window. Unlike the CAEPs which had robust amplitudes, the acoustic change complex was small relative to the noise floor in the recordings and hence window-based mean amplitudes were computed to improve SNR (Luck, 2005). For each stimulus triplet, the peak of the acoustic change complex was identified from the grand-averaged waveform and a time window was selected that included voltage points within +/- 25 ms surrounding this peak. Using this time window each participant's mean acoustic change complex voltage was calculated and used for statistical analysis. Separate statistical analyses were performed for the obligatory components (N1, P2) and acoustic change complex amplitudes. Two (group: NH vs. SNHL) × 2 (stimulus conditions: N6, N12) × 3 (stimulus triplets: deltaF) × 2 (electrode: Cz, Fz) mixed-model ANOVAs were used to find statistical differences for the evoked potentials.

Thresholds obtained from the behavioural TFS1 test for both groups were subjected to a 2 (group: NH vs. SNHL) x 2 (stimulus conditions: N6, N12) mixed-model ANOVA. Interaction effects were explored using one-way ANOVAs to examine each of the effects separately. Tests of simple effects were conducted using paired- and independent-samples t-tests. A significance level of 0.05 was used for statistical analyses. Greenhouse-Geisser corrections (Greenhouse & Geisser, 1959) were used when the assumption of sphericity was not met.

Results

Behavioural sensitivity to pitch cues

Table 5 shows the mean and individual thresholds of each subject group for the N6 and N12 stimulus conditions. Two of the NH participants and all nine of the HI participants were not able to discriminate the maximum deltaF (50 Hz) for the N12 stimulus condition. Moore et al., (2009a) described a procedure to estimate deltaF values based on the detectability index, *d*', calculation in conditions where the participants were unable to reach a threshold less than deltaF = 50 Hz in the adaptive task. Only two participants (NH9 and SNHL2, Table 5) who failed the adaptive task had a score of >26 correct responses out of 40 trials for the fixed delta F = 50 Hz value, indicating some ability to use TFS cues. The other participants failing the adaptive task performed at chance levels and hence, rather than estimating the delta F using the d-prime procedure for the statistical analysis, a fixed deltaF level of 55 Hz was assigned to all participants failing the adaptive task.

Table 5. Individual and mean TFS1 thresholds (deltaF, Hz) for the N6 and N12 stimulus conditions, for the NH and SNHL participants

	NH		SNHL	
Participants	**N6**	**N12**	**N6**	**N12***
1	5.2	9.7	9.7	19/40 (0.47)
2	6.5	5.8	12.0	30/40 (0.75)
3	10.5	17.0	9.4	19/40 (0.47)
4	8.3	22.9	29.3	20/40 (0.50)
5	8.3	5.8	24.7	23/40 (0.57)
6	6.9	14.1	13.0	19/40 (0.47)
7	5.7	13.6	8.1	25/40 (0.62)
8	11.6	23.6	12.9	25/40 (0.62)
9	17.1	28/40*****	41.0	15/40 (0.37)
10	3.7	17/40*****		
Mean	**8.38 (Hz)**	**22.25 (Hz)**	**17.7 (Hz)**	0**.54**
(SD)	**(3.8)**	**(18.3)**	**(11.3)**	**(11.2)**

Note: * number of correct responses out of 40

The mixed-model ANOVA showed significant main effects for subject group [$F(1,17) = 30.8$, $p < 0.001$] and stimulus condition [$F(1,17) = 56.5$, $p < 0.001$]. Figure 12 shows that the N12 condition was more difficult for both groups; however for the HI group the difference in performance was much greater. There was also a significant interaction between subject group and stimulus condition [$F(1,17) = 11.8$, $p = 0.003$].

Post hoc t-tests were used to determine whether the mean thresholds differed between subject groups within each stimulus condition. For the N6 stimulus condition, the NH

listeners typically performed somewhat better than the SNHL listeners and the difference in mean scores was statistically significant (t(17) = -2.47, p = 0.024). For the N12 stimulus condition, all of the participants in the SNHL group failed to do the task and hence overall performance was poorer than that of the NH group (Figure 12).

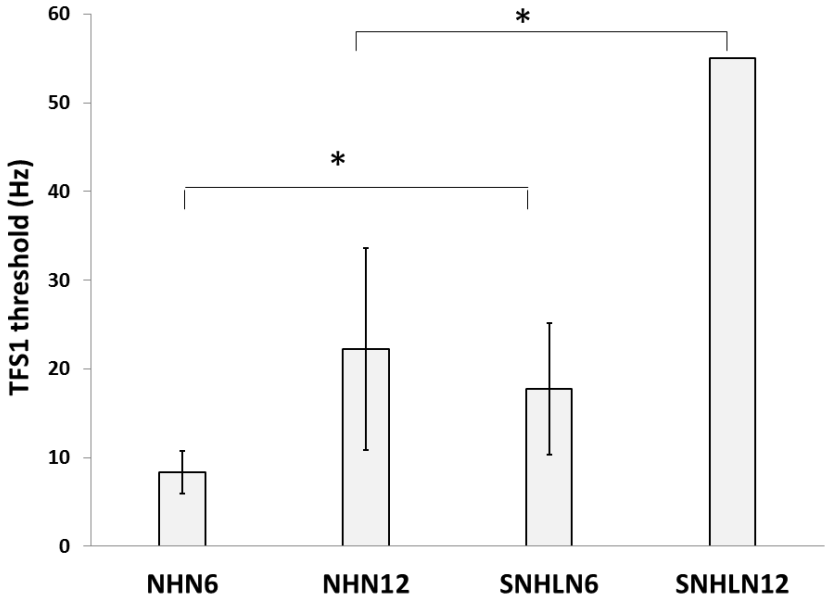


Figure 12. Mean TFS1 thresholds of each subject group for the N6 and N12 stimulus conditions. Error bars show the 95% confidence interval of the mean.

***= p < .05**

Electrophysiology

Electrophysiological recordings were used to investigate differences in neural encoding of pitch cues between the NH and SNHL groups and to explore pitch-related effects on the auditory evoked potentials. We hypothesised that group differences (NH vs. SNHL) would mainly be seen in the acoustic change complexes as the stimuli were all very detectable and hence should have generated robust CAEPs in both groups. We anticipated that pitch-related stimulus acoustic differences would primarily affect the P2 (Crowley et al., 2004) component of the CAEPs and the amplitudes of the acoustic change complex

(Martin et al., 1999). Results of the mixed-model ANOVA, conducted separately for the obligatory CAEPs and the acoustic change complexes are shown in Table 3.

Obligatory CAEPs

Though no main effect of group was found for any of the CAEP components, there was a significant interaction between stimulus condition and group for N1 latency (Table 6). Overall N1 latency increased for the NH group for the N12 (144.5 ms) compared to the N6 (137.8 ms) condition but was essentially unchanged for the SNHL group (138.7 vs. 137.7 ms). To further explore the interaction between stimulus condition and group, response latencies were averaged across deltaF and electrodes and post hoc independent t-tests were conducted to comparing N1 latencies between NH and SNHL groups for the N6 and N12 stimulus conditions. These comparisons revealed no significant differences between groups for the N6 ($t(17) = -0.1$, $p = 0.868$) or N12 ($t(17) = 1.1$, $p = 0.252$) stimulus conditions.

The main effect of stimulus condition was significant only for P2 latency and amplitude ($p < 0.05$; see Table 6). P2 responses averaged across electrodes and groups were earlier and larger for the stimulus condition N6 [234 ms (SE:4.1), 1.0 µV (SE: 0.28)] compared to N12 [243 ms (SE: 4.6), .04 µV (SE: 0.23)] (Figure 13). Overall, stimulus condition effects on CAEPs were mainly seen for P2.

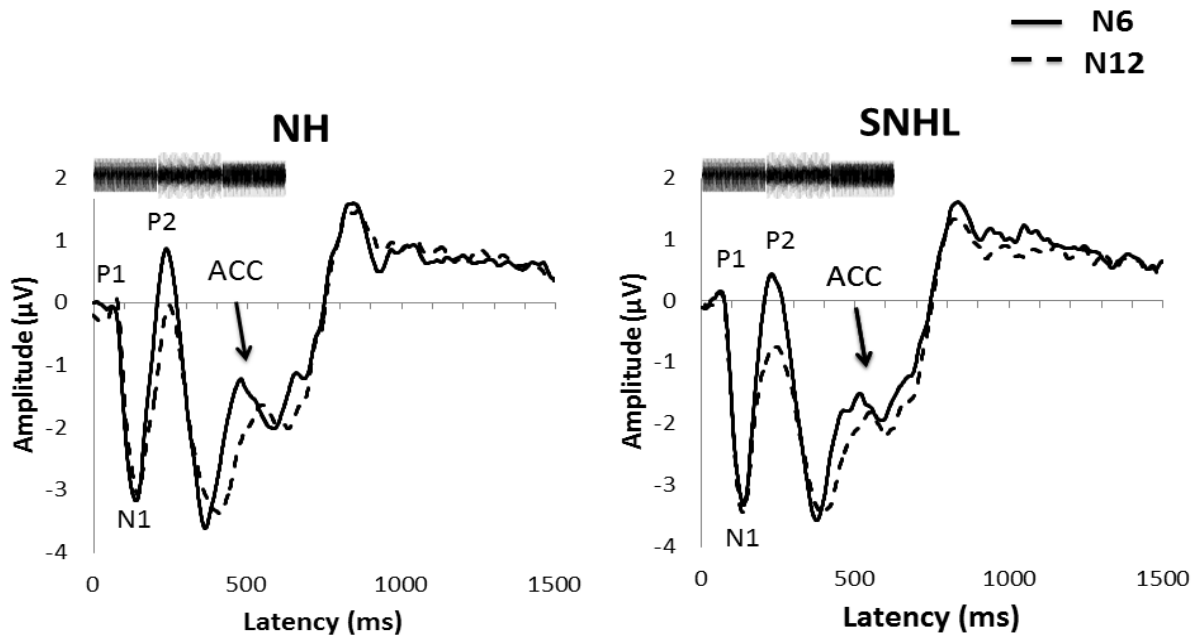


Figure 13. Grand mean Cz waveforms for the adults with NH (n = 10) and SNHL (n = 9) elicited in response to the N6 (solid line) and N12 (dashed line) stimulus conditions, averaged across the three stimulus triplets. Arrow marks approximate N1, P2 peaks corresponding to the stimulus onset, and acoustic change complex. Overlaid time waveform of the stimulus demonstrates the correspondence to the ERP waveform.

With regards to differences across electrodes, although there were no significant main effects, a significant two-way interaction was found for P2 latency, between stimulus condition and electrode (Table 6)**.** Post hoc comparisons indicated that differences between stimulus conditions (N6 vs. N12) were evident at both Cz [$t(18) = 4.0$, $p = 0.001$] and Fz [$t(18) = 2.9$, $p = 0.010$], but the N6 vs. N12 difference was greater at Cz. There was also a three-way interaction between stimulus condition, electrode, and group for P2 amplitude (Table 6). In order to explore the effects of electrode site on differences across stimulus conditions, a 2 x 2 ANOVA was conducted separately for each group. This analysis revealed that only Cz showed significant differences across stimulus

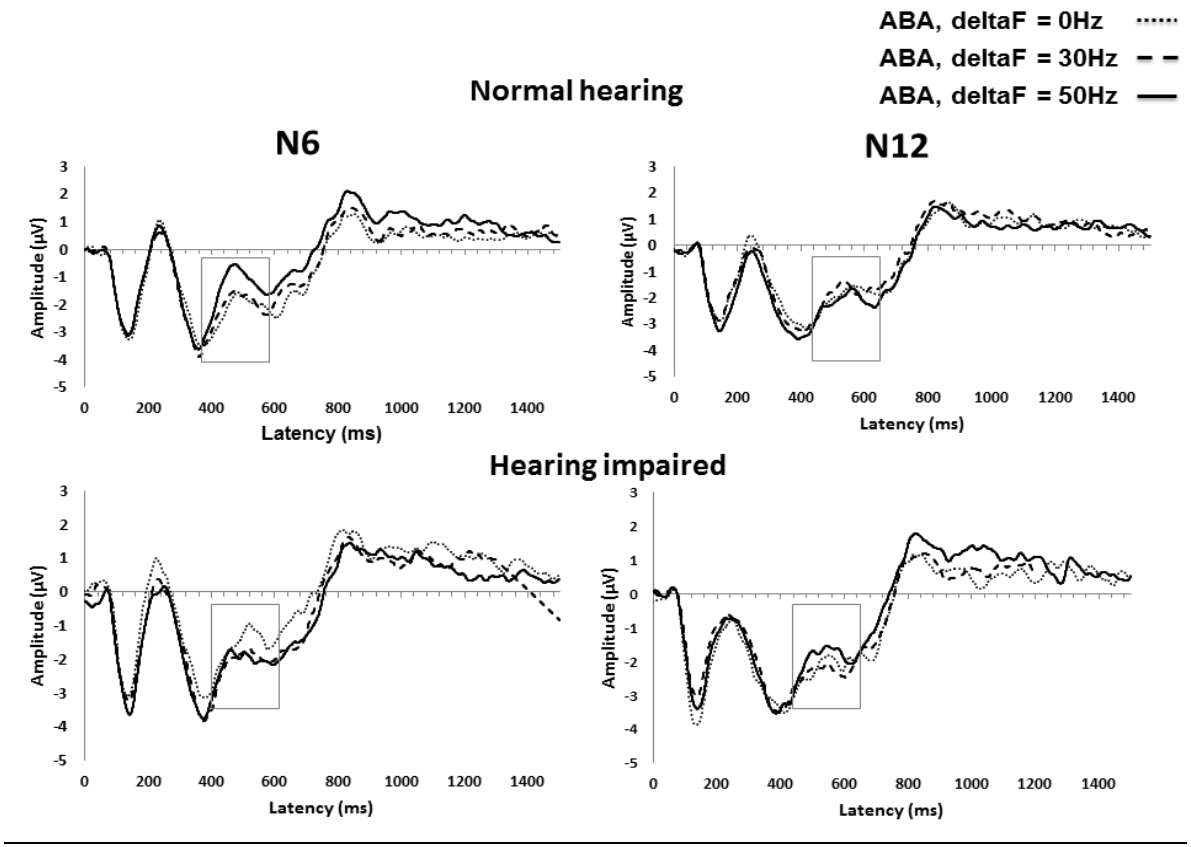
conditions and this was present only for the NH group ($F(1,19) = 4.7$, $p = 0.042$). Overall, recordings at Cz were more affected by changes in stimulus condition than Fz.

Acoustic change complex

Figure 14 shows the average amplitudes of the acoustic change complexes to the various deltaF values for the two stimulus conditions in the NH and SNHL groups. As was observed for P2, stimulus conditions also had a significant effect on the acoustic change complex (see Table 6). The acoustic change complex evoked by the stimulus condition N6 was more robust [-1.5 µV (SE: 0.17)] than that evoked by N12 [-1.8 µV (SE: 0.17)]. Although there were no significant main effects of group and deltaF, an interaction effect was found between stimulus condition, deltaF, and group (Table 6). Figure 15 illustrates the interaction between stimulus condition and deltaF values separately for the NH and SNHL groups. From this figure it is evident that for the N6 stimulus condition, the NH group showed a monotonic increase in acoustic change complex amplitude with increase in deltaF from 0 to 50 Hz ($0 < 30 < 50$ Hz). Mean acoustic change complex amplitudes for each stimulus triplet for the NH group were 0 Hz: -1.8 µV (SE 0.27), 30 Hz: -1.6 µV (SE 0.31), and 50 Hz: -0.7 µV (SE 0.31). Thus, as expected robustness of the acoustic change complex increased with increasing pitch shift in the NH group. This pattern was not observed for the N12 condition for the NH group and was not evident for either condition for the SNHL group. To further explore the three-way interaction between stimulus condition, deltaF, and group, a 3 x 2 ANOVA was conducted separately for each stimulus condition. This analysis showed that a group difference was present with significantly higher acoustic change complex amplitude only for the 50 Hz deltaF, N6

condition ($F(1,18) = 7.7$, $p = 0.013$). No other significant differences were observed between subject groups.

Figure 14. Grand mean Cz waveforms are displayed as a function of deltaF stimulus triplets (N6 stimulus condition), for the participants with NH and SNHL. The acoustic change complex is indicated by the grey inset box.



A main effect of recording electrode on acoustic change complex amplitudes was also found (Table 6). Mean acoustic change complex amplitude at Cz, averaged across stimuli was more positive [-1.6 µV, SE 0.17] than at Fz [-1.8 µV (SE: 0.17)]. The analysis also revealed a four-way interaction between stimulus condition, deltaF, group, and electrode (Table 6). Figure 15 illustrates this four-way interaction. The pattern of ACC amplitudes differed, as already noted, between groups for one combination of stimulus condition and deltaF (N6, deltaF = 50 Hz), however, the effect of deltaF and stimulus condition varied across electrode and group. For the SNHL group the effect of deltaF differed between electrodes for the N6 but not the N12 condition. For the NH group results were consistent across electrode.

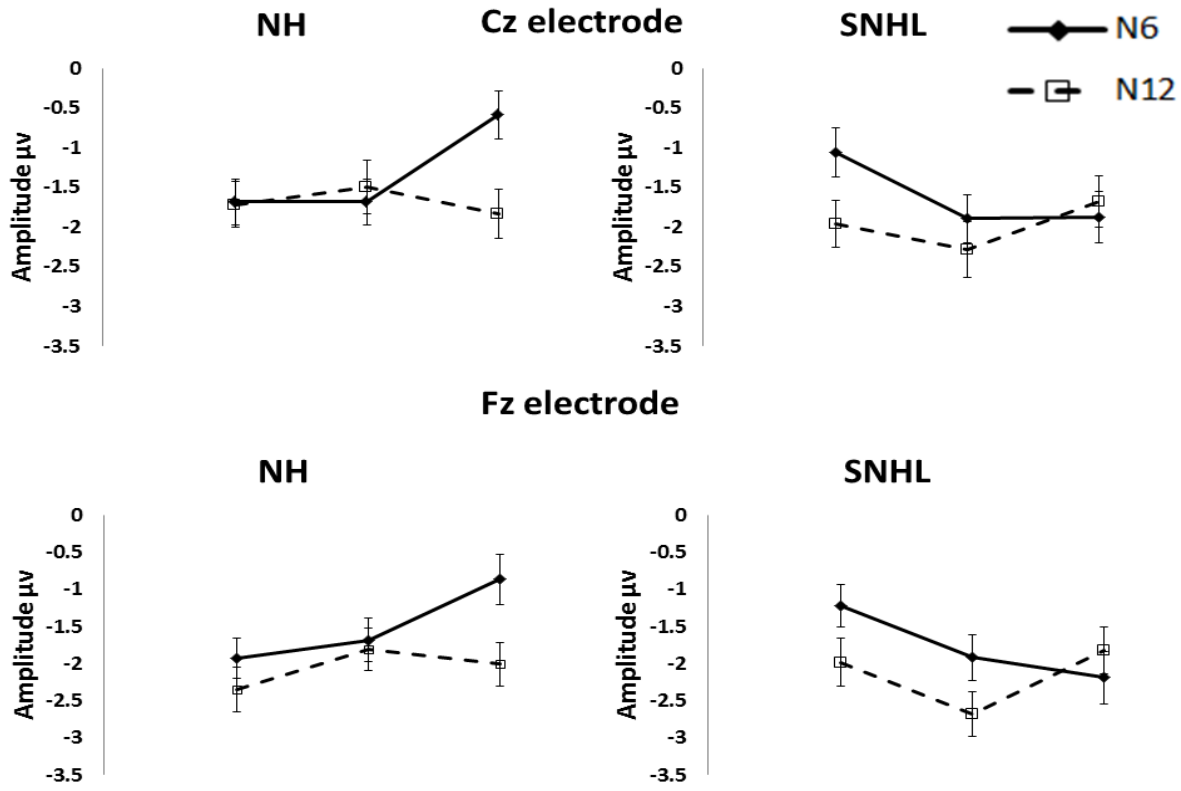


Figure 15. Mean acoustic change complex amplitudes (Cz & Fz) for the N6 and N12 stimulus conditions plotted across stimulus triplets are shown for the NH and SNHL groups.

Table 6. Results of mixed-model ANOVA of N1, P2, and acoustic change complex for all conditions.

	N1 Latency		**N1 Amplitude**	
	F(df)	***p***	**F(df)**	***p***
Group	0.298(1,17)	0.592	0.012(1,17)	0.915
Stimulus condition	2.794(1,17)	0.113	2.659(1,17)	0.121
Stimulus condition * Group	4.866(1,17)	0.041	2.226(1,17)	0.154
deltaF	1.031(2,34)	0.368	2.905(2,34)	0.068
deltaF * Group	0.639(2,34)	0.534	0.555(2,34)	0.579
Electrode	2.608(1,17)	0.125	0.019(1,17)	0.891
Electrode * Group	0.033(1,17)	0.858	0.966(1,17)	0.339
Stimulus condition * deltaF	0.664(2,34)	0.521	0.559(2,34)	0.577
Stimulus condition * deltaF * Group	2.358(2,34)	0.110	1.191(2,34)	0.316
Stimulus condition * electrode	0.287(1,17)	0.599	2.802(1,17)	0.112
Stimulus condition * electrode * Group	2.095(1,17)	0.166	0.333(1,17)	0.571
deltaF * electrode	1.649(2,34)	0.207	1.698(2,34)	0.198
deltaF * electrode * Group	3.055(2,34)	0.060	0.496(2,34)	0.613
Stimulus condition * deltaF * electrode	.375(2,34)	0.690	0.890(2,34)	0.420
Stimulus condition * deltaF * electrode * Group	.143(2,34)	0.867	0.901 (2,34)	0.416
	P2 Latency		**P2 Amplitude**	
	F(df)	***p***	**F(df)**	***p***
Group	.430(1,17)	0.521	0.137(1,17)	0.716
Stimulus condition	11.871(1,17)	0.003	34.138(1,17)	0.001
Stimulus condition * Group	0.404(1,17)	0.534	0.020(1,17)	0.890
deltaF	0.006(2,34)	0.994	0.738(2,34)	0.486
deltaF * Group	1.212(2,34)	0.310	0.089(2,34)	915
Electrode	0.263(1,17)	0.615	0.059(1,17)	0.811

Electrode * Group	2.498(1,17)	0.132	1.194(1,17)	0.290
Stimulus condition * deltaF	2.004(2,34)	0.150	0.634(2,34)	0.537
Stimulus condition * deltaF * Group	1.609(2,34)	0.215	1.830(2,34)	0.176
Stimulus condition * electrode	5.360(1,17)	0.033	0.101(1,17)	0.755
Stimulus condition * electrode * Group	.426(1,17)	0.523	7.026(1,17)	0.017
deltaF * electrode	.369(2,34)	0.694	0.635(2,34)	0.536
deltaF * electrode * Group	.721(2,34)	0.494	1.174(2,34)	0.321
Stimulus condition * deltaF * electrode	1.506(2,34)	0.690	2.106(2,34)	0.137
Stimulus condition * deltaF * electrode * Group	.545 (2,34)	0.585	0.879(2,34)	0.421
	ACC amplitude			
	F(df)		***p***	
Group	0.869(1,17)		0.364	
Stimulus condition	9.966(1,17)		0.006	
Stimulus condition * Group	0.046(1,17)		0.834	
deltaF	1.588(2,34)		0.219	
deltaF * Group	2.821(2,34)		0.074	
Electrode	17.057(1,17)		0.001	
Electrode * Group	1.029(1,17)		0.057	
Stimulus condition * deltaF	0.102(2,34)		0.903	
Stimulus condition * deltaF * Group	5.329(2,34)		0.010	
Stimulus condition * electrode	0.512(1,17)		0.484	
Stimulus condition * electrode * Group	2.024(1,17)		0.173	
deltaF * electrode	0.547(2,34)		0.584	
deltaF * electrode * Group	0.432(2,34)		0.653	
Stimulus condition * deltaF * electrode	6.731(2,34)		0.003	
Stimulus condition * deltaF * electrode * Group	4.712(2,34)		0.016	

Discussion

The current study showed differences in pitch discrimination abilities for bandpass-filtered harmonic (A) and inharmonic (B) tone complexes, containing resolved (N6) and mostly unresolved (N12) components, for NH and SNHL participants. Overall results indicate that listeners in the SNHL group showed poorer pitch processing abilities than the NH group for all stimulus conditions. Furthermore, perceptual processing abilities and neural encoding of pitch information depended on the stimulus condition (pitch salience). The P2 component of the CAEPs reflected pitch salience; across both groups P2 was smaller and later for the weak pitch (N12) stimulus condition. Acoustic change complexes were equally sensitive to the stimulus conditions and to the frequency shift in the stimulus triplets (deltaF 30 and 50 Hz) for the N6 stimulus condition only, for NH participants. For the HI group the frequency shift in the stimulus triplets did not produce a consistent acoustic change complex response.

Behavioural sensitivity to pitch cues

Discrimination thresholds for the stimulus condition N6 were significantly better than for N12 for both subject groups. Thus the ability to extract pitch using shifts in excitation pattern and TFS cues are better for stimuli containing resolved components (N6). This is in agreement with previous reports describing TFS sensitivity (Bernstein et al., 2003; Houtsma et al., 1990; Moore et al., 2003). Peripheral pitch encoding of complex tones containing low-numbered harmonic components (e.g. N6 stimulus condition) involves two processes (Moore & Gockel., 2011). Firstly, it is presumed that the harmonic components are spatially resolved on the basilar membrane, and secondly neural firing patterns phase lock to the TFS peaks at the envelope maxima. This result in a clear pitch

percept and changes in deltaF are easier to discriminate (Houtsma et al., 1990; Shackleton & Carlyon, 1994). In contrast, tones with high-numbered harmonics (e.g. N12 stimulus condition) produce a weaker pitch percept and poorer deltaF detection thresholds (Brenstein et al., 2003; Houtsma et al., 1990; Moore et al., 2009b), consistent with the results of the current study for both NH and SNHL participants.

The NH group performed significantly better than the SNHL group for all stimulus conditions. For the N6 condition, the listeners with NH had an average discrimination threshold of 8.3 Hz (0.7% of centre frequency). This is consistent with the literature (1% or less) for tone complexes containing resolved harmonics (Moore et al., 2006a). This discrimination threshold was much lower than that obtained by listeners with SNHL (17.7 Hz, 1.5%). Poorer discrimination thresholds for the SNHL group, with only a slight hearing loss in the mid frequencies (0.5, 1, 2 kHz), could be explained by broader auditory filters and/or decrease in phase locking in the auditory nerve compared with listeners who have NH (Moore, 2008). While most listeners in the NH group were able to perform the TFS1 task for the N12 stimulus condition, most listeners in the SNHL group scored no better than chance (Table 5). Poor discrimination scores suggest that participants with SNHL could not perceive differences in TFS cues between A and B tone complexes. These results corroborate findings of previous studies showing lack of sensitivity to TFS cues in adults with SNHL compared to NH controls (Ardoint et al., 2010; Hopkins et al., 2008; Hopkins et al., 2010; Lorenzi et al., 2006). These earlier studies showed reduced TFS sensitivity in adults with mild to moderate SNHL; the current study showed similar effects in adults with lesser degrees of hearing loss.

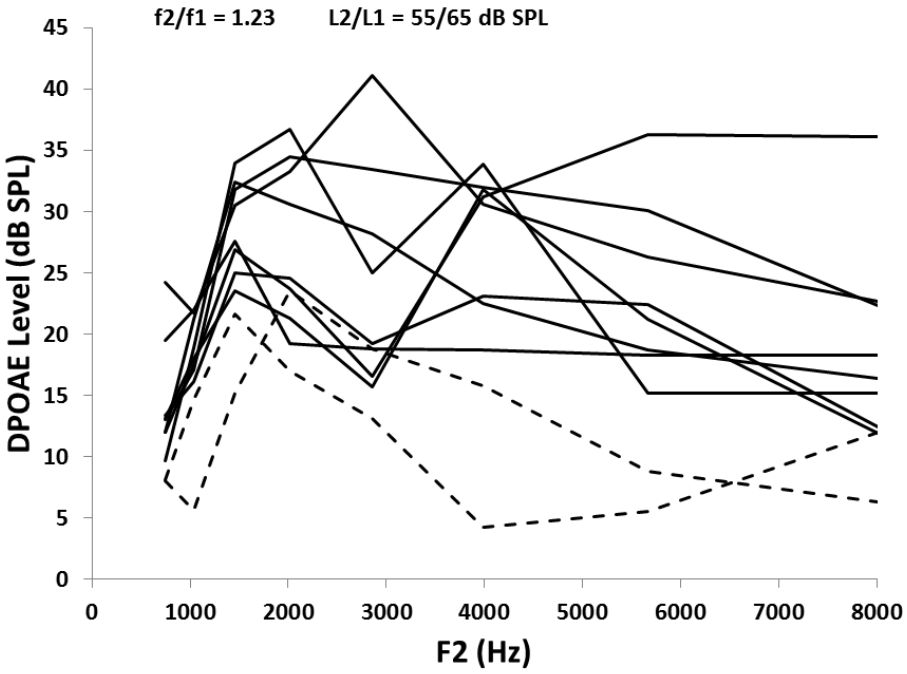


Figure 16. Distortion-product otoacoustic emission signal-to-noise ratio (DPOAE SNR) values for the right ear in the NH group (n=10). The dashed lines show the results of the two participants with lower DPOAE SNR values who had difficulty with the N12 behavioural discrimination task.

Previous studies have reported that TFS information is independent of the audiometric configuration (Hopkins et al., 2007; Strelcyk et al., 2009). Two of the participants aged 24 and 36 years with normal hearing thresholds (PTA < 15 dB HL) were not able to do the behavioural discrimination task for the N12 stimulus condition. The comparison of TFS1 thresholds for the N12 stimulus condition with audiometric thresholds assessed by means of Pearson correlations also showed no significant association between these variables ($r = 0.040$, $n = 10$, $p = 0.913$). Poor TFS1 performance could be indicative of a subclinical hearing loss that was not detected using conventional audiometry. To further explore this, we measured distortion-product otoacoustic emission (DPOAE) for all participants in the NH group. Interestingly, the two participants in the NH group who

could not do the N12 task showed lower-amplitude and/or absent DPOAEs at higher frequencies compared to other NH participants (Figure 16). Within the NH group, there was greater variation in DPOAE strength (Figure 16) and audiometric thresholds (Figure 9) at the higher frequencies. Moore (2007) proposed that limited ability to use TFS information for SNHL listeners relates to lower-amplitude OAEs. In the two NH participants described here, a sub-clinical hearing loss as indicated by the DPOAEs, may account for their poor performance on the N12 task. Thus adults with audiometrically-normal hearing can still experience TFS deficits. Füllgrabe (2013) similarly showed evidence of reduced sensitivity to TFS cues in young adults with clinically normal hearing. The TFS1 test may be a good screening tool for either mild or sub-clinical hearing impairment (Hietkamp et al., 2010).

Obligatory CAEPs

Obligatory CAEPs are sensory responses that depend on the physical characteristics of the stimulus (Martin et al., 2008). The presence of CAEPs indicated that the stimulus was detected at the auditory cortex (Hyde, 1997). In the current study CAEPs were present for each individual in each subject group and there were no substantive morphological differences in P1-N1-P2 across subject groups. This confirmed that stimuli were presented at a suprathreshold level, making them audible and equally detectable for all participants.

CAEPs have been used to show differential neural encoding of stimulus onset characteristics (Agung et al., 2006; Beukes et al., 2009; Digeser et al., 2009; Purdy et al., 2005; Whiting et al., 1999). N1-P2 CAEPs recorded using signal triplets (ABA) showed some onset-dependent changes when compared across stimulus conditions (N6 vs. N12).

Previous studies have shown that N1 morphology mainly reflects changes in stimulus envelope/rise time (Kodera et al., 1979; Onishi et al., 1968), but we gated stimuli on (and off) using a constant rise time, and onset phases of the components were selected randomly for every complex which could account for the lack of stimulus effects on N1.

Previous investigations have associated P2 with pitch processing and musical training (Istók et al. 2013; Tong et al., 2009); consistent with our finding that P2 differed between clear (N6) and weak (N12) pitch stimuli (see Figure 11). The N6 and N12 conditions tested here had the same centre frequency and fixed bandwidth but differed in their absolute F0 and harmonic components, resulting in pitch differences (Figure 10). P2 sources have been identified in the planum temporale and the lateral part of Heschl's gyrus (Crowley et al., 2004; Ross & Tremblay, 2009), and functional magnetic resonance imaging (fMRI) has shown enhanced activity in in these site's response to stimulus pitch differences (Barker et al., 2011; Schadwinkel & Gutschalk, 2010).

We found P2 to be significantly earlier and larger for the N6 stimulus condition than N12, which could be due to the better resolution of the components on N6 producing a clearer pitch percept, and inducing faster neural processing and stronger neural activation. Penagos et al. (2004) has also showed evidence of lower cortical activation for complex tones with unresolved than resolved components. Alternatively, P2 differences between N6 and N12 conditions could have also occurred because F0 differed across stimulus conditions. Although differences in P2 arising from contrasts in pitch salience (N6 vs. N12) and F0 were not separately studied here, the results suggest that P2 is reflective of pitch processing.

Acoustic change complex

Acoustic change complexes were recorded to examine the processing of pitch differences in two stimulus conditions with varying pitch shifts (deltaF), comparing adults with NH and SNHL. Overall results indicated that responses were larger and more discriminable at Cz than at Fz. This is consistent with previous studies showing larger ACC amplitudes near the vertex; at or lateral to Cz and FCz (Martin et al., 2010; Tremblay et al., 2006). However, amplitude differences seen across electrodes may result from the underlying volume conduction and inverse problems. As was seen for P2 amplitude and latency, acoustic change complexes were dependent on the salience of the pitch-evoking stimuli. Acoustic change complexes evoked using N6 stimuli were significantly more robust and produced clearer waveforms than those evoked by N12 stimuli (Figure 11). This suggests that neural encoding of pitch information at the auditory cortex is predominantly driven by the presence of resolved harmonic components. This is supported by a recent fMRI study that showed stronger activation of cortical pitch-sensitive regions in response to spectrally resolved harmonic tones than to frequency-matched noise and unresolved harmonic tones (Norman-Haignere et al., 2013*).*

Acoustic change complexes evoked using deltaF = 0 Hz control stimulus

From Figure 17 it can be seen that acoustic change complexes were recorded even with the control stimulus (AAA) for both subject groups. This could reflect the phase discontinuities, the brief temporal gap due to ramping off and on of the stimuli at the transition points, and/or spectral splatter at the transition points within the stimulus triplets. However, this would not have led to artifactual results because the phase randomisation at each stimulus triplet transition meant that the amplitude discontinuity varied randomly, independent of the frequency shift size and was thus evenly distributed

across groups and conditions. Thus the group difference observed only for the deltaF = 50 Hz condition are likely to reflect differences in processing of pitch cues.

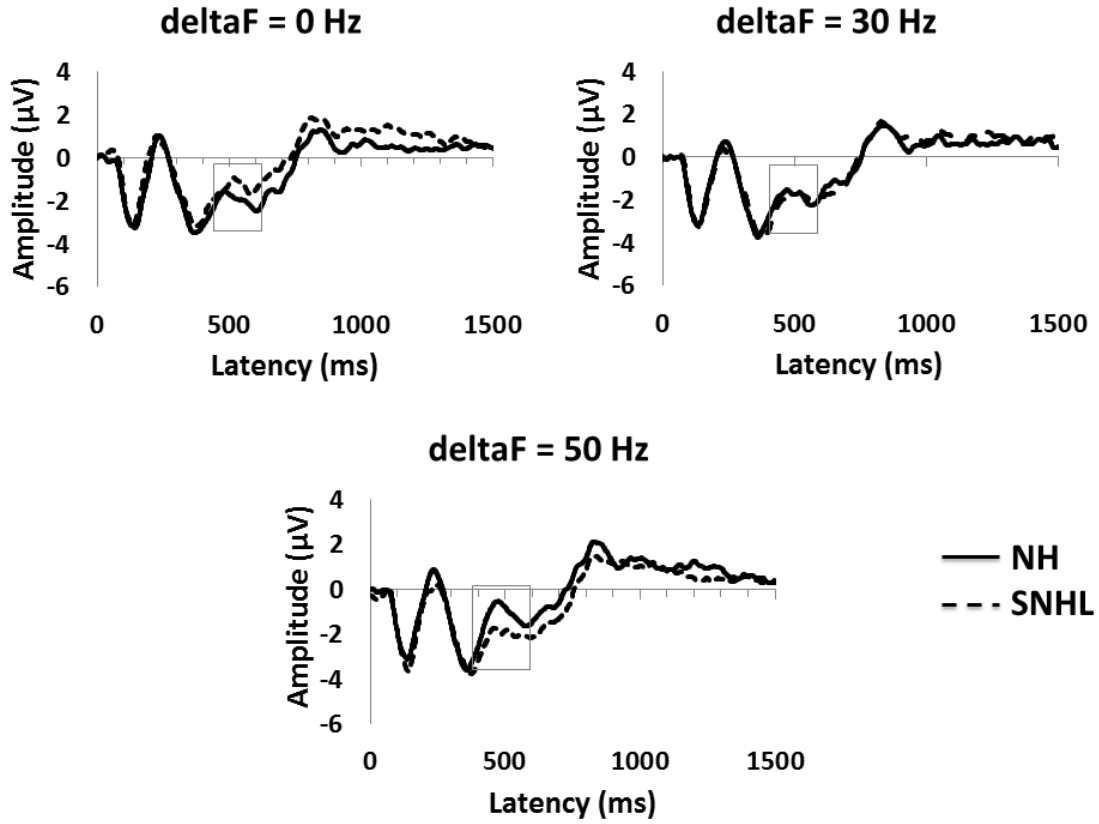


Figure 17. Grand mean Cz waveforms are displayed as a function of deltaF stimulus triplets (N6 stimulus condition), are overlaid for the participants with NH and SNHL. The acoustic change complex is indicated by the grey inset box.

Acoustic change complexes evoked by increasing deltaF

Acoustic change complexes demonstrated cortical sensitivity to pitch change only for the N6 stimulus condition with strong pitch salience. As expected, the NH group showed a monotonic increase in acoustic change complex amplitude with increasing pitch change (0 < 30 < 50 Hz). The 50 Hz change was perceptually discriminable but did not elicit an

acoustic change complex for the N12 condition for the NH participants. Results for the N12 stimulus condition are supported by the findings of neuroimaging studies showing weak and distributed pitch responses in the auditory cortex when using unresolved pitch-stimuli (Barker et al., 2011; Norman-Haignere et al., 2013*)*. Unlike the NH group, the SNHL group didn't show an increase in amplitude with increasing pitch shifts for either condition.

The finding of significantly larger acoustic change complex amplitudes in the NH group compared to the SNHL group (N6 stimulus condition, deltaF = 50 Hz) provides objective evidence for differences in the processing of complex acoustic cues between subject groups (see Figure 17). Although the acoustic change complexes were evoked using a stimulus condition that produced a clear pitch percept (N6 stimulus condition), the response amplitude differed between NH and SNHL participants. This aligns with the results from the behavioural measures in the current study. These findings suggest that pitch processing can be affected in frequencies where the absolute audiometric thresholds are only slightly affected.

Overall, the acoustic change complexes measured here did not show significant differences consistently for all perceptually discriminable pitch shifts in the NH (N12) and SNHL (N6 & N12) groups. Although the behavioural and electrophysiological measure did not produce parallel results, the absence of a significant difference does not indicate a lack of discriminability. Rather, discriminability is more likely for the deltaF shift that shows differential neural encoding which is essentially a prerequisite for successful perception. For example, on comparing the responses evoked using the largest

shift (deltaF = 50 Hz, N6 stimulus condition), the NH group showed significantly larger amplitude responses and better behavioural thresholds than the SNHL group. Additionally, evoked potentials were recorded using a passive listening paradigm as we were interested in obligatory encoding of pitch cues. It may be easier to demonstrate differential neural encoding to pitch change with an active oddball paradigm, in which participants are required to focus attention on the stimulus change.

Conclusion

The current study utilized behavioural and electrophysiological measures to show processing of complex acoustic cues important for pitch perception. Both behavioural performance and neural representation depended on stimulus pitch salience. Overall the study showed that sensitivity to TFS cues is adversely affected in individuals with hearing loss. Some limitations of this study include small sample size, the wide age range of participants and the lack of an age-matched control group, and Bonferroni adjustments were not used for multiple comparisons. This is the first time a relationship between stimulus triplets of varying deltaF and amplitude of acoustic change complexes has been described, and hence further research is required to clarify these findings. The combined electrophysiological and behavioural approach may be a useful for evaluating the benefit of training and amplification in individuals who experience difficulties understanding speech in noise.

Chapter 4

Auditory training of people with hearing loss: Effects on auditory event-related potentials in the absence of behavioral change

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This chapter includes content from the article "**Auditory training of people with hearing loss: Effects on auditory event-related potentials in the absence of behavioral change**" submitted to Clinical Neurophysiology, 2015.

Introduction

Acquired hearing loss is associated with changes in neural and perceptual processing (Dietrich et al., 2001; Syka, 2002; Wingfield and Peelle, 2015). Perceptual difficulties experienced by listeners with sensorineural hearing loss (SNHL) can be characterized as the *attenuation* associated with reduced thresholds and the *distortion* that can occur at supra-threshold levels (Plomp, 1986). For individuals with mild to moderate SNHL, hearing aids can restore the speech spectrum to a comfortable and intelligible level in quiet (Scollie et al., 2010). The distortional aspect of SNHL, i.e. difficulty understanding speech in noise, is not completely compensated for by a hearing aid, however (Kricos, 2006).

Deficits in auditory perceptual skills may be partially rehabilitated using auditory training in adults with mild to moderate SNHL (Bronus et al., 2011). Improvements in pure-tone frequency discrimination following adaptive training have been shown in adults with normal hearing using training approaches with a range of stimuli, sensory modalities, and training regimens (Amytay et al., 2005; Amitay et al., 2006; Molloy et al., 2012). In general, for a sinusoidal frequency discrimination task, most of the learning occurs early within a few sessions and is dependent on task difficulty (Molloy et al., 2012). A growing body of research has provided evidence for auditory-training-induced neurophysiological changes using auditory ERPs (Tremblay et al., 2001; Atienza et al., 2002; Tremblay and

Kraus, 2002; Bosnyak et al., 2004; Sheehan et al., 2005; Alain et al., 2007; Tremblay et al., 2009; Alain et al., 2010; Shtyrov et al., 2010; Tremblay et al., 2014; Barlow et al. (in press)). In most of these studies N1-P2 responses were passively recorded in adults with normal hearing following single or multiple short sessions of auditory training using speech stimuli. Changes in ERPs in people participating in auditory training using speech stimuli are mostly evident as an enhanced P2 response, with reduced effects on N1 (e.g. Barlow et al., (in press)).

Cortical evoked potentials reflect different levels of auditory processing, from detection of audible signals to neural processing of differences in signal acoustics, reflected in the P1-N1-P2 complex occurring at about 50-200 ms after stimulus onset, to conscious sound discrimination indicated by the P3 response occurring at about 300 ms after stimulus onset (Martin et al., 2008). In a passive ERP paradigm N1and P2 components mainly reflect effects of signal acoustics, however, these components are also influenced by top-down effects (Näätänen & Picton, 1987; Woldorff & Hillyard, 1991; Woods, 1995). Enhanced synchronized evoked responses reflecting training effects may be more evident during active listening paradigms. For example, rapid changes in the auditory neurons of trained ferrets were observed when they actively listened and performed a target-tone detection task as opposed to when they listened passively (Fritz, Shamma, Elhilali, & Klein, 2003). Reinke, He, Wang, and Alain (2003) showed that training-induced behavioral changes were associated with decreased N1 and P2 latencies and enhanced P2 amplitudes when ERPs were recorded using an active recording paradigm using speech stimuli in human participants with normal hearing.

Changes in ERPs could reflect a task repetition or stimulus exposure effect associated with the data acquisition process (Sheehan et al., 2005; Tremblay et al., 2010), and may not be due to auditory learning effects resulting from participation in auditory training. Changes in ERPs observed in auditory training studies may also reflect top-down processes such as auditory attention or memory (Näätänen et al., 1993; Amitay et al., 2006; Seppänen et al., 2012; Tremblay et al., 2010, 2014). Auditory learning is generally believed to be a result of rapid changes in sensory processing following training ('bottom-up' processes), however research also suggests the involvement top-down processes (Moore & Amitay, 2007). Enhanced top-down attentional skills following linguistic and nonlinguistic auditory training have been reported in clinical populations (Stevens et al., 2008; Murphy et al., 2015). For example, Stevens et al. used an active ERP paradigm along with behavioral measures to show enhanced selective attention skills in children with specific language impairment compared to controls after training.

Perception of pitch cues is important as it largely accounts for the ability to attend to a single speaker in the presence of multiple competing talkers (Oxenham, 2008). Cues to pitch include 'spectral' excitation cues that arise from the tonotopic arrangement in the basilar membrane and 'temporal' fine structure cues (TFS) that arise from the patterns of phase locking in the auditory nerve (Moore, 2008). The pitch of complex tones is primarily determined by the lower resolved harmonics that are coded both temporally and according to place in the output of the cochlea (Plomp, 1966). Moore and Sek (2009) developed the TFS1 test to determine sensitivity of an individual to specific pitch cues; the task involves a two- interval, two-alternative forced choice adaptive paradigm, where one interval contains four identical reference tones and the other contains two frequency shifted tones interleaved with two reference tones. A behavioral and

electrophysiological study using the TFS1 and ERPs showed that listeners with mild to moderate SNHL had poorer ability to use spectral and temporal fine structure (TFS) cues to discriminate the pitch of complex tones than listeners with normal hearing (Kuruvilla-Mathew et al., 2015).

Benefits from training for pitch (due to change in F0) and timbre (due to change in centre frequency) discrimination and melody contour identification have been observed in cochlear implant users (Galvin et al., 2007; Vandali et al., 2014). Vandali and colleagues trained experienced adult cochlear implant users over a period of two months (30 minutes each day) using adaptive training software (aTune) that employed complex tones varying in pitch or timbre. This software applied elements of gaming to encourage compliance and help participants to accomplish required tasks (Greitzer et al., 2007). The aTune software was hence used in the present study to train adults with either mild or high frequency SNHL over multiple training sessions (within a 9-day period).

The present study examined whether pitch and timbre-training resulted in improved perceptual discrimination and cortical representation of tonal complex contrasts driven by pitch cues (spectral and TFS). Possible training or other effects on ERPs were evaluated using *training-related*, tonal complex, contrasts varying in spectral and TFS cues and *non-related*, speech stimulus contrasts varying in voice onset time (VOT) cues, presented in an active oddball ERP paradigm. The purpose of this exploratory study was to determine whether any changes post-training might reflect improved sensitivity to pitch cues rather than some other mechanism. The effects of training were evaluated using behavioral discrimination and an active oddball ERP paradigm to explore

neurophysiological changes in adults with SNHL. We hypothesized one of two possible outcomes: 1) no significant change and 2) significant changes measured behaviorally and electrophysiologically. This was a single-group exploratory study where all participants received training.

Materials and Methods

2.1 Participants

Nine right-handed English-speaking adults (five females), with either mild or high frequency SNHL participated in this study (mean age = 42 years, SD = 15.5 age range = 24 – 62 years). The audiograms for the test (right) ears are shown in Figure 18. Participants had normal Type A tympanograms and no history of neurological disorders. No participants had formal musical training and none reported currently playing a musical instrument or taking part in any musical activity. Two participants were part-time hearing aid users (inexperienced ~4 months). Participants gave their informed consent in accordance with the University of Auckland Human Participants Ethics Committee requirements.

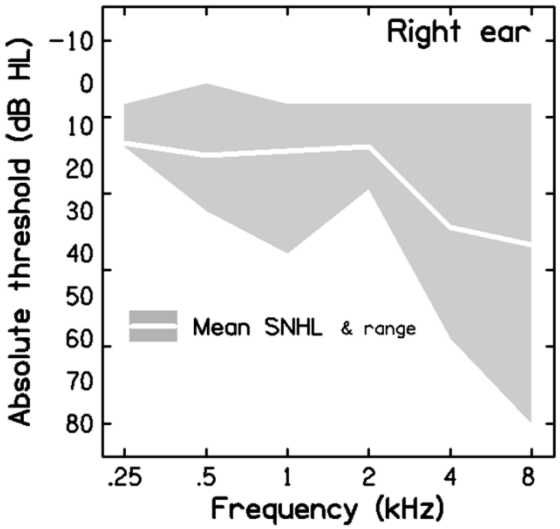

Figure 18. **Audiogram for the right ears of the nine SNHL participants.** The thick white lines and associated light-grey shaded areas represent the mean audiograms and ranges of audiometric thresholds for the participants, respectively.

2.2 Procedure

Each participant completed the experiment over a course of 16 days. On Day 1, pre-training cortical ERPs (oddball task) and behavioral measurements were obtained during the same session, lasting about 2.5 hours, inclusive of breaks. The oddball task required the participants to press a button whenever they heard the infrequent target signal. No performance feedback was given during this task. The participants subsequently trained at home using custom software (Vandali et al., 2015) installed on a laptop. The participant's task was to discriminate the pitch of complex tones which gradually grew closer in interval as participants advanced through the game. The training regimen for each participant lasted approximately 14 days and was self-paced. Participants were asked to train formally for 30 minutes for 9 days during the 14 day period. On day 16, participants

returned to undergo the same testing regime as in the pre-training session. Participants were reimbursed for their time with an additional bonus for progressing through the levels of the training program.

2.3 Electrophysiology

2.3.1 Stimuli and Maskers - Stimuli were presented monaurally to the right ear through Etymotic ER-2 insert earphones with a suprathreshold stimulus presentation level of 70 dB SPL and an overall root-mean-square (RMS) masker level of 55 dB SPL, producing a signal to noise ratio (SNR) of + 15 dB. Stimuli comprised speech and non-speech (complex tones) stimulus contrasts (Figure 19). Speech stimuli were natural consonant-vowel (CV) syllables (/di/, /ti/) produced by a native New Zealand female speaker recorded in a sound-attenuated room via a AKG HC 577 L omnidirectional headset microphone placed 3 cm from the lips of the speaker attached to an M-Audio MobilePre, using Adobe Audition version CS6 sound editing software, with a sampling rate of 44.1 kHz and 16 bit quantization rate. The CV syllables /di/ and /ti/ have the same place of articulation but differ in their voice onset times (VOT), the time lag between consonant release and the onset of voicing for the following vowel (*Abramson & Lisker, 1964*). Neural encoding of tonal pitch differences was studied using complex tones generated digitally in MATLAB 2012b (Mathworks, Inc.). Tonal stimuli were bandpass filtered harmonic and frequency shifted (Δf) inharmonic complex tones. The two tonal complexes (Δf = 0 and 50 Hz) had the same envelope rate equal to the fundamental frequency (F0), 200 Hz, but differed in their TFS. There were three components in the passband: 1000, 1200, 1400 Hz for Δf = 0 Hz and 1050, 1250, 1450 Hz for Δf = 50 Hz (see Figure 19). These stimulus parameters resulted in tone complexes with mainly low-numbered harmonics i.e. $N < 8$, which may be resolved at the level of the cochlea due to access to

both spectral (place) and TFS (temporal) cues (Plomp, 1966; Moore & Gockel, 2011). The filter center frequency was 1200 Hz for both tone complexes. Separate oddball sequences were generated for the tonal complex and speech stimuli for evoking the cortical ERPs. Total duration of the tonal and speech stimuli was 200 ms. The speech stimuli were shortened from the original stimuli (~ 700 ms) by removing segments of the steady-state part of the vowel..

The background masker was a threshold-equalising-noise (TEN), extending from 200 to 16,000 Hz (Moore et al., 2000). Background noise was continuously presented for the entire duration of the stimulus sequence presentation, and began before the presentation of the stimulus. TEN was used to mask combination tones for the tonal complexes (Moore & Sek, 2009). The same masking noise was used for the speech stimuli. A +15 dB SNR was used to study neural encoding at a favorable SNR.

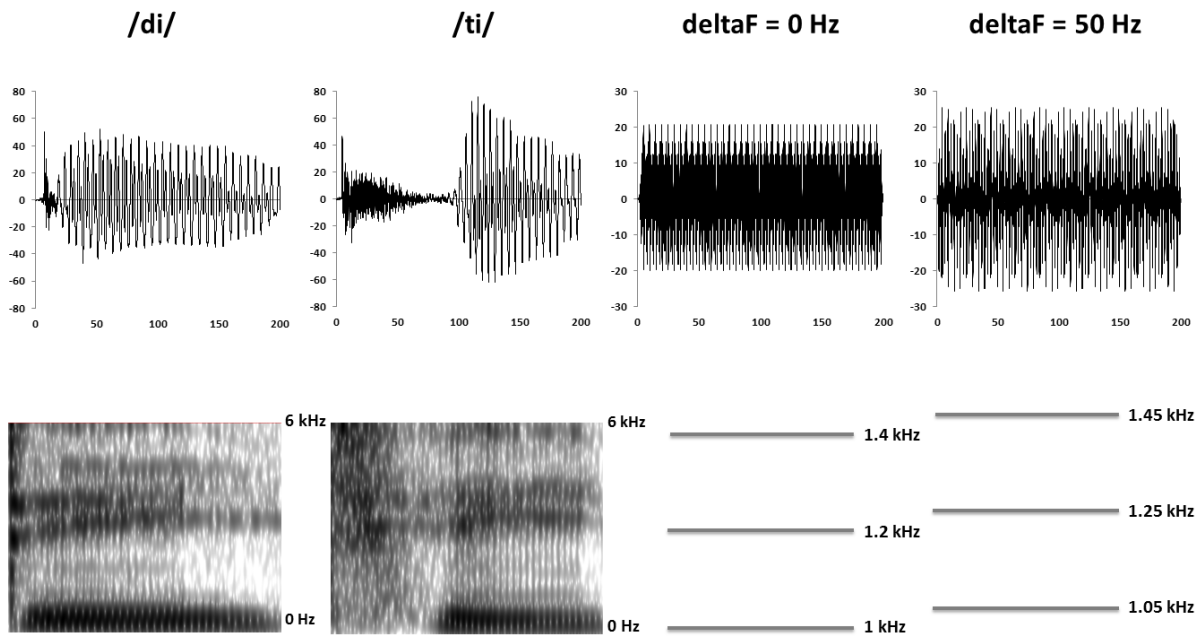

Figure 19. Acoustic waveforms of the stimulus types used here. Top portion shows the time waveforms for the speech stimuli (/di, ti/) and the tonal complexes (Δf = 0, 50 Hz). The bottom portion shows the respective spectrograms of the stimuli.

2.3.2 Cortical Event-Related Potential measurements

All testing was performed in a double-walled sound attenuating booth. During the recordings, participants were instructed to fixate on a small dot located ~2 m in front of them. The Neuroscan SCAN^{TM} (version 4.5) software and Synamps2 was used for recording electrophysiological data. ERP components, N1, P2, and P3 were recorded using the speech (/di/ - /ti/) and tonal complex (Δf = 0, 50 Hz) contrasts in separate oddball paradigms. Probability of presentation was 0.80 for standards and 0.20 for deviants. For each pair of stimulus contrasts (speech/tonal complex), the standard stimulus in one run became the deviant in the next; this ensured equal probability of each

and every stimulus (/di vs. ti/; (Δf = 0 vs. 50 Hz) to evoke both pre-attentive (N1, P2) and attentive (P3) ERP components. This allowed comparison of stimulus contrasts (speech vs. tone; /di vs. ti/; (Δf = 0 vs. 50 Hz)) across all ERP components (pre-attentive/attentive) and controlled for acoustic differences between the pairs of stimuli (Flip-Flop method, Sharma et al., 2004). For each contrast, two blocks of 250 tokens were presented for a total of 400 standard and 100 deviant stimulus presentations. An interstimulus interval of 900 ms (offset to onset) was used. Stimulus presentation was pseudorandomized so that a deviant was not the initial stimulus of a block and no two deviants were presented consecutively. The presentation order for stimulus contrasts was randomized across participants. Prior to recording, participants were given time to become familiar with the oddball discrimination task.

ERPs were recorded from surface electrodes placed along the midline of the scalp at frontal (Fz), central (Cz), and parietal (Pz) locations and the mastoids (reference electrodes, M1, M2). The ground electrode was located on the forehead and eye blink activity was monitored using electrodes placed over the right supraorbital ridge of the frontal bone and on the zygomatic bone under the right eye. Inter-electrode impedance was < 3 kΩ. N1 and P2 were analyzed at Cz and P3 at Pz electrode sites, where the response was largest, using an average mastoid reference. Evoked responses were amplified (gain = 1000), filtered (0.1 to 100 Hz, 6 dB/octave), and digitized at 20 kHz using a 1600 ms analysis time (including a 100 ms pre-stimulus period). Eye-blink artifacts were corrected offline, using Neuroscan SCAN 4.5 software (Neuroscan, Inc 2007). After eye-blink correction, trials containing artifacts exceeding +/− 75 microVolts were rejected. The ERP waveforms were epoched (-100 to 900 ms), baseline corrected, digitally filtered (30 Hz low pass filter 12 dB/octave) and averaged to obtain responses to

standards and deviants separately. Response amplitudes, relative to the baseline, and response latencies, relative to the signal onset were determined by agreement of two judges.

2.4 Behavioral measures

2.4.1 Reaction time (RT) and d-prime sensitivity (d')

The speed (RT) and accuracy (d') of the participants' response during the oddball discrimination task was recorded and analyzed. RT was measured from signal onset and the d' scores were calculated from measurements of hit and false alarm rates (Swets, 1973; Yanz, 1984). A hit was defined as a response occurring 100 to 900ms after the onset of an oddball stimulus. A false alarm was defined as any response that occurred outside this time window.

2.4.2 TFS1 test

Behavioral sensitivity to pitch was measured using the TFS1 test (Moore et al., 2009) which was set to detect changes in pitch of low-numbered harmonic complex tones. The task involved discrimination of a harmonic complex tone (H) from an inharmonic complex tone (I)Δf. Both 'H' and 'I' tone complexes had the same envelope repetition rate (equal to F0), but differed in their TFS due to the Δf. The TFS1 test was a two-interval forced-choice task. On each trial, two consecutive intervals were presented, separated by 300 ms. Each interval contained four successive 200 ms tones, separated by 100 ms in either HHHH or HIHI sequences. Participants were instructed to indicate the interval that had pitch fluctuations; they were given visual feedback (correct/incorrect).The starting phases of the components were randomly selected for every presentation. This prevented envelope shape from being used as a discrimination

cue (Moore & Sek, 2009). Both tone complexes were passed through a bandpass filter with a flat passband and skirts that decreased in level at a rate of 30 dB/octave. A background threshold-equalising-noise (TEN), extending from 200 to 16,000 Hz, was used to mask combination tones and to limit the audibility of components on the lower side of the bandpass filter (Moore et al., 2000). Δf was the manipulated variable, initially set to 0.5F0. Δf varied from trial to trial according to a 2-down 1-up procedure, in order to estimate the value of Δf producing 70.7% correct responses (Levitt, 1971). The value of Δf was changed by a factor of 1.953 until the first reversal, then by a factor of 1.5625 until the second reversal, and by a factor of 1.25 thereafter (Moore et al., 2009). After eight reversals, the run was terminated and the threshold was estimated as the geometric mean of Δf values at the last six reversals. The TFS1 test was installed on a DELL Latitude 6420 laptop and stimuli were presented via Sennheiser HD 25 1-ii headphones in a double-walled sound-attenuating booth. All participants completed a practice run to ensure they understood the task. The stimulus was presented monaurally to the right ear at 70 dB SPL. The background TEN was presented at +15 dB SNR, started 300 ms before the first tone in the interval and ended 300 ms after the last tone in the second interval. The stimulus parameters (F0, number of components) were identical to those used for the tonal complex electrophysiology recordings (Figure 19).

2.5 Training program

Participants received a brief period of training with the aTune (v 1.3) pitch and timbre training program (Vandali et al., 2015). Training was completed over the course of 14 days and participants were required to complete 9 days (30 minutes/day) of training. aTune is an adaptive computer-based training program in which listeners match acoustic patterns of pitch and spectral timbre to visual patterns on the computer screen. aTune

incorporates training of discrimination under various stages and combination of acoustic cues. For this study, participants were instructed to complete only two stages of the training program: (1) pitch and (2) spectral timbre, at the easiest difficulty Level in the initial stages of training and were encouraged to repeat the Levels in Stage 1 (pitch) at a harder difficulty Level in the later stages. Each Stage comprised 12 levels and the training required participants to progress through each Level and Stage sequentially, without skipping. Training incorporated tasks that increased in difficulty with increasing Levels. For example, in the initial Levels, the F0 interval of the tokens was set to a size that the participant could easily discriminate (as determined by discrimination tests conducted prior to the training). As the participants progressed through the Levels, the F0 interval was reduced and the velocity of the visual patterns increased, adding additional challenge and diversity to Levels.

For both training Stages, synthetic harmonic complex tones were used to train discrimination of a single cue by varying F0 or center frequency, for Stage 1 and 2 respectively. All tone complexes were 500 ms long. F0 for the tone complexes ranged from 87.3 Hz to 784 Hz. A Level commenced with the F0 of the tones chosen from the available range and was fixed within the Level. Center frequency was chosen from one of the possible ranges for the passband depending on the Level: "low" which spanned a range of 2 to 5 × the highest F0 of the tones to be presented within the training run; "mid" (6 to 14 × highest F0); and "high" (10 to 31 × highest F0). Each level lasted for 5 minutes and if the task was not completed, it would be recorded as a failed attempt and that Level would need to be repeated. The aTune software provided helpful hints and visual/acoustic feedback during each level and after the completion of a level. The software logged all session times, dates and scores.

Results

Training information was retrieved from log files. All participants completed the minimum requested training amount of 4 hours and 30 minutes, with a mean duration of 5.6 hours (SD 2.3). Only five out of the 9 participants progressed up the levels/stages. Although most participants experienced frustration initially, the tasks got easier as they progressed to a higher level in the game through the sessions. Two participants who progressed to advanced levels in the game more quickly than others were the youngest in the group.

Cortical Event-Related Potentials

Robust ERPs were obtained from all subjects for each stimulus in each session. The response window for identification of each ERP components (N1, P2, P3) was defined using the grand averaged waveforms for each stimulus type (speech/tonal complex), as depicted in Figure 20. N1 was defined as the largest negative peak at Cz in response to the standard stimuli within the 90 – 150 ms time window; P2 was defined as the largest positive peak at Cz to the standard stimuli within 180 – 280 ms time window; and P3 was defined as the largest positivity at Pz to the deviant stimuli within 290 – 800 ms time window, measured from the deviant-standard difference waveform. Latencies were identified at the center of the peak within the time window. Amplitudes of N1 and P2 were calculated as the mean voltage within +/-25 ms window centered on the peak determined from the grand mean waveform, P3 amplitude was calculated as the mean voltage across a +/-50 ms window.

Repeated measures analyses of variance (rmANOVA), with $\alpha = 0.05$, were separately performed using IBM SPSS 20 for each ERP component (N1, P2, and P3) to determine the statistical significance of 2 x test sessions (pre, post), 2 x stimulus types (tonal complex, speech), 2 x stimuli (Δf (0, 50 Hz) or /di, ti/) on latencies and amplitudes. Tests of simple effects to explore interaction effects were conducted using t-tests. Magnitudes of effect sizes were estimated from partial eta-squared (η_p^2) and Cohen's (*d*) values, for rmANOVA and t-tests respectively. The results describing the main and interaction effects of the rmANOVA are summarised in Table 7, and findings that are relevant to the experimental hypotheses are described below.

Table 7. Results of repeated measures ANOVA of N1, P2, and P3

	Latency						Amplitude					
	N1		P2		P3		N1		P2		P3	
	F	*p value*	F	*p value*	F	*p value*	F	*p value*	F	*p value*	F	*p value*
Sessions	2.67	0.14	36.02	< 0.001	29.89	0.001	0.01	0.904	12.48	0.008	0.82	0.389
Type	5.78	0.04	12.02	0.008	84.84	< 0.001	5.16	0.053	0.60	0.458	12.20	0.008
Stimuli	13.31	0.007	0.02	0.882	1.65	0.234	27.99	0.001	5.66	0.044	0.04	0.831
Sessions x Type	1.00	0.34	13.34	0.006	0.93	0.361	0.02	0.878	9.92	0.014	0.34	0.575
Sessions x Stimulus	0.31	0.58	16.76	0.003	1.27	0.292	0.53	0.486	0.02	0.871	1.02	0.342
Type x Stimulus	2.30	0.16	0.16	0.696	2.16	0.179	22.79	0.001	6.13	0.038	1.36	0.276
Session x Type x Stimulus	1.76	0.22	9.59	0.015	0.79	0.399	0.11	0.746	1.94	0.200	2.64	0.142

3.1.1 Latency

Grand averaged waveforms for pre-training and post-training sessions for the two stimulus types are shown in Figure 20 and 21. Analysis of peak latencies revealed a significant main effect of test session for the P3 component ($F(1,8) = 29.8$, $p = 0.001$, $\eta_p^2 = 0.7$). P3 latency decreased by 23.1 ms (SE 1.0), on average, following brief training. A significant two-way interaction between session and stimulus type was seen only for P2 latency ($F(1,8) = 13.3$, $p = 0.006$, $\eta_p^2 = 0.6$). To further explore this interaction, response latencies were averaged across stimuli (0, 50 Hz or /di, ti/) and t-tests were conducted to compare P2 latencies between the pre and post sessions for each stimulus type (tonal complex, speech). These comparisons revealed significant differences between test sessions only for the speech stimulus type ($t(8) = -5.6$, $p < 0.001$, $d = 2.2$); P2 latencies increased following training (see figure 22). Although not significant, P2 latency for tonal complex also showed a trend for increased latencies post-training ($t(8) = -1.5$, $p = 0.159$).

3.1.2 Amplitude

On average, P2 amplitude increased from 1.7 μV (SE 0.4) to 2.7 μV (SE 0.5) at the second recording session after the brief training period. A significant two-way interaction between session and stimulus type was seen for P2 amplitude ($F(1,8) = 9.9$, $p = 0.014$, $\eta_p^2 = 0.5$), similar to that observed for latencies. Simple effects analyses comparing amplitude changes between the pre and post sessions showed a difference only for the speech stimulus type ($t(8) = -4.4$, $p = .002$, $d = 1.1$). Although not significant, P2 latency for tonal complex also showed a trend for longer latencies post-training ($t(8) = -1.9$, $p = 0.091$) (see figure 22).

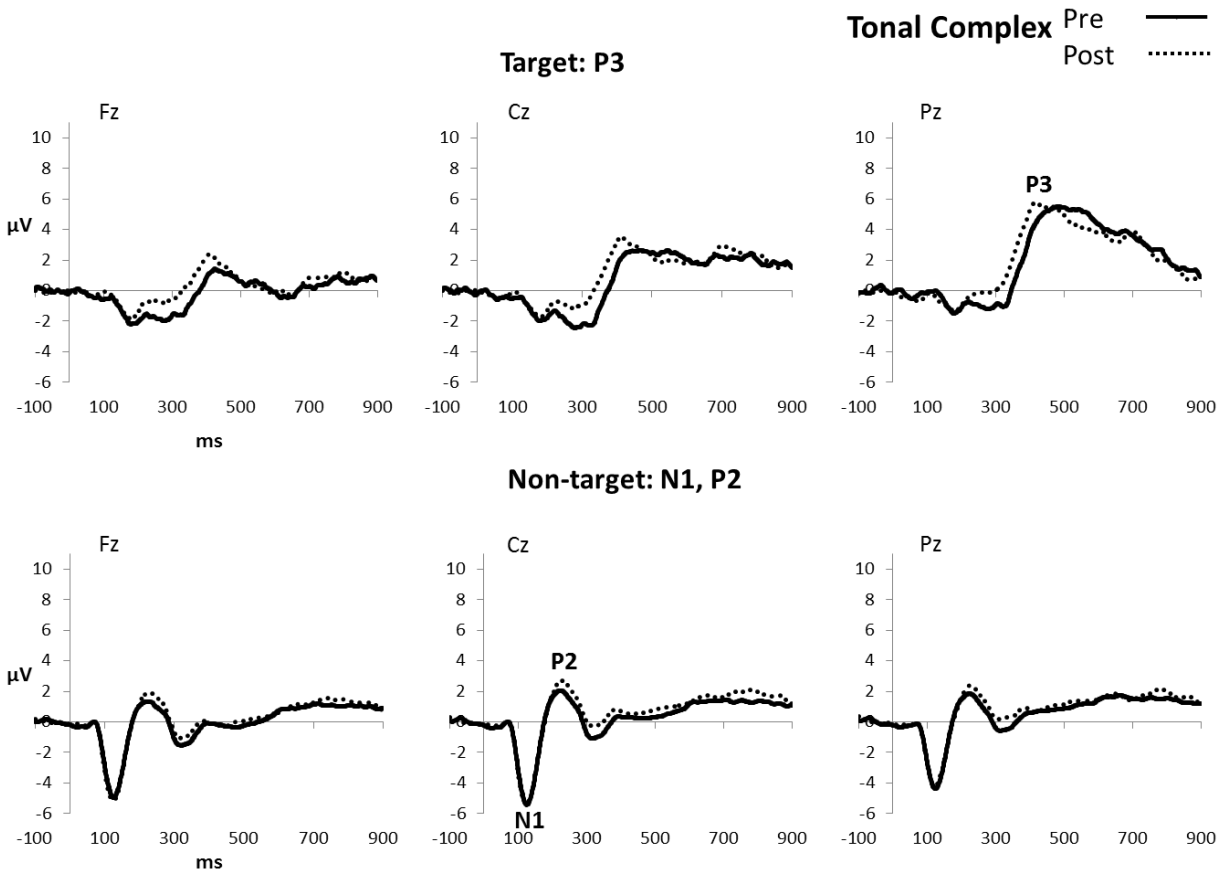

Figure 20. Grand average Fz, Cz, and Pz waveforms are overlaid for the pre-training (solid line) and post-training (dotted lines) sessions averaged across the tonal complexes. Top portion shows the P3 response recorded from the target stimulus and bottom portion shows the N1, P2 peaks corresponding to the non-target stimulus in the oddball sequence, respectively.

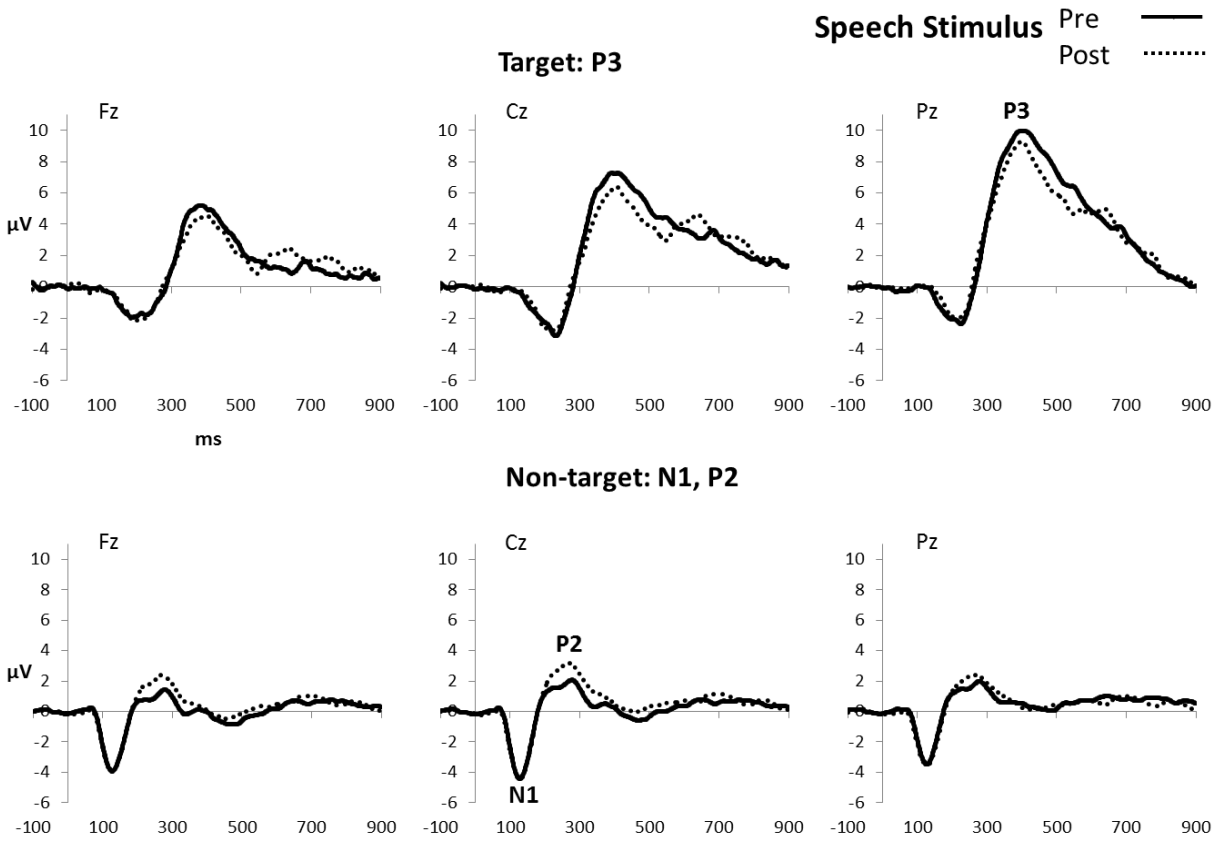

Figure 21. Grand average Fz, Cz, and Pz waveforms are overlaid for the pre-training (solid line) and post-training (dotted lines) sessions averaged across the speech stimuli. Top portion shows the P3 response recorded from the target stimulus and bottom portion shows the N1, P2 peaks corresponding to the non-target stimulus in the oddball sequence, respectively.

3.1.2.1 Inter-subject variability

Figure 22 shows training-related changes (post minus pre) across participants for all ERP components. To further explore the variability of training-related changes across the study sample, participants were divided into 'good' (n = 5) versus 'poor' (n = 4) players based on their training 'mastery' (defined by the time taken to progress up the levels of the training software). There was a statistical trend for larger P2 amplitudes in good (M = 1.2, SE 0.2) versus poor (M = -0.1, SE 0.2) players (U = 2, p = 0.063). There were no

other statistical links between evoked potential or behavioral measures and training mastery.

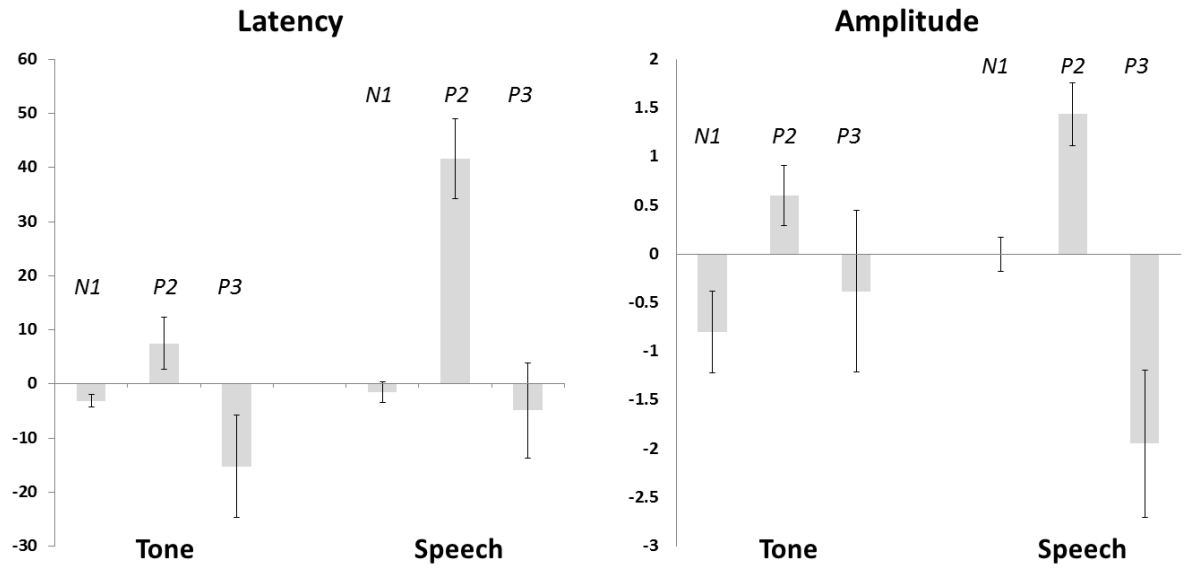

Figure 22. Training-related changes (post minus pre) across participants for all ERP components.

Behavioural measures

4.1 RT and d' sensitivity

Figure 23 displays mean RT and d' scores for the two stimulus types (tonal complex and speech) for the pre and post training sessions. RT and d' scores for speech stimuli were better than those for the tonal complexes. There were no main effects of session or

interactions between session and stimulus type for RT or d' scores (rmANOVAs, p > .05).

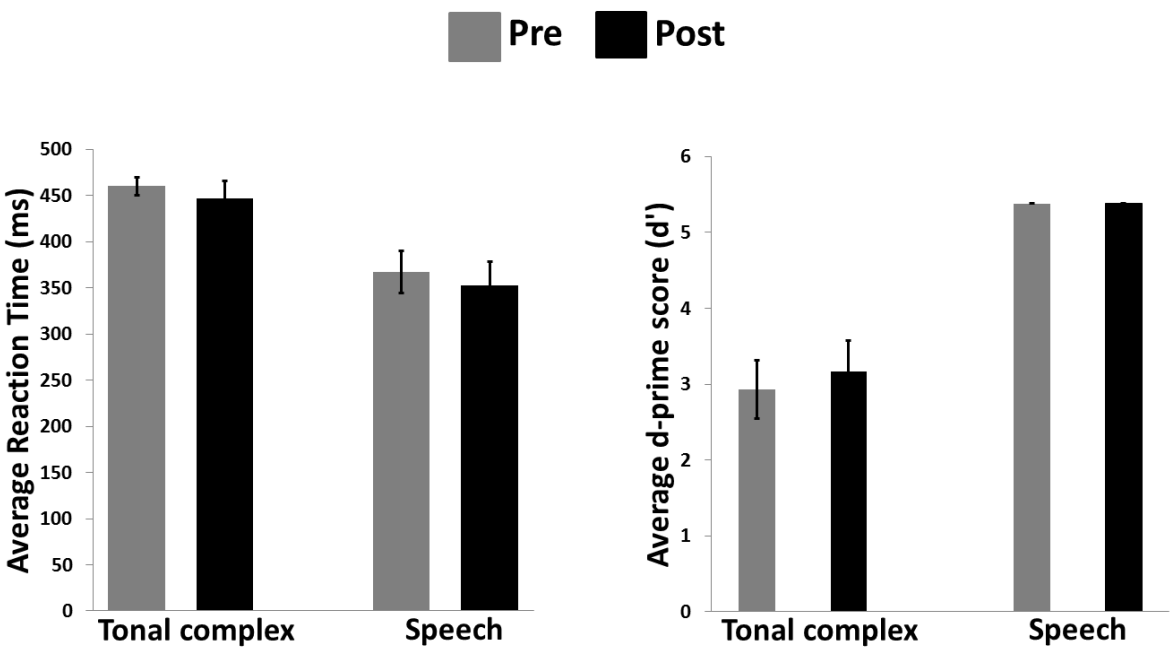

Figure 23. Average reaction time (left column) and d-prime (right column) scores for pre-training and post-training sessions using the oddball sequence during cortical ERP recording. Scores are averaged across each stimulus type. Error bars represent one standard deviation.

4.2 TFS1 test

Figure 24 displays the mean thresholds for the TFS1 test in the pre and post training sessions. One outlier with a very poor TFS threshold of 38 Hz pre training improved to 15 Hz post training (oldest participant in group). A Wilcoxon signed-rank test showed that the pre and post training scores did not differ significantly ($p = 0.762$, $r = -0.07$). Median post-training thresholds for the nine participants (14.4, SE 1.6) were significantly poorer

than thresholds of adults with normal hearing (8.3, SE 1.2; U = 13, p = 0.014) from our previous study (Kuruvilla-Mathew et al., 2015).

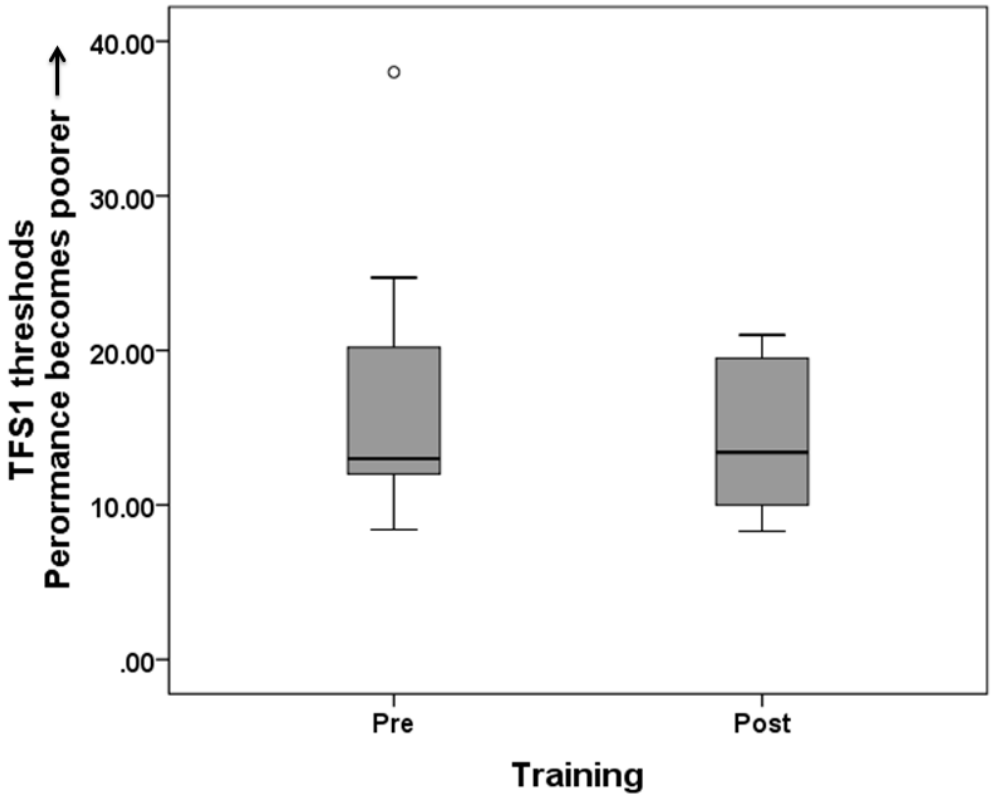

Figure 24.Box plots representing TFS1 thresholds obtained by participants for the pre-training and post-training sessions. The median scores are indicated by the thick horizontal line. Boxes indicate the data falling between the 25th and 75th percentile and the whiskers indicate the 95% confidence intervals.

Discussion

Testing after a brief period of training (9 days) using the aTune software was associated with changes in the amplitude and latency of the P2 and the latency of P3 cortical ERPs. The brief training was not associated with a significant improvement in pitch processing

measured behaviorally (RT, d', TFS1 test). Links between ERPs and behavioral data have been explored to establish the 'brain-behavior' relationship (Ross et al., 2013). The reason for the lack of behavioral change in the presence of a change in ERPs in the current study is not clear.

Previous studies have associated long-term musical training (Seppänen et al., 2012) and short-term auditory training (Atienza, et al., 2002; Bosnyak, et al., 2004) with enhanced P2 amplitudes. In the current study, P2 amplitude and latency showed an overall trend for longer latencies and larger amplitudes post-training for both stimulus types but this effect was significant only for the speech stimuli. The differential increase in P2 amplitude for speech versus tonal stimuli is not completely consistent with reports of enhanced P2 responses occurring simply as a result of repeated stimulus exposure during the data acquisition process (Sheehan et al., 2005). Thus the increase in P2 amplitude could reflect training-related learning, however, as the training involved pitch and timbre discrimination of complex tonal stimuli it was anticipated that learning effects would be greater for the tonal complex than for the speech stimuli with VOT differences. Ross et al. (2013) suggested that P2 gains preceding behavioral improvements may be an indicator of learning; at first a strong representation of the auditory object is necessary (memory updating), which in turn allows learning of fine differences between stimuli with further auditory training. The increase in P2 amplitude observed in the current study could reflect the first phase of enhanced object representation through memory updating. This would be consistent with the trend observed for P2 amplitude for good versus poor players. P2 latency effects seen in the current study could reflect changes in P2 morphology.

It is possible that the change in P2 seen only for the speech stimuli reflects greater stimulus salience of the speech than the tonal stimuli. Differential P2 enhancement across stimulus types could be due to greater task familiarization, attention or motivation for the more salient speech stimulus rather than the training. In support of this rationale, behavioral scores were better for the speech stimulus, which produced shorter RTs and better d' scores than the tonal complex contrast (see Figure 23). This stimulus salience effect was also observed as an early and robust P3 evoked by the speech stimulus contrast compared to the tonal stimuli (see Figure 20 & 21). The inclusion of a control group who had undergone testing but not formal training would have provided evidence regarding this possible explanation.

P3 latency and amplitude are related to stimulus evaluation and confidence of detection in an oddball discrimination task (Weinstein, 2000). This is consistent with generally faster and more accurate behavioral responses and larger and earlier P3 responses for the speech contrast than the tonal complex. Decreased P3 latency suggests easier discrimination after training, which is consistent with previous studies that have shown P3 latency to increase with the complexity of stimulus evaluation and more difficult discrimination tasks (Ritter, Simson, & Vaughan, 1972; Walton et al., 1986). As the effects on P3 latency were not specific to the tonal complex stimuli these changes could also be a result of non-training related effects. It is possible that participants were more relaxed or more motivated in the second recording session (e.g. because of familiarity with experimental procedures), leading to a reduction of physiological noise and consequent changes in ERP latencies,

however there were no behavioral changes or changes in P3 amplitude which might be expected if these factors accounted for the P3 change.

Most studies have tested training-related neural and behavioral changes in adults with normal hearing (e.g. Alain et al., 2010; Atienza et al., 2002; Tremblay et al., 2001, 2014). Adults with SNHL may require much more auditory training exposure to improve behavioral performance noticeably. Training effects may also be related to the complexity of the stimulus and the difficulty of the listening task (Watson, 1991). Consistent with this, Fu et al. (2005) found that the amount and time course of training-related improvement varied greatly across cochlear implant participants, with some showing rapid improvement after just a few training sessions and others changing much more slowly. Tremblay, Kraus, and McGee (1998) suggested that tracking the time course of neural changes and behavioral learning associated with training may be useful to determine efficacy of the training method, i.e., whether the training method should be continued or whether it should be adjusted/improved. Testing training effects using a randomized control group design, using training-task related stimuli and non-task-related stimuli and inclusion of both behavioral and non-behavioral measures of training effects is needed to establish whether auditory training is worthwhile for people with SNHL.

Because of the exploratory nature of the current study it is not clear whether the P2 and P3 changes reflected perceptual learning or other effects. Comparing ERP changes to a control group trained on a task unrelated to pitch and timbre would be helpful in future studies to determining whether pitch perception and speech discrimination is improved by computer-based auditory training. Furthermore, inclusion of another control group

involved in no formal training and comparing pre/post results might help differentiate P2 enhancement due to training vs repeated stimulus exposure. The current finding of differential P2 changes post-training for speech and tonal stimuli suggests that it would be useful to explore different combinations of training and test stimuli coupled with speech perception measures to determine whether any changes in the discriminability of brief stimuli generalise to speech perception in noise. Future studies using a larger and more homogeneous sample (age, hearing loss, a control group, additional baseline and follow-up assessments, and ERP recordings from multiple electrode sites to enable source analysis) are needed to better establish links between ERP and perceptual changes associated with short-term auditory training in people with SNHL.

Chapter 5
Discussion

This chapter summarises the major findings of the three studies presented here, implications of results and suggests future directions.

The overall purpose of this research was to investigate the role of electrophysiology in understanding cortical processing of signals in noise in adults with normal hearing and in adults with mild to moderate SNHL. In general the results show that it is possible to use a range of auditory evoked potentials (obligatory CAEPs, ACC, P3) to show differences in cortical encoding of natural speech stimuli and complex tonal stimuli. Study 1 showed that CAEPs were influenced by both spectral and temporal features of the speech stimuli. Study 2 utilised tonal stimuli differing in spectral and/or TFS cues to explore the effect of these cues on perceptual sensitivity and cortical processing of obligatory (N1, P2) and discriminative (ACC) cortical evoked potentials. Cortical responses were sensitive to the effects of background noise (Study 1), SNHL (Study 2) and auditory training (Study 3). The link between performance on perceptual tasks and evoked potential characteristics was explored in Study 2 and Study 3. These studies showed that, in general, there was

some correspondence between neural stimulus-encoding and overall performance in the behavioural task. The use of complex tonal stimuli in this research was valuable for investigating processing of specific acoustic cues that are important for speech perception in noise.

Key findings of the three studies are as follows:

Study 1 "*Cortical encoding of speech acoustics: effects of noise and amplification*"

1. The characteristics of the speech stimulus and the presence of background noise affected CAEP N1 and P2 components for both unaided and aided conditions.
2. CAEP latencies were more sensitive than amplitude measures in reflecting differences across all conditions tested here.
3. The addition of background noise (+10 dB SNR) significantly increased N1 and P2 latencies and decreased P2 amplitudes for most stimuli.
4. CAEP responses were earlier and larger for stimuli with shorter rise times and low frequency spectral content at the stimulus onset.
5. CAEPs (across all conditions) reflected acoustic differences consistently only for those stimulus contrasts with salient acoustic differences.
6. The CAEP findings indicated that the hearing aid did not eliminate any spectro-temporal stimulus features but did add some, depending on the aided SNR and the hearing aid's frequency-specific gain characteristics
7. Aiding had an overall impact on P2 peak (latency and amplitude) and this was linked to acoustic characteristics of the hearing aid transduced speech stimuli.

Study 2 "*Electrophysiological and behavioural processing of complex acoustic cues*"

1. Tonal stimuli differing in spectral and/or TFS cues were discriminated better by people with normal hearing than by participants with SNHL, even though the frequency content in the stimuli fell within the participants' normal hearing range.
2. Stimuli containing low numbered harmonics (N6) had greater pitch salience than stimuli with high numbered harmonics (N12). Greater pitch salience was determined by the availability of both TFS and spectral cues for discrimination.
3. Behavioural discrimination abilities and cortical encoding measured using N1, P2, and ACC evoked potentials depended on the stimulus condition (pitch salience).
4. P2 and ACCs were sensitive to differences in pitch cues between tonal stimuli.
5. Behavioural and electrophysiological measures showed that individuals with SNHL have reduced sensitivity to complex acoustic cues compared to controls with normal hearing.

Study 3 "*Auditory training of people with hearing loss: Effects on auditory event-related potentials in the absence of behavioral change*"

1. Testing after a brief period of training (9 days) resulted in neurophysiological changes without significant improvements in behavioural discrimination abilities.
2. The P2 response to training-related (tonal) and non-related (speech) stimuli was differentially affected post-training.
3. Changes in P3 (reduced latency) that were non-stimulus-specific after training suggest easier (less effortful) discrimination abilities. There was an association between P2 change and training mastery.

4. Overall results cannot be attributed to training-related neuroplasticity without the use of a randomized control study design, especially when behavioural changes did not accompany the ERP changes.

The set of studies in this thesis explored various types of signal (speech and non-speech) and recording conditions (noise, aiding, training) influencing cortical ERPs. In previous studies cortical ERPs have been used to understand physiologic processing of speech and non-speech stimuli (e.g. Agung et al., 2006; Digeser et al., 2009), the effects of hearing aid processing (e.g. Billings et al., 2007, 2012; Easwar et al., 2012) and the impact of auditory training on neural processing of speech contrasts (e.g. Tremblay et al., 2014). The research conducted here is consistent with these earlier studies, but adds to this literature by demonstrating the effects of specific acoustic cues in individuals with SNHL. The use of complex tonal stimuli differing in TFS cues to evoke ACC and P3 is another novel feature of this research.

Combined behavioural and electrophysiological approach

Behavioural studies have shown that psychophysical abilities are related to speech-in-noise perception abilities in adults with SNHL (Bernstein et al. 2013; Glasberg & Moore, 1989; Phatak et al., 2009). Although behavioural measures are useful as they characterise the performance of people with SNHL, there is variability across participants, including those with similar hearing configurations. Other tools are needed to understand these individual differences. In Study 2 participants with SNHL had a range of abilities on the TFS1 test, and performance on this task was not correlated with audiometric thresholds. A goal of the current research was to find links between evoked potential and behavioural findings, however in Study 2 no ACCs were recorded for the deltaF=50 Hz stimulus

condition for the SNHL group even when they could do the discrimination task behaviourally. Thus in Study 2 the electrophysiology underestimated behavioural performance on the TFS1 task. ACCs are pre-conscious discrimination potentials and the TFS1 behavioural task required effortful listening for the participants with SNHL. Thus top-down attentional processing during the behavioural task may have contributed to this mismatch between ERPs and behaviour.

Study 3 provided ERP evidence for rapid changes within the cortex without any behavioural change in discrimination performance. ERP studies such as this provide objective evidence for differences in neural activity in response to different stimuli; the mismatch between ERP and behavioural findings suggests that auditory cortex encoding of stimulus differences alone may not be sufficient for stimuli to be behaviourally discriminated. In Study 3 an active ERP protocol was used in order to evoke P3 responses. In this case brain responses to stimulus contrasts were enhanced after training with no change in behavioural discrimination. Additional training may be needed to make the difference between stimuli more salient in order for perception to occur. Longer duration studies, with repeated measurement of ERPs and behavioural discrimination may be needed to better characterise this brain-behaviour link. In Study 2 a passive ERP protocol was used and some of the stimulus conditions (N12) that were behaviourally discriminated did not generate a difference in brain responses. Differences in top-down attentional processes may account in part for varying results across studies when brain and behavioural responses are compared (Ross et al., 2013). Understanding this brain-behaviour relationship and finding a common ground between evoked potential and behavioural measures such that one measure can be used to predict the other is an important goal of this research. This would enable CAEPs to be used to assess benefits of

hearing rehabilitation (hearing prostheses, auditory training) objectively in difficult-to-test populations.

Participant characteristics

There is a lack of published data related to the underlying neural processing of acoustic cues important for speech perception for different age groups and different severities and configurations of hearing loss. Some studies have measured auditory evoked responses in participants with normal hearing (e.g. Agung et al., 2006, Billings et al., 2007), others have investigated participants with simulated hearing loss (Martin & Stapells, 2005) and other studies have used participants with SNHL (Carter et al., 2013). There is evidence for changes in neuroplasticity following SNHL (Campbell & Sharma, 2013) however the time since onset of the hearing loss is not usually considered in ERP studies unless the participants are cochlear implant users. Individuals with SNHL demonstrate large variability in performance for suprathreshold listening tasks (Moore, 2008). Partly because of this variability, suprathreshold cortical ERP studies typically focus on examining adults with normal hearing to control for variables associated with SNHL. Studies of participants with normal hearing such as Study 1 can provide baseline information about the influence of stimulus acoustics and CAEPs, however, caution is needed in applying aided results to participants with SNHL using hearing aids (Carter et al., 2013). In people with SNHL factors such as hearing aid noise and loudness recruitment may make it difficult to extrapolate findings from participants with normal hearing to participants with hearing loss. In Study 1 listeners had normal hearing and were tested at relatively high sensation levels and the acoustic measurements showed that the signal level was high relative to the hearing aid noise floor. Under these conditions aiding did have a significant effect on CAEPs. Extrapolation of these findings to listeners

with SNHL should be done with caution and requires consideration of factors such as stimulus sensation level and hearing aid noise.

Sample size and evoked potential reliability

One disadvantage of far field scalp ERP recordings is that small amplitude responses are obtained. ERPs always contain noise not related to the evoking stimulus and hence the process of determining whether stimulus and other parameters affect ERPs in a systematic way is a signal to noise problem. Luck (2005) listed some of the factors contributing to noise in ERP recordings, including insufficient number of trials per participant and stimulus condition, EEG activity (alpha waves), eye blinks, high electrode impedance and electrical noise from the environment. Effects of various experimental conditions described in this thesis were demonstrated using a relatively small sample size (~N = 10). Despite the small sample size it was possible to show differences in ERP morphology between stimulus and other conditions with moderate effect sizes. For example in Study 3 effect sizes (η_p^2) of > 0.1 were evident when comparing pre- and post-training N1 and P3 latencies. Effect size statistics provide a better estimate of treatment effects and clinical significance than *p* values alone. The ability to record reliable ERPs depends on many factors including the test design and experimental conditions, salience of the evoking stimuli, number of trials per participant and nature of the evoked component. For example, in the third study the effects of training were investigated by recording early and late ERP components (N1, P2, P3) using an active listening paradigm. The focus on later and large components such as P3 has the advantage of high SNR (Luck, 2005). In general, the ERP components investigated here and the experimental conditions generated robust ERPs, particularly for the active listening paradigm in Study 3. The experiments described here involved ERP recordings at suprathreshold levels and were presented at

relatively high sensation levels for both normal hearing and SNHL participant groups. Despite the use of high stimulus levels response identification was problematic for some conditions when the stimuli were less salient. For example, in Study 1, CAEPs evoked by voiceless stimuli (e.g. /si/) were difficult to distinguish from the noise floor for the noise condition in a few individuals (Figure 3). A similar problem occurred for the N12 condition (weak pitch) in Study 2. Although a window-based calculation of mean ERP amplitudes was used to improve SNR (Luck, 2005), response identification (at least in the grand average waveform) in these conditions would have benefited from a larger sample size.

Behavioural studies have reported variability in TFS abilities depending on age, degree of hearing loss, and cognitive abilities (Moore, Vickers, & Mehta, 2012; Neher, Lunner, Hopkins, & Moore, 2012). Picton et al. (2000) noted that it is important to include a homogenous group with similar profiles when looking at data from individual subjects using a small sample size. Study 2 showed significant differences in TFS processing between participant groups with just a small sample size (NH, n = 10; vs. SNHL, n = 9). Participants were selected to have good low-mid frequency hearing and mild-moderate high frequency hearing loss. Because a small homogeneous group of participants with SNHL was investigated in both Study 2 and Study 3 it is difficult to generalise the results across the entire population of people with SNHL.

Choice of ERP recording electrodes

ERP analyses and recordings described in this thesis were made using only a few adjacent active electrodes (Fz, Cz, Pz). The ERP protocols utilised selected electrodes with future clinical applications in mind. In each study recordings were made at the vertex which consistently generates robust ERP responses (Picton, Lins, & Scherg, 1995). Other electrode sites were selected because of differences in scalp distribution for specific peaks. In addition to Cz, Fz was also used to record N1 in Studies 1 and 3 (Korczak & Stapells, 2010). A Pz electrode was used to record P3 in Study 3 (Picton, 1992). A limitation of this approach is that auditory cortical ERPs have multiple generators (Crowley & Colrain, 2004; Näätänen & Picton, 1987; Picton, 1992) and the use of just a few electrode sites limits the ability to look for differences in scalp distribution (e.g. Wood & Wolpaw, 1982), global field power (e.g. Murray, Brunet, & Michel, 2008) and ERP generators (e.g. Näätänen & Picton 1987).

In Study 1, N1 amplitudes were bigger in the presence of multi-talker babble and with the hearing aid at the Fz electrode site, which may indicate a more frontal distribution for N1 under those conditions. The observed enhancement of N1 amplitude at Fz could possibly reflect a frontal component of N1 related to higher levels of alertness or cerebral arousal activated for speech in noise and with a hearing aid in place (at a high sensation level) (Näätänen & Picton 1987). Participants in this study were involved in a passive listening task (watching subtitled videos), and were instructed to ignore the stimuli. The use of multi-channel recordings for the speech in noise paradigm would allow source analysis (Leavitt, Molholm, Gomez-Ramirez, & Foxe, 2011) to determine whether N1 generators shifted with the addition of noise and amplification, and could help delineate the contribution of top-down attentional processes using spectral analysis of the EEG recordings (Cervenka, Nagle, & Boatman-Reich, 2011). EEG oscillations may provide

additional information on attentional processes during auditory ERP recordings (Kisley & Cornwell, 2006).

Poeppel (2003) proposed that while the left hemisphere is good at processing slow changes in signals (25 to 50 Hz, ENV cues), the right hemisphere preferentially extracts information from rapid changes in the signal (4 to 7 Hz, TFS cues). Thus Study 2 could have benefited from measures of hemispheric asymmetry using multiple electrode recordings. There have also been reports of hemispheric asymmetry in auditory training effects (Tremblay & Kraus, 2002) and multichannel hemispheric recordings could have been a useful addition to Study 3. Non-stimulus specific changes in P2 were evident in Study 3 after training. Seppannen et al. (2012) showed increased P2 source activation after an active adaptive discrimination task in musicians compared to non-musicians and linked this effect to rapid neuroplasticity, and hence source analysis based on multichannel recordings could have been a useful addition to Stduy 3. Study 3 also showed stimulus-specific ERP changes (decrease in N1 and P3 latencies for tonal complexes). N1 and P3 waves have several underlying components that are thought to reflect different sensory and/or cognitive processes (Hillyard et al., 1973; Näätänen & Picton 1987; Picton, 1992). Source analyses based on multiple electrode recordings could help to determine the relative contribution of different processes to the observed changes in P2, N1 and P3 relative to the stimulus following training.

Top-down versus bottom-up processing

In Study 3 N1 and P3 latencies were shorter after training for the training-related (tonal) stimuli. Shorter P3 latencies are thought to reflect less attentional allocation associated

with less effortful listening (Polich, 2003). Recent reviews of speech recognition and cognition have concluded that when listening in adverse conditions, listeners' use top-down skills to enhance auditory discrimination (Akeroyd, 2008; Humes, 2007). Individuals with hearing loss may rely more on their cognitive processes when listening to signals in noise than people with intact hearing and this can have adverse effects on cognitive processes such as working memory and attention due to the increased resource allocation associated with more effortful listening in people with hearing loss (Akeroyd, 2008; Pichora-Fuller & Singh, 2006). Cortical ERPs are sensitive to bottom-up sensory processing and top-down cognitive processes. As the balance of top-down versus bottom-up processes may differ between normal hearing controls and participants with hearing loss, it would be useful to extend Study 3 to include a control group of participants with normal hearing undergoing auditory training to see if they showed the same changes in P3 latencies.

Cortical encoding of signal acoustics

Study 1 showed significant differences in latencies and amplitudes for a number of speech contrasts (see Table 3). Several speech-evoked CAEP studies have showed evidence of different response patterns depending on the speech features (voice onset time (VOT); place of articulation; manner of articulation) being examined (e.g. Agung et al., 2006; Kaplan-Neeman et al., 2006; Koch, McGee, Bradlow, & Kraus, 1999; Sharma & Dorman, 1999; Tremblay et al., 2003). However, there are some inconsistencies in the findings of speech-evoked studies depending on the type of stimulus used i.e., natural versus edited and/or synthesized speech stimuli. For example, studies investigating the effects of VOT differences have shown differential effects on N1 and P2 morphology, with most reporting differences only for N1 morphology (Dimitrijevic et al., 2013; Horev

et al., 2007; Korczak & Stapells, 2010; Sharma & Dorman, 1999; Sharma et al., 2000). These studies indicate that N1 and P2 responses are objective indicators of stimulus differences. In contrast to many earlier studies, Study 1 only showed effects of voicing on P2 morphology. The use of naturally produced speech stimuli may account for this difference between Study 1 and earlier publications. In Study 1 no CAEP components showed consistent differences for all speech stimuli and across all conditions (quiet/noise; unaided/aided) (see Table 3). Rosen (1992) suggested that information on speech features is conveyed by the acoustic characteristics of the stimulus. Specifically, while manner of articulation features are cued by the ENV, place of articulation and voicing information are cued by the TFS of the speech stimuli.

Although N1 and P2 components co-vary they can be differentiated using experimental conditions such as those used in Study 2. In the second study the early cortical ERPs (P2, ACCs) were sensitive to acoustic characteristics (TFS and spectral cues) other than the stimulus envelope rise time. Study 2 showed a link between behavioural and electrophysiological results as adults with SNHL were less sensitive to changes in TFS cues compared to adults with normal hearing and had reduced ACC amplitudes for the N6 stimulus condition (spectral and TFS cues). Deficits in TFS processing in individuals with SNHL are thought to contribute to difficulties understanding speech in noise (Moore, 2008). Although poor speech perception in noise was a complaint of the participants with SNHL in Study 2, speech scores were not measured directly.

An interesting finding also observed in Study 2 was that for the control condition (no change in place or TFS cues), adults with SNHL showed an enhanced response compared

to the normal hearing group (Figure 9). This might have occurred due to an exaggerated response to the rise time and spectral changes at the transition point between complexes. Individuals with SNHL could have greater encoding of envelope information that suppresses important TFS information (Anderson et al., 2013). When they are unable to encode one cue, listeners may rely on the available cues to make discriminations and hence may have enhanced skills for processing alternative cues. This is consistent with the findings of Kale and Heinz (2010) who found enhanced coding of ENV cues in chinchillas with mild to moderate noise-induced hearing loss compared to normal hearing. Thus, an important goal of research in this area would be to understand the relative contribution of neural encoding of ENV, spectral, and TFS cues to speech perception in noise in people with normal hearing and SNHL, in different age groups and with different severities and configurations of hearing loss.

Hearing aids and ERPs

Commercial technology has become available for using CAEPs to assess detection of amplified signals (Carter et al., 2013; Munro et al., 2011). CAEPs are a useful tool for assessing response detection at threshold levels (McNeil et al., 2005; Munro et al., 2011; Korczak et al., 2005) and can be used to check speech stimulus audibility in children with hearing loss (Golding et al., 2007). Caution is needed however when using CAEPs to obtain unaided and aided CAEP recordings at suprathreshold levels as there can be complex interactions between the stimulus, noise and amplification effects (Billings et al., 2012). Consistent with this, Study 1 revealed complex effects of speech features, background noise and amplification on CAEPs. Although detectable CAEPs were recorded for all stimuli and conditions, CAEPs to speech contrasts in quiet and in noise did not show the same differences for the unaided and aided conditions (see Table 3).

This was linked to the frequency specific stimulus onset modifications by the hearing aid and the signal to noise ratio of the speech stimulus at the onset, as revealed by in-the-canal acoustic measurements of the speech stimulus (see Figure 5). Acoustic modifications introduced by the hearing aid should be verified using in-the-canal acoustic measurements in future aided CAEP studies and would facilitate understanding of the link between the stimulus and cortical response characteristics. Easwar et al. (2012) showed that hearing output levels for the single phonemes that are usually used for CAEP recordings were lower than levels estimated for a more naturalistic running speech context. This further reinforces the need for hearing aid output measurements as the settings from one hearing aid cannot be generalised to all hearing aids, and the stimulus context is also relevant. The ACC paradigm in which cortical responses are recorded to stimulus change may be useful for addressing the problem of stimulus context since longer duration stimuli can be used.

ACCs have been previously recorded in listeners wearing a hearing aid and are less sensitive to onset related modifications than the N1 component (/si, ʃi/, Tremblay et al., 2006). Hence, recording aided ACCs evoked using a discriminative stimulus paradigm such as that used in Study 2 is recommended for future studies exploring hearing aid transduced speech. Moore (2008) suggested that individuals with SNHL with some ability to process TFS cues might benefit more from fast acting compression than from a slow acting compression. Future studies using ACC recordings could be used to determine cortical processing of hearing aid transduced speech for these different types of compression. Future studies looking at the acoustic output of the hearing aids for various hearing aid settings (noise reduction, compression) and input signals (speech and speech-like complex stimuli in different noise contexts) and correlating features of the stimulus

with ACC evoked response characteristics would help to determine what stimulus and hearing aid contrasts can be detected electrophysiologically.

Recommendations for future research

Speech in noise perception was not measured directly in any study presented here; this would be a useful addition to future work in this area in order to link psychophysical and electrophysiological findings to functional listening results. The inclusion of participants with a wider range of severities and configurations of hearing loss would allow generalisation of findings to a wider group of people with SNHL. Inclusion of a control group of untrained participants with SNHL and trained and untrained listeners with normal hearing could aid understanding of the stimulus-specific training effects observed in Study 3. For example, reduced P3 latency after auditory training was evident for people with SNHL and this was hypothesised to reflect the reduced need for allocation of attention for the discrimination task after training. If the stimulus contrast was very salient for listeners with normal hearing, then perhaps no change in P3 latency would be evident.

It will be worth exploring the relationship between cognitive, psychophysical and electrophysiological measures in people with SNHL experiencing listening difficulties in noise, particularly using ERP stimulus and recording techniques that enable the relative contribution of sensory versus cognitive processes to be evaluated, such as studies using a directed attention protocol (e.g., Neelon, Williams, & Garell, 2006). Multichannel

recordings would facilitate understanding of the observed stimulus and training effects by enabling source and other more sophisticated analyses of the ERP findings.

Overall, the results show that a range of cortical evoked responses provide objective evidence for differences in neural encoding of auditory stimuli in people with hearing loss, and that these responses could be useful clinically for determining whether hearing aids and training enhance auditory processing of spectral and temporal cues that are important for speech understanding in noise in people with SNHL.

Appendices

APPENDIX 1: Means and SDs of (A) Latencies and (B) Amplitudes (Cz Electrode) Across All Conditions in Study 1

(A)

			LATENCY (ms)						
			/di/	/gi/	/mi/	/pi/	/ti/	/si/	/ʃi/
Unaided	Quiet	P1	65.5 (4.7)	78.9 (16.1)	79.9 (17.9)	71.7 (18.9)	66.5 (21.1)	84.6 (32.2)	88.2 (22.8)
		N1	128.8 (14.9)	143.7 (14.2)	138.9 (15.8)	127.6 (20)	120.6 (20.3)	145.4 (22.1)	140 (21.7)
		P2	214.5 (19.4)	215.0 (15.5)	225.6 (31.8)	184.3 (24.2)	188.1 (16.2)	216 (28.3)	203.4 (19.5)
	Noise	P1	78.4 (9.1)	98.6 (22.7)	99.8 (12.9)	119.8 (31.2)	88.2 (27.1)	86.5 (41.3)	88.2 (19.8)
		N1	156 (14.3)	181.6 (14.5)	181.1 (17.4)	189.3 (25.2)	141.1 (23.1)	158 (17.3)	146.9 (15.8)
		P2	240.3 (30.2)	266 (24.3)	272.1 (17.8)	279.4 (40.7)	189.4 (25.8)	207.2 (21.6)	216.3 (16.3)
Aided	Quiet	P1	75.8 (20.6)	79.6 (20.1)	94.8 (23.2)	76.5 (27.6)	71.8 (17.5)	92.5 (29.1)	85.3 (23.7)
		N1	134.8 (16.1)	144.8 (13.5)	157.5 (31.2)	132.6 (20.7)	126.6 (13.9)	147.8 (28.3)	139.4 (24.8)
		P2	224.4 (20.5)	234.9 (23.9)	240.4 (33.4)	185.1 (42.7)	190.7 (23.3)	202.9 (32.2)	209.2 (19.7)
	Noise	P1	88.9 (21.3)	121.5 (17.1)	112.9 (26.6)	123.8 (29.7)	83.9 (25.1)	28.5 (28.5)	16.9 (16.9)
		N1	166.4 (17.3)	186.8 (17.2)	185 (19.8)	200.9 (17.8)	142.0 (29.5)	139.1 (29.8)	149.2 (16.1)
		P2	246.2 (21.3)	274.2 (22.3)	286.3 (26.7)	307.7 (16.2)	195.7 (34.3)	198.5 (17.8)	212.8 (17.6)

(B)

			Amplitude (μV)						
			/di/	/gi/	/mi/	/pi/	/ti/	/si/	/ʃi/
Unaided	Quiet	P1	.5 (.8)	.7 (.6)	.4 (.9)	.6 (.6)	.7 (.8)	.3 (.3)	.6 (.9)
		N1	-4.6 (3.2)	-3.5 (2.4)	-3.4 (2.9)	-3.1 (2.3)	-3.4 (2.2)	-1.9 (1.5)	-2.1 (1.8)
		P2	2.7 (1.7)	1.9 (1.9)	1.5 (1.4)	.5 (2.8)	1.4 (2.3)	.8 (1.2)	1 (1.5)
	Noise	P1	.3 (.4)	.4 (.8)	.8 (.6)	0 (.6)	.5 (.5)	.4 (.5)	.4 (.8)
		N1	-4.2 (2.6)	-3.5 (1.6)	-3.4 (1.7)	-3.1 (1.9)	-1.4 (1.5)	-0.9 (1.2)	-1.8 (.8)
		P2	.7 (1.1)	.5 (1.1)	.9 (.9)	.9 (.9)	.2 (1.4)	-.1 (.9)	.6 (1.3)
Aided	Quiet	P1	1.1 (.7)	.4 (.8)	.4 (.9)	.9 (.7)	.7 (.9)	.8 (.7)	.7 (.8)
		N1	-3.4 (2.7)	-3.1 (2.6)	-2.7 (2.6)	-2.5 (2.5)	-3.3 (2.5)	-.80 (1.3)	-2.1 (2.0)
		P2	3 (2.5)	2.8 (2.3)	1.6 (1.4)	1.2 (2.5)	1.8 (2.3)	0.5 (1.5)	1.7 (1.6)
	Noise	P1	.2 (1.3)	.2 (.5)	-.1 (.5)	.0 (.4)	.4 (.8)	.4 (.5)	.2 (.6)
		N1	-4.1 (2.4)	-3.4 (2.2)	-3.4 (1.6)	-3.6 (2.0)	-2.2 (.9)	-.6 (1.1)	-2.1 (1.1)
		P2	1.1 (1.5)	.9 (1.1)	1 (.9)	1.7 (.9)	1.8 (2.3)	.1 (.9)	.6 (.7)

APPENDIX 2: Ethics Approval

Office of the Vice-Chancellor
Research Integrity Unit

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UNIVERSITY OF AUCKLAND HUMAN PARTICIPANTS ETHICS COMMITTEE

23-Apr-2012

MEMORANDUM TO:

Prof Suzanne Purdy
Psychology

Re: Application for Ethics Approval (Our Ref. 8023)

The Committee considered your application for ethics approval for your project on 23-Apr-2012.

Ethics approval was given for a period of three years with the following comment(s).

1. In the clinic manager CF rather than stating "I agree that participation or non-participation …" please state "I provide my assurance that participation or non-participation …".

The expiry date for this approval is 23-Apr-2015.

All communication with the UAHPEC regarding this application should include this reference number: **8023**.

APPENDIX 3: Ethics Approval Amendment

Office of the Vice-Chancellor
Finance, Ethics and Compliance

UoA logo

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UNIVERSITY OF AUCKLAND HUMAN PARTICIPANTS ETHICS COMMITTEE (UAHPEC)

15-Dec-2014

MEMORANDUM TO:

Prof Suzanne Purdy
Psychology

Re: Request for change of Ethics Approval Ethics Approval (Our Ref. 8023): Amendments Approved

The Committee considered your request for change for your project entitled **Cortical encoding of amplified speech signals in noise** and approval was granted for the following amendments on 15-Dec-2014.

The Committee approved the following amendments:

1) Inclusion of training program using computer software.

2) Minor additions to the CF.

3) Current testing time to be three 2.5 hour session for each participant.

4) Koha of $ 40 offered to the participants.

Researcher should note,
1) Statements of the time involved for participants need to be corrected in PISs. The PIS for Clinic Manager says the time involved is only 2.5 hrs, and should say this is 2.5 x3 sessions plus the time using the programme at home (9 days by 15 mins). The PIS for participants notes that there are 3 sessions (of 2.5 hrs) but does not include the time at home in the statement about the time involved in participation.
2) If a programme is to be added to participant's personal computers, then they should be informed whether or not they can participate without consenting to this, and an explicit consent to a programme being added to their personal computer should be included in the CF (not just 'I understand that', but 'I agree to'). The PIS should also make clear if participants are meant to bring their own computer to somewhere to have this added (and if so when), or whether they will be expected to install this themselves somehow.

The expiry date for this approval is 23-Apr-2015.

If the project changes significantly you are required to resubmit a new application to the Committee for further consideration.

In order that an up-to-date record can be maintained, it would be appreciated if you could notify the

Appendix 4: Recruitment Brochure

THE UNIVERSITY OF AUCKLAND

Can you help with a research study investigating how the brain processes sounds?

This study will be conducted over the period 2012/2015 by researchers at the *University of Auckland*.

We aim to test around 40+ adults aged 18 years or older, with and without a known hearing loss.

Testing will involve a single 2 hour session conducted in the University of Auckland, Tamaki Campus, Building 730 Speech Science and Audiology clinics.

A number of hearing tests, both behavioural and objective, will be conducted. More details about the test are provided in the information sheet.

If you are interested in participating and would like copies of the information sheet/consent form please email Abin K Mathew:
amat527@aucklanduni.ac.nz

Also please feel free to email Professor Suzanne C Purdy (Chief investigator): sc.purdy@auckland.ac.nz

APPROVED BY THE UNIVERSITY OF AUCKLAND HUMAN PARTICIPANTS ETHICS COMMITTEE ON 23/04/2012 for (3) years, Reference Number 8023

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