The impact of frequency compression on cortical evoked potentials and perception

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THE IMPACT OF FREQUENCY COMPRESSION ON CORTICAL EVOKED POTENTIALS AND PERCEPTION

by

Benjamin James Kirby

A thesis submitted in partial fulfillment of the requirements for the Doctor of Philosophy degree in Speech and Hearing Science in the Graduate College of The University of Iowa

May 2014

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Benjamin James Kirby

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for the thesis requirement for the Doctor of Philosophy
degree in Speech and Hearing Science at the May 2014 graduation.

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Shawn S. Goodman

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Christopher W. Turner
First say to yourself what you would be; and then do what you have to do.

Epictetus, *Discourses*, Book III, Chapter 23
ACKNOWLEDGMENTS

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Wu Yu-Hsiang and Shawn Goodman provided invaluable technical assistance without which this work could not have been completed.

Many thanks are owed to my family and friends.
ABSTRACT

Nonlinear frequency compression is a signal processing technique used to increase the audibility of high frequency speech sounds for hearing aid users with sloping, high frequency hearing loss. However, excessive compression ratios may reduce spectral contrast between sounds and negatively impact speech perception. This is of particular concern in infants and young children, who may not be able to provide feedback about frequency compression settings. This study explores use of an objective cortical auditory evoked potential that is sensitive to changes in spectral contrast, the auditory change complex (ACC), in the verification of frequency compression parameters.

We recorded ACC responses in adult listeners to a spectral ripple contrast stimulus processed with a range of frequency compression ratios (1:1 to 4:1). Vowel identification, consonant identification, speech recognition in noise (QuickSIN), and behavioral ripple discrimination thresholds were also measured under identical frequency compression conditions. In Experiment 1, these measures were completed in ten adult normal hearing individuals to determine the effects of this type of signal processing in individuals with optimal hearing. In Experiment 2, these same measures were repeated in ten adults with sloping, high frequency hearing loss, which is the clinical population for whom this signal processing technique was intended.

Increasing the compression ratio did not affect vowel identification for the NH group but did cause a significant decrease in vowel identification for the hearing impaired listeners. Changes in compression ratio were associated with significant changes in ACC amplitude, consonant identification, ripple discrimination threshold, and speech perception in noise scores.

These results indicate that the ACC response, like speech and non-speech perceptual measures, is sensitive to frequency compression ratio. However, it was observed that the amplitude of ACC responses elicited by our single ripple contrast
stimulus was not strongly correlated with behavioral ripple discrimination or measures of speech perception in quiet or in noise. It may be the case that using a different ripple density in the evoking stimulus (e.g. 1 RPO, 2 RPO), or determination of ACC threshold by using multiple ripple densities, would result in stronger correlation to behavioral measures across frequency compression conditions.
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<tr>
<td>ABR</td>
<td>auditory brainstem response</td>
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<td>ACC</td>
<td>auditory change complex</td>
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<td>AEP</td>
<td>auditory evoked potential</td>
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<td>AFC</td>
<td>alternative forced choice</td>
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<td>ASSR</td>
<td>auditory steady state response</td>
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<td>CAEP</td>
<td>cortical auditory evoked potential</td>
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<tr>
<td>CI</td>
<td>cochlear implant</td>
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<tr>
<td>FFT</td>
<td>fast Fourier transform</td>
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<tr>
<td>IHC</td>
<td>inner hair cell</td>
</tr>
<tr>
<td>OHC</td>
<td>outer hair cell</td>
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<tr>
<td>RPO</td>
<td>ripples per octave</td>
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<td>SNHL</td>
<td>sensorineural hearing loss</td>
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<td>SNR</td>
<td>signal-to-noise ratio</td>
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CHAPTER 1
INTRODUCTION

1.1 Background

One of the central challenges in the field of audiology is the verification of optimal hearing aid fitting in infants and young children who cannot reliably provide detailed subjective feedback about hearing aid settings. The audiologist often depends on parental report of the child’s aided listening performance to guide selection of hearing aid settings, information that may take weeks or months to establish. Cortical auditory evoked potentials (CAEPs) are long-latency responses to auditory stimuli recorded using surface electrodes on the scalp. They have been used to help establish audibility of sound for infants using various hearing aid gain characteristics (Golding et al. 2007; Carter et al. 2010). At this time, however, no objective measure exists to evaluate the suitability of different hearing aid frequency compression parameters in this population.

Frequency compression is a technique that has been shown to be effective at increasing the audibility of high frequency sounds for individuals with high frequency hearing loss. It can, however also cause decreases in speech perception scores and reduction in the spectral contrast of speech sounds, particularly when higher frequency compression settings are used (Turner and Hurtig, 1999). In published work to date on the efficacy of hearing aid frequency compression in adults and older children, frequency compression parameters have typically been established by verbal report of the hearing aid user and set to provide the minimum amount of compression needed to improve both audibility and discriminability of high frequency speech sounds (Glista et al. 2009; Wolfe et al. 2010 and 2011). Anecdotal reports of pediatric hearing aid fitting practices indicate that some clinicians are often far less conservative in prescribing frequency compression, and frequency compression is enabled as a default in commercial hearing aid programming software, even for users with flat hearing loss configurations.
The acoustic change complex (ACC) is a CAEP that has been shown to be correlated with detection of spectral changes in an ongoing stimulus (Ostroff, Martin and Boothroyd, 1998; Martin and Boothroyd, 2000). The ACC reflects auditory discrimination at the level of the cortex and it can be evoked using spectrally complex signals that can be processed by a hearing aid. The ACC may also be recorded in young children, including infants (Small and Werker, 2012). These properties may make the ACC an ideal tool that could help clinicians better establish benefit or lack thereof from use of frequency compression in hearing aid users who may be difficult or impossible to test using behavioral approaches.

1.2 Goals and Hypotheses

The aim of this work is to describe relationships between behavioral measures of perception and cortical auditory evoked potentials in conditions of frequency compression. To accomplish this aim a series of experiments were conducted with adults with normal hearing (Experiment 1) and with high frequency sensorineural hearing loss (Experiment 2). The stimuli used in these physiological and behavioral assessments were processed using a custom MATLAB program to simulate different hearing aid frequency compression conditions. In Experiment 1, ten normal hearing listeners were tested and for each frequency compression condition, vowel identification was assessed using a ten alternative forced choice procedure. The effect of frequency compression was also tested using behavioral discrimination of complex non-speech noise stimuli (also known as spectral ripples) in an adaptive, three alternative forced choice (3AFC) procedure. CAEPs, including the ACC, were recorded using spectral ripple contrasts (0.5 ripples per octave, 1500-6000 Hz broadband noise) stimuli. It was hypothesized that for this group of normal hearing listeners, vowel identification, consonant identification, and ACC amplitudes would decrease and that signal-to noise ratio threshold for speech identification and spectral ripple detection thresholds would increase (become poorer).
with increasing frequency compression ratios. In Experiment 2, ten individuals with high frequency SNHL were tested, and high frequency gain derived from each individual hearing-impaired participant’s prescribed audibility targets was applied to the stimuli and the tasks of Experiment 1 were repeated. For this group of hearing-impaired listeners it was hypothesized that mild compression settings would either have no impact or possibly improve performance on the ACC, vowel, consonant, behavioral ripple, and speech identification in noise tasks but that as the amount of compression was increased ACC amplitude would decrease and performance on the behavioral tasks would become poorer.
CHAPTER 2
REVIEW OF THE LITERATURE

2.1 Effects of High Frequency Sensorineural Hearing Loss

High frequency, sloping hearing losses constitute the most commonly observed audiometric configuration in adults and children (Cruickshanks et al. 1998, 2003; Niskar et al. 1998). Conventional hearing aids may restore the audibility of high frequency speech sounds for some hearing-impaired individuals, though more severe high frequency hearing loss can present challenges to hearing aid fitting. Some of these challenges are technical, such as limited output bandwidth of hearing aids (Killion, 1997) or limited high frequency gain before onset of acoustic feedback (Lybarger, 1982; Kates, 1999). However, the impaired auditory system itself may also limit how much improvement in performance can be achieved by restoring audibility of high frequency sounds. Hogan and Turner (1998) and Ching, Dillon, and Byrne (1998) showed that extending the audible bandwidth to frequencies above approximately 4000 Hz can result in minimal benefit for the listener or cause decrements in performance on speech perception tasks for listeners with more severe high frequency hearing losses. These investigators hypothesized that in addition to loss of outer hair cells (OHCs), hearing thresholds greater than approximately 55 dB HL may be also associated with poorer survival of inner hair cells (IHCs), causing disruption in the transmission of neural signals to the central auditory nervous system from the periphery. Restoration of high frequency audibility in such cases might result in undesirable distortion or masking of the speech signal and negatively affect performance.

2.2 Frequency Lowering Signal Processing in Hearing Aids

Frequency lowering signal processing schemes have been developed to address the problems associated with limited high frequency gain and decreased benefit from high frequency audibility in listeners with more severe high frequency hearing losses
The goal of these processing schemes is to shift high frequency sounds in the acoustic input to lower frequency regions where the listener has better residual hearing.

Braida et al. (1979) reviewed several of the early laboratory-based techniques for accomplishing frequency lowering, which were categorized according to six basic types: 1) slow-playback, 2) time-compressed slow-playback, 3) frequency shifting, 4) vocoding, 5) zero-crossing-rate division, and 6) frequency transposition. Speech perception performance by hearing-impaired listeners using these strategies was variable, and temporal and harmonic distortions of the processed speech sounds contributed to decrements rather than improvements in performance for many listeners when compared to standard linear gain amplification. Extensive training with frequency-lowered speech materials was necessary for the subset of individuals who did enjoy modest benefit from use of that technology. Individuals who experienced slight increases in speech perception with small shifts in frequency typically experienced significant decrements in performance and reported unacceptable sound quality when greater compression ratios or frequency shifts were applied. Taken together, these early works indicated that a subset of hearing-impaired listeners could potentially benefit from conservative frequency lowering signal processing and efforts to develop wearable frequency lowering hearing aids continued.

Early hearing aids that were capable of lowering high frequency sounds typically used transposition techniques, which involve linear shifts of high frequency speech sounds by a fixed step (in Hz) which are then mixed with unaltered low frequency bands. The first of these aids was a body-worn device, the Oticon TP72 (Johansson, 1961; Wedenberg, 1961). This device extracted temporal modulations at frequencies between 3 and 6 kHz and applied these modulations to a lower frequency broadband noise, which
was mixed in the output with unaltered low frequency sounds. Listener performance with this device was mixed (Johansson, 1966; Ling, 1968; Foust and Gengel, 1973). Studies using later transposition aids such as the AVR TranSonic and ImpaCt, which used manipulations of sampling rates to shift high frequency sounds, also failed to show consistent benefit (Davis-Penn and Ross, 1993; Parent, Chimiel and Jerger, 1997; McDermott et al. 1999; McDermott and Knight, 2001). Nonetheless, Kuk et al. (2009) and Auriemmo et al. (2009) reported some improvements in fricative perception with a current generation linear transposition hearing aid (Widex) following training.

Another frequency lowering technique used in current hearing aids is nonlinear frequency compression. This technique is in contrast to the proportional, whole band frequency compression used in earlier works (e.g. Turner and Hurtig, 1999), where all frequencies in the sound input were lowered by a constant ratio. As implemented in current generation Phonak hearing aid fitting software, frequency compression (referred to as SoundRecover™) is characterized by a “knee point”, or start frequency above which compression occurs, and a compression ratio, the amount of compression applied. Using this method, sounds in the low frequency band, below the start frequency, are unaltered and high frequency sounds are not mixed with the low frequency band in the output, as is the case with transposition aids. Figure 1 shows spectral input output functions for various frequency compression ratios with a single start frequency for a nonlinear frequency compression algorithm described by Simpson, Hersbach, and McDermott (2005), as adapted by the author. Knee point and compression cannot be selected independently in clinical programming software, though some investigators (Glista et al. 2009) have used research programming interfaces capable of independent knee point and compression ratio adjustments. Clinicians may also select a default frequency compression setting determined by the fitting software based on the individual hearing impaired individual’s thresholds.
Studies to date of the effectiveness of nonlinear frequency compression have differed in their conclusions. This may be due, in part, to differences between studies in 1) the protocols used to determine individual frequency compression settings, 2) the degree and configuration of hearing loss of the subjects, 3) ages of the participants (adults vs. adolescents vs. younger children), 4) the types of speech perception tests used as outcome measures, and 5) the length of participants’ pre-test familiarization period with frequency compression sound processing. Several longitudinal studies (Glista et al. 2009; Bohnert, Nyffeler, and Keilmann, 2010; Wolfe et al. 2010 and 2011) indicate that some but not all hearing aid users show improvements in the perception of consonant sounds (especially high frequency sibilants such as /s/, and /ʃ/) with nonlinear frequency compression enabled after an acclimatization period. Glista et al. (2012) found that hearing-impaired children (ages 11-18) showed varied time courses of realization of benefit (or lack thereof) in speech perception of sibilants (/s/ and /ʃ/) from use of nonlinear frequency compression hearing aids. An acclimatization period prior to full benefit from frequency compression is suggested by these longitudinal studies. However, possible maturational influences on speech test performance in hearing-impaired in children, as noted by Alexander (2013), may also have contributed to gains in test performance. It should also be noted that audibility of live-voice presentations of the phonemes (/s/ and /ʃ/) guided the nonlinear frequency adjustments used with the participants of some of these studies (Glista et al. 2009, 2012). It is, therefore, not surprising that modest gains in perception of this limited subset of speech sounds was observed.

Others have reported that while many individuals benefit from use of nonlinear frequency compression, a subset of adult listeners with high frequency SNHL may experience significant decrements or fail to show significant improvements in word recognition (Simpson, Hersbach, and McDermott, 2005). Simpson, Hersbach, and McDermott (2006) found no benefit, on average, in a group of adult listeners with steeply
sloping hearing loss, and suggested a possible influence of severity and configuration of hearing loss on performance with nonlinear frequency compression. Hillock-Dunn et al. (2014) found no benefit of nonlinear frequency compression compared to conventional sound processing for perception of consonant-vowel syllables and spondees in noise in school-age hearing aid users. It is noteworthy that participants in this study had extensive experience with nonlinear frequency compression in their personal hearing aids (ranging from approximately 1 month to nearly 3 years), suggesting that the overall lack of benefit is not attributable to an insufficient acclimatization period.

The participants in these aforementioned studies were adults and older children (ages 6-17 years for Glista et al. 2009; ages 5-13 years for Wolfe et al. 2010, 2011; ages 11-18 years for Glista, 2012; ages 9-17 for Hillock-Dunn et al. 2014) who were typically capable of providing feedback about the audibility and sound quality of high frequency speech sounds as part of the fitting process. Relatively few studies have examined the impact frequency compression hearing aids in pre-school aged children (<5 years old). Ching et al. (2013) randomly assigned 2-3 year olds with early-identified hearing loss (mean age of first hearing aid fitting to a controlled trial of conventional processing (n=19) or nonlinear frequency compression aids (n=25) with a double-blinded design (patient’s families and examiners were blinded to the hearing aid processing condition). Performance on standard speech production, receptive language, and expressive language tests was then measured after a trial period (mean duration of 7.7 months) at approximately three years of age. It was found that receptive and expressive language scores were significantly higher in the nonlinear frequency compression group but that receptive vocabulary and consonant articulation scores were significantly lower compared to the conventional processing group. Speech error analysis found a significant increase in substitution of affricate sounds for sibilants in the nonlinear frequency compression group. A multivariate regression analysis (including gender, severity of hearing loss, and maternal education as predictors of performance at age 3) found no
significant effect of hearing aid signal processing condition in global language ability scores.

It is clear then that the use of frequency compression is not always beneficial for the listener and that benefit in perception of high frequency consonant sounds can be highly variable for individual users. Perreau, Bentler, and Tyler (2013) reported significantly *poorer* vowel and spondee-in-noise performance in adult listeners with moderate to severe hearing loss using frequency compression hearing aid signal processing compared to conventional hearing aid settings after two months of use. This finding was in contrast to the lack of significant effects of frequency compression on vowel recognition reported by Glista et al. (2009). There are multiple reasons for observed decrements in vowel recognition performance. Unlike proportional frequency compression strategies (Turner and Hurtig, 1999), nonlinear frequency compression strategies used in hearing aids also have the potential to alter the ratio of formant peaks frequencies, an important cue for vowel identification (Peterson and Barney, 1952). Shifts in the frequency of vowel formant peaks can decrease listeners’ ability to perceive separate formants (Chistovich and Lublinkskaya, 1979). The reported nonlinear frequency compression start frequencies in Glista et al. (2009) were higher, on average, than those of Perreau, Bentler, and Tyler (2013), where a majority of the subjects were fit with the lowest available frequency compression knee point (1.5 kHz), which is below the typical second formant frequency for some vowels. As a result, the frequency compression implementation in Perreau, Bentler, and Tyler (2013) may have caused shifts in formant ratio or other distortions that negatively impacted performance.

Given the demonstrated potential for lack of benefit or even for measurable decrements in speech perception performance with frequency compression in some hearing aid users, an experimentally validated method for selecting optimal frequency compression parameters would be of great value in clinical practice. Audiogram-based prescriptions of nonlinear frequency compression have been proposed. Values for knee
point and compression ratios can be set using the SoundRecover Fitting Assistant software designed by Alexander (2009). Using a fuzzy logic model, this algorithm generates an estimate from a range of knee point values and compression ratios input by the audiologist that maximizes the contribution of high frequency audibility to speech perception for a given audiogram. McCreery et al. (2013) reported that this fitting method resulted in improved speech sound identification compared to manufacturer default frequency compression settings and conventional sound processing in normal hearing listeners with simulated high frequency hearing loss. Frequency compression in the McCreery et al. (2013) study was implemented using a PC-based hearing aid simulator and the nonlinear frequency compression algorithm described by Simpson, Hersbach, and McDermott (2005). Using this same stimulus processing method McCreery et al. (2014) found a significant improvement in recognition of words (chosen with emphasis on fricatives and affricate phonemes) with SoundRecover-optimized frequency compression in hearing-impaired adults and children when compared to conventional sound processing. However, the word recognition stimuli in the conventional sound processing condition in this study were low-pass filtered at 5000 Hz to simulate a maximum audible frequency with a Phonak hearing aid. The authors indicated that actual audibility with a wearable hearing aid might extend above this frequency in some hearing-impaired listeners, such that their results would tend to overestimate gains in audibility with nonlinear frequency compression for these users.

These results provide some evidence that audiogram-based methods may be a viable alternative to reliance on using behavioral tests of audibility of phonemes /s/ and /ʃ/ to guide fitting of nonlinear frequency compression. Nonetheless, Hillock-Dunn (2014) used SoundRecover Fitting Assistant to set nonlinear frequency setting for their participants and found no improvement compared to conventional hearing aid processing. Perreau, Bentler, and Tyler (2013), who observed decrements in vowel and spondee performance with nonlinear frequency compression, also used SoundRecover Fitting
Assistant to program knee point and compression ratio settings for their subjects. Participants in this study tended to have poorer audiometric thresholds than those who participated in the McCreery et al. (2014) study. It may be the case that frequency compression settings necessary to maximize the audible bandwidth in individuals with severe hearing loss cause excessive distortion in the compressed band, contributing to poorer performance. The validity of the assumption underlying SoundRecover Fitting Assistant, which is that maximized audible high frequency bandwidth should result in improved performance, merits further investigation.

2.3 Electrophysiology in Hearing Aid Fitting

Audiometric thresholds for infants, younger children and individuals with multiple disabilities are often estimated from electrophysiological thresholds. For example, auditory brainstem responses (ABRs) have been used for many years in the objective assessment of hearing sensitivity in infants (Hecox and Galambos, 1974; Galambos and Despland, 1980; Sininger, Abdala, and Cone-Wesson, 1997; Stapells & Oates, 1997). Estimates of behavioral threshold from the ABR are not without error, particularly for frequencies above and below 2-4 kHz (Gorga et al. 1985, 2006; Oates and Stapells, 1998; Stapells, 2000). More recently the auditory steady state response (ASSR) has been used to estimate audiometric thresholds in infants and young children. Studies have shown that ASSR-based estimates of audiometric threshold also may differ considerably from “true” audiometric thresholds measured using standard behavioral techniques (Stueve and O’Rourke, 2003; Sininger, Abdala, and Cone-Wesson, 1997; Stapells & Oates, 1997; Herdman and Stapells, 2001; Cone-Wesson et al., 2002; Vander Werff et al. 2002). Under- or over-estimation of behavioral thresholds from physiological responses (error may exceed ± 10 dB) in young hearing-impaired listeners may interfere with the application of audiogram-based frequency compression fitting algorithms previously described.
In the early years after ABR was first described there were attempts to use the ABR to verify audibility with personal amplification in this population (Kiessling, 1982; Gorga, Beauchaine, and Reiland, 1987). Unfortunately, stimuli like clicks or brief tone bursts that are optimal for evoking an ABR often result in significant output distortion when processed by a wide dynamic range compression hearing aid, especially a hearing aid with higher gain characteristics (Garnham et al. 2000). Aided ASSR threshold and aided behavioral thresholds also may differ (Picton et al., 1998; Stroebel, Swanepoel, and Groenewald, 2007), and artifacts may be mistaken for responses when high level stimuli are used to elicit ASSRs (Small and Stapells, 2004). ASSRs are also small in amplitude compared to the ABR, which has been a concern when attempting to estimate aided thresholds from ASSRs in listeners with severe and profound hearing losses wearing wide dynamic range compression hearing aids.

Cortical auditory evoked potentials have been used with some success to estimate audibility for hearing-impaired individuals who use conventional hearing aid technology (Korczak, Kurtzberg, and Stapells 2005; Golding et al. 2007). CAEPs have advantages over ABRs or ASSRs in that the responses can be evoked using a variety of naturally produced speech stimuli and complex non-speech sounds that can be processed with reasonable fidelity by the hearing aid (Ostroff, Martin and Boothroyd, 1998; Tremblay et al. 2003). CAEPs can be recorded in a passive listening paradigm, allowing for application to younger children and infants (Golding et al. 2007; Carter et al. 2010), and the presence of aided speech-evoked CAEPs in infants has been found to be positively correlated with higher scores on parental questionnaires of aided listening performance (Golding et al. 2007).

Several different CAEPs have been described in the literature. To date, most published research has focused on the P1-N1-P2 complex. This evoked potential is recorded following the onset of an acoustic stimulus and is characterized by a series of positive and negative peaks, typically labeled P1, N1 and P2, recorded within a latency
range of about 50 and 300 milliseconds post stimulus onset. A second type of cortical evoked potential, the auditory change complex (ACC), has also been described (Ostroff, Martin and Boothroyd, 1998; Martin and Boothroyd, 2000). The ACC is recorded in response to spectral and/or intensity changes in ongoing stimuli and has morphology similar to the onset P1-N1-P2 response, occurring at an equivalent latency following a change in the stimulus. It has been demonstrated that the ACC response may be recorded in response to contrasting speech sounds in infants as young as 4 months (Small and Werker, 2012).

Both evoked potentials (the P1-N1-P2 complex and the ACC) may have advantages relative to the ABR and the ASSR for assessing benefit from amplification in that they can both be measured using longer duration and/or naturally produced speech stimuli. The ACC may allow for comparison to behavioral speech perception and discrimination tasks, which have been reported to show decrements in performance when higher frequency compression ratios are used (Turner and Hurtig, 1999). Furthermore, discrimination thresholds for ACC responses elicited by spectrally contrasting noise stimuli, often referred to as spectral ripples, correlate well with behavioral discrimination thresholds elicited using identical stimuli (Won et al. 2011). Both speech and complex non-speech stimuli might be processed with different frequency compression signal processing settings to evaluate the effects of these settings on ACC response morphology.

P1-N1-P2 responses have been successfully recorded in children wearing frequency compression hearing aids. Glista et al. (2012) recorded CAEPs from five adolescent hearing aid users with sloping high-frequency hearing loss wearing Phonak Naida IX SP behind-the-ear hearing aids. A direct audio input presentation of 2- and 4 kHz tone burst stimuli was used to elicit onset CAEPs with frequency compression applied in one condition and disabled in the other. Knee points of frequency compression ranged from 1.6-2.2 kHz; compression ratio was fixed across individuals at 4:1. Results showed that replicable N1-P2 responses could be recorded in children wearing frequency
compression hearing aids. Enhanced detectability of the CAEP in response to the 4 kHz tone burst was noted with frequency compression enabled. This report did not, however, specifically address the application of the CAEPs to the evaluation of multiple frequency compression conditions, or attempt to compare results to speech perception performance.

While it has been demonstrated that it is feasible to record cortical AEPs from hearing aid users, including those using frequency compression hearing aids, the interpretation of these results may not straightforward. For example, studies have indicated that amplitude and latency of aided N1-P2 responses are proportional to the signal-to-noise ratio (SNR) of the hearing aid output rather than absolute stimulus intensity levels (Billings et al. 2007, 2009, 2011, 2012). The authors of these studies indicated the need to consider the effects of hearing aid signal processing, which may be reflected in the output SNR, in the interpretation of aided cortical evoked potentials. It is not known how the choice of frequency compression parameters such as the knee point and/or the compression ratio might influence output SNRs or impact CAEPs. As much as possible, then, it would be desirable to control for the effects of various hearing aid characteristics, such as gain and system noise, when examining the effects of frequency compression on physiological and psychophysical responses. That information is necessary if the N1-P2 response and the ACC are to be used in the selection of hearing aid frequency compression parameters.

2.4 Discrimination of Spectrally Complex Non-speech Sounds

Ripple spectrum stimuli were first used to assess discrimination of spectral peaks in speech-like acoustic signals in normal hearing listeners (Supin et al. 1994, 1997). The acoustic frequency spectrum of this type of noise is often described as having a series of peaks and troughs resembling ripples and therefore is often referred to as ripple noise. The starting ‘phase’ of the spectral ripple can be shifted (or inverted) to create spectral contrasts. As the separation of the center frequencies of spectral peaks becomes smaller,
listeners experience increasing difficulty distinguishing standard (sine) vs. inverted (cosine) phase stimuli. **Figure 2** shows examples of spectra for representative standard and inverted ripples.

Ripple stimuli have been used to assess spectral-peak discrimination ability in listeners with a range of different hearing losses, including hearing aid and cochlear implant users (Henry and Turner, 2003; Won et al. 2007; Anderson et al. 2011). Anderson et al. (2011) found statistically significant correlations of discrimination thresholds for 2 octave bandwidth ripple stimuli (amplitude spectra consisting of a 1 octave pass band and ½ octave raised cosine high and low pass slopes) and spatial tuning curve bandwidth, a more direct but time-consuming measure of spectral discrimination in cochlear implant users. Ripple discrimination thresholds have been found to be both and stable and reliable measure, based on test-retest analyses (Won et al. 2007; Anderson et al. 2011).

Discrimination of spectral peaks in ripple stimuli has been hypothesized to correlate with identification of important spectral cues in speech such as vowel formants, though reports differ on the strength of the relationship of ripple discrimination threshold and measures of speech perception. Henry and Turner (2003) found a significant correlation of ripple discrimination resolution and vowel recognition in CI users. Henry, Turner, and Behrens (2005) reported significant correlations of ripple discrimination with vowel and consonant recognition in quiet when data was pooled across normal hearing, hearing-impaired, and cochlear implant subjects: within-group correlations for hearing-impaired and cochlear implant subjects were significant but less robust. Won et al. (2007) reported significant correlations of ripple threshold with word recognition in quiet and speech reception threshold (in dB SNR) in multi-talker babble and steady-state noise backgrounds for normal hearing individuals listening to vocoded stimuli. Anderson (2011) reported a significant correlation of ripple discrimination threshold with sentence recognition scores in quiet with cochlear implant users, though correlations of ripple
discrimination threshold with vowel recognition in quiet, sentence recognition in noise, and vowel recognition in noise were not found to be significant. The authors of this last study attributed these apparent discrepancies to lack of statistical power, given the small sample size used (N=15) compared to the number of participants (N=31) of Won et al. (2007).

Ripple discrimination thresholds have been shown to be sensitive to changes in the spectral detail provided by auditory prostheses such as cochlear implants. Henry and Turner (2003) found that ripple discrimination in CI users improved when the number of independent spectral channels was increased, with performance reaching an upper limit at 4-6 channels. Normal hearing participants listening to CI simulations continued to show improvements in ripple discrimination when the number of channels increased from 1 to 16 channels. Significant improvements in spectral ripple discrimination in the absence of significant changes in speech perception have been reported when cochlear implant signal processing approaches designed to enhance spectral representation are used (Berenstein et al. 2008; Drennan et al. 2010). These findings suggest that ripple discrimination threshold may be more sensitive to changes in the spectral representation of stimuli by auditory prostheses than speech perception measures. It might then be presumed that, likewise, any changes in the spectral representation of high frequency sounds caused by nonlinear frequency compression may result in larger effects on ripple discrimination threshold than speech recognition measures.

As previously discussed, spectral ripple stimuli have been used in electrophysiology experiments in an ACC paradigm (Won et al. 2011), allowing for comparison to behavioral ripple discrimination. In that study, the number of independent vocoder channels was varied in a cochlear implant simulation presented to normal hearing listeners; ripple discrimination threshold and cortical ACC responses were elicited by stimuli 4 seconds in duration with either a standard-inverted ripple phase change or a standard-standard stimulus with no change in ripple phase at 2 seconds post-
stimulus onset. Behavioral thresholds were observed to improve and cortical change response amplitudes for the standard-inverted stimuli increased respectively with the introduction of additional independent channels in the vocoder simulation. Further, ACC amplitudes were reported to be significantly correlated to the behavioral discriminability of the ripple stimuli (represented as a d-prime value) across the vocoder channel conditions in individual subjects. These results indicate that both physiological and psychophysical measures of ripple discrimination are sensitive the degree of spectral fidelity preserved in signal processing manipulations and therefore may be useful for the purpose of evaluating effects of various nonlinear frequency compression settings on the discriminability of spectral peaks. The reported correlation of these measures indicates that ACC responses might be used to estimate ripple discrimination performance in individual listeners under conditions of varying frequency compression ratios.

In summary, frequency lowering techniques have been used in hearing aids for over two decades for the purpose of enhancing the audibility of high frequency speech sounds. One of these methods, nonlinear frequency compression, has been shown to result in improvements in the perception of a subset of high-frequency speech sounds, though it may result in loss of spectral contrast and poorer speech perception performance in a subset of users. Both behavioral and audiogram-based methods of selecting appropriate nonlinear frequency compression parameters have been used, though these methods may not be ideal in infants and other hearing aid users. While electrophysiological measures such as the ABR and ASSR have significant disadvantages for assessments of aided benefit in these populations, CAEPs appear to be well-suited to this application. Specifically, the ACC may be ideal for evaluating the effects of frequency compression on the discriminability of high-frequency spectral contrasts. It is the goal of the following work, then, to investigate whether the ACC is sensitive to changes in frequency compression settings.
Figure 1. Input-output functions of various nonlinear frequency compression ratios. The compression algorithm used to produce this function was that described by Simpson, Hersbach, and McDermott (2005). Knee point, or start frequency of compression is fixed at 1500 Hz in this example.
Figure 2. Ripple spectra for standard (sine) and inverted (cosine) ripple stimuli. The bandwidth of the ripple in this example is 1500-6000 Hz; ripple density is four ripples per octave. A raised cosine window with a 1 octave plateau and ½ octave high- and low-pass bands was applied to the ripple spectra, similar to that described by Anderson et al. (2011).
CHAPTER 3
METHODS

3.1 Overview and Participant Information

Experiment 1 involved the participation of ten (N=10) normal hearing adults (9 F, 1 M, Mean age=23.8, SD = 3.7, Range = 19-32). Normal hearing was defined as air conduction thresholds ≤ 15 dB HL for octave intervals from 250 to 8000 Hz. Experiment 2 involved the participation of ten (N=10) adults (5 F, 5 M, Mean age=54.7, SD = 9.0, Range = 38-65) with sloping sensorineural hearing loss. For this group of study participants, audiometric thresholds for air conduction stimuli were measured at octave intervals from 250 to 8000 Hz and at 6000 Hz using standard clinical procedures. Based on their audiometric thresholds, all ten of the hearing impaired participants were candidates for hearing aids. However, one participant declined to use hearing aids, another was in the process of undergoing an evaluation for his first hearing aid, and another was a former (but not current) hearing aid user. Figure 3 shows mean and individual audiometric thresholds for this hearing impaired group. Table 1 provides greater detail about the individuals who took part in Experiment 2. One participant (HI5) was unable to complete the study protocol due to fatigue and was replaced by another hearing impaired listener (HI11).

All subjects were native speakers of English and had negative histories for neurological or cognitive impairment. Initially, consent was obtained and audiometric testing was completed. The study protocol had three components: perceptual testing using speech, perceptual testing using ripple noise stimuli, and evoked potentials testing, details of which are provided below. Completion of the full protocol required approximately 5-6 hours. Breaks were provided as needed and in some cases the results were obtained at two or more appointments. Study participants were compensated for their participation and all of the procedures were approved by the University of Iowa Institution Review Board.
3.2 ACC Stimuli

The stimulus used to elicit the ACC was a spectral ripple contrast. An advantage of ripple stimuli is that they can be designed so that energy in the signal occurs only above the knee point of typical hearing aid frequency compression, which was 1.5 kHz in these experiments. Each half of the ripple contrast stimulus consisted of a broadband noise, 400 ms in length, constructed from 400 summed sinusoids with frequencies distributed logarithmically between 1500 and 6000 Hz; the spectrum of the broadband stimulus was then modulated (maximum modulation depth of 30 dB) using a rectified sinusoidal spectral envelope with sine or cosine starting phase. The ripple contrast stimulus was created by concatenating the leading sine-phase ripple with its cosine counterpart. Figure 4 shows examples of spectrograms for the ripple contrast stimulus. To mitigate spurious spectral “edge effects” on ripple threshold observed in pilot studies, which tended to elevate (improve) behavioral discrimination thresholds for the high frequency ripple stimuli, the spectrum of each ripple stimulus was filtered using a modified raised sine window (symmetrical in log frequency) with a 1 octave plateau and ½ octave ramps. A single ripple density, 0.5 ripples per octave (rpo), was used for the ACC experiment. This single value was selected to decrease test time (compared to a physiological threshold procedure, where recordings would be made at multiple ripple densities in each frequency compression condition) and because pilot data indicated that ripple contrasts of this density (0.5 RPO) are discriminable for normal hearing and hearing impaired listeners.

3.3 Electrophysiology Procedures

ACC responses were recorded from each subject using the vowel contrast and ripple contrast stimuli presented monaurally via an ER-3A insert earphone at 70 dBA peak level. The non-test ear was plugged. Stimuli were presented with a 1.2 second
interstimulus interval. The test ear was counterbalanced across subjects. During testing, study participants sat in a reclining chair and were encouraged to read, play with an iPad, or watch captioned videos in order to stay alert.

Surface electrodes were applied to the scalp on the side of the head contralateral to the test ear. A ground electrode was placed off center on the forehead and EEG activity measured differentially from six separate recording channels: vertex (+) to contra mastoid (-), temporal lobe (+) to contra mastoid (-), high forehead (+) to contra mastoid (-), vertex (+) to occiput (-), temporal lobe (+) to occiput (-), and high forehead (+) to occiput (-). Eyeblinks were monitored using two additional electrodes placed above and lateral to the stimulus-contralateral eye; sweeps with excessive eye blink contamination were rejected online. Consistent with pilot data where the greatest onset and ACC amplitudes were consistently observed in the vertex to contralateral mastoid channel, subsequent analyses were applied to this channel only. Responses were band-pass filtered between 1 and 30 Hz and amplified (gain=10,000) prior to averaging using custom LabView software. Each stimulus was presented 200 times, in blocks of 100 sweeps. Stimulus conditions were randomized and counterbalanced across the recording session to avoid effects of adaptation. Responses were averaged together prior to off-line analysis using MATLAB. Peaks were selected using an automated, custom MATLAB algorithm, with corrections made as necessary to picked peaks by a trained examiner.

ACC amplitudes were measured offline between N1 and P2 and were normalized to stimulus onset amplitude, allowing for an internally consistent metric of relative amplitude. Absolute amplitudes of ACC responses may be more susceptible to changes in patient state or other conditions of the experimental apparatus across trials.

3.4 Behavioral Assessment Procedures

The perceptual consequences of frequency compression were assessed in multiple ways. First, closed set vowel identification was measured across conditions of frequency
compression using a ten alternative forced choice procedure. The vowel stimuli consisted of medial vowels (e.g. /hVd/ for the ten English vowels /i/, /u/, /ai/, /ei/, /ʌ/, /ɛ/, /a/, /ɔ/, /ɪ/, and /ʊ/) recorded from ten adult female talkers (Hillenbrand et al., 1995), for a total of 100 stimuli per condition. Second, closed set consonant identification was measured using a test developed by Turner et al. (1995). This test consists of sixteen consonant sounds, in an /aCa/ context, produced by two adult males and two adult females for a total of 64 trials per condition. Stimuli for each of these tests were presented and participant responses were registered using custom MATLAB software in conjunction with PsyLab (Hansen, 2006) experimental interfaces. Participants indicated the perceived vowel using a touch sensitive monitor and graphic user interface. Stimulus intensity was calibrated at a peak presentation level of 70 dBA. As with the ACC task, stimuli were presented monaurally using an ER-3A insert earphone, with the non-test ear plugged. With the exception of a brief familiarization with the task, no feedback on performance was given during testing.

Speech perception in noise for these same frequency compression conditions was evaluated using the QuickSIN test (Killion et al. 2001). The test consists of key word identification from sentences spoken by an adult female talker in multi-talker babble background noise. Two lists, selected at random, were used to estimate the signal to noise ratio at which performance was approximately 50% correct (aka ‘SNR-50’) for each frequency compression condition. Choice of this particular test was motivated by the results of Ellis and Munro (2013), who found significant negative effects of frequency compression (using a knee point of 1600 Hz and multiple frequency compression ratios) on speech in noise performance in normal hearing individuals listening to IEEE sentences from which the QuickSIN test was derived.

Another method of assessing how frequency compression alters perception was behavioral spectral ripple discrimination. Spectral ripple discrimination thresholds (determination of the maximum number of spectral peaks per octave for which the
listener can discern standard vs. inverted spectral ripples) were measured using a standard, adaptive, three interval, forced-choice (3AFC) procedure (Levitt, 1971) comparable to that described in earlier works (Henry, Turner, and Behrens, 2005). A ±1 dB level rove was used across the three intervals to control for possible loudness difference cues. Threshold was estimated from the mean ripples per octave of the last four reversals in a run. Three runs were completed per compression condition and threshold was calculated as the mean of the three runs. Again, stimuli were presented and participant responses were registered using custom MATLAB software and the PsyLab interface. Frequency compression conditions applied to the raw stimuli were the same as those used in the electrophysiology task. Stimulus intensity was calibrated for a peak presentation level of 70 dBA. All stimuli were presented monaurally with an ER-3A insert earphone. Response feedback was enabled throughout the test interval.

3.5 Frequency Compression

Manipulation of frequency compression parameters, while simultaneously controlling for circuit noise and the effects of wide dynamic range amplitude compression, was accomplished using PC signal processing software. Four experimental frequency compression conditions were created. Using custom MATLAB software, the raw sound files used for each experimental task (.WAV format, 44100 Hz sampling rate, 16 bit) were segmented (using a moving, 882 point, raised cosine window), filtered (using a Fast Fourier Transform (FFT) filter with 300 dB attenuation slopes within the bandwidth of single FFT bin) into low pass and high pass components, and converted by FFT to the frequency domain. Frequency compression was applied to the high pass input frequency vector with the compression function used by Simpson, Hersbach, and McDermott (2005) in their wearable experimental device:
\[ F_{out} = \begin{cases} F_{in}, & F_{in} < F_{knee} \\ \frac{1}{F_{knee}^{1-CE}} \times F_{CE}^{CE}, & F_{in} > F_{knee} \end{cases} \]

No compression was applied to the low frequency component of the input segment. The low pass component and compressed high pass were then converted to the time domain using and inverse FFT and summed. The next segment was likewise compressed and summed with the preceding compressed segment using a half-overlap increment. For this study, start frequency was fixed at 1500 Hz, defining the boundary for the low pass and high pass segments, for each condition, to coincide with the lowest typically used in commercial frequency compression hearing aids. The five signal processing conditions were defined by the compression exponents applied in the compression function: 1, 0.824, 0.71, 0.56, and 0.465. In terms of compression ratio of the high pass component, these exponents correspond to ratios of 1:1 (no compression), 1.5:1, 2:1, 3:1, and 4:1, respectively. These ratios were selected to encompass the range of compression ratios used in commercial frequency compression aids.

3.6 Gain Prescription

In addition to the nonlinear frequency compression manipulations described above, individually prescribed gain settings were applied to the experimental stimuli for the hearing-impaired participants in Experiment 2. Each participant’s gain profile was calculated by subtracting DSL i/o (Scollie et al., 2005) audibility targets for a normal hearing listener (0 dB HL thresholds, 250-8000 Hz) for a 65 dB SPL speech stimulus from targets calculated based on the hearing-impaired subject’s own audiometric thresholds. These calculated gain settings were implemented using an audiometer (Grason-Stadler GSI-61) to adjust overall level in line with a multi-band digital programmable equalizer (Alesis DEQ 830) to create band-specific gain adjustments.
Figure 3. Individual thresholds of the hearing impaired participants of Experiment 2. The solid gray line shows mean thresholds.
<table>
<thead>
<tr>
<th>Subject ID</th>
<th>Age</th>
<th>Gender</th>
<th>Age at which hearing loss was identified (years)</th>
<th>Duration of HA use (years)</th>
<th>HA Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>HI1</td>
<td>64</td>
<td>F</td>
<td>55</td>
<td>Y, 8 years</td>
<td>BTE (AU)</td>
</tr>
<tr>
<td>HI2</td>
<td>40</td>
<td>F</td>
<td>40</td>
<td>Y, HA eval(^1)</td>
<td>-</td>
</tr>
<tr>
<td>HI3</td>
<td>38</td>
<td>M</td>
<td>~2-4</td>
<td>Y, &lt;1 year</td>
<td>BTE (AU)</td>
</tr>
<tr>
<td>HI4</td>
<td>58</td>
<td>M</td>
<td>44</td>
<td>No</td>
<td>-</td>
</tr>
<tr>
<td>HI6</td>
<td>58</td>
<td>M</td>
<td>28</td>
<td>Y, ~1 year(^2)</td>
<td>ITE (AD)</td>
</tr>
<tr>
<td>HI7</td>
<td>59</td>
<td>M</td>
<td>45</td>
<td>Y, 6 years</td>
<td>BTE (AU)</td>
</tr>
<tr>
<td>HI8</td>
<td>54</td>
<td>F</td>
<td>16</td>
<td>Y, 20 years</td>
<td>ITE (AS)</td>
</tr>
<tr>
<td>HI9</td>
<td>55</td>
<td>M</td>
<td>48</td>
<td>Y, 1 year</td>
<td>BTE (CROS)</td>
</tr>
<tr>
<td>HI10</td>
<td>56</td>
<td>F</td>
<td>46</td>
<td>Y, &lt;1 year</td>
<td>BTE (AU)</td>
</tr>
<tr>
<td>HI11</td>
<td>65</td>
<td>F</td>
<td>61</td>
<td>Y, 3 years</td>
<td>BTE (AU)</td>
</tr>
</tbody>
</table>

**Table 1.** Participant information for Experiment 2

\(^1\)Subject HI2 was undergoing a hearing aid evaluation concurrent with participation in this study.

\(^2\)Subject HI6 previously wore a ITE hearing aid for approximately 1 year, but had discontinued use at the time of his participation in this study.
Figure 4. Spectrogram of ripple contrast stimulus. 0.5 ripples per octave. 800 ms duration.
CHAPTER 4
RESULTS

4.1 Experiment 1 Results

Planned analyses for the data in Experiment 1 included univariate repeated measures ANOVAs with post-hoc paired t-tests with frequency compression ratio as the within subjects main effect of interest. P-values were adjusted for multiple comparisons according the Holm-Bonferroni method. Results for each frequency compression condition were compared to the non-compressed condition as baseline performance on each task.

Analysis of the vowel perception data revealed no effect of frequency compression ratio on identification score in percent correct (F(4,36)=0.2278, p=0.921). Figure 5 shows group results for this measure. Performance was generally high on this task and remained so across the frequency compression conditions. This lack of significant effects of frequency compression ratio on vowel perception was unexpected, though previous reports of deleterious effects of frequency compression on vowel identification were in individuals with moderate-to-severe sensorineural hearing loss (Perreau, Bentler, and Tyler, 2013), rather than in individuals with normal hearing.

Analysis of the consonant perception data revealed a significant main effect of frequency compression condition on identification score in percent correct (F(4,36)=11.355, p < 0.001). Figure 6 shows group results for this measure. Post-hoc paired t-tests indicated that consonant identification score was significantly poorer in the 3:1 (t(9) = 3.0928, p-value = 0.03861) and 4:1 (t(9) = 7.0716, p-value < 0.001). This result was consistent with our hypothesis that greater frequency compression ratios (e.g. 3:1 and 4:1) would have a negative impact on consonant perception. As may be observed in derived consonant confusion matrices for select compression conditions (Appendix A) errors were not randomly distributed; rather, it was primarily the voiceless fricatives that were incorrectly identified.
Figure 7 shows the effect that frequency compression had on perception of speech in noise as measured using the QuickSIN test. A repeated measures analysis of variance procedure revealed a significant main effect of frequency compression on the SNR-50 ($F(4,36)=17.650$, $p=4.217\times10^{-8}$). Post-hoc paired t-tests indicated that speech identification in noise was significantly poorer in the 2:1 ($t(9) = -3.9545$, p-value = 0.006664), 3:1 ($t(9) = -4.7805$, p-value = 0.003003), and 4:1 ($t(9) = -6.7036$, p-value < 0.001) frequency compression conditions compared to the no compression baseline condition. This result was consistent with our hypothesis that frequency compression would have a negative effect on speech perception in noise for listeners with normal hearing. It is also consistent with the findings of Ellis and Munro (2013), who found significant negative effects of nonlinear frequency compression ratio on normal hearing listeners’ speech perception in noise.

Figure 8 shows the effect that changes in the frequency compression ratio had on behavioral measures of spectral ripple discrimination thresholds. Ripple discrimination thresholds tend to decrease (become poorer) as the amount of frequency compression increases. Results of a repeated measures ANOVA revealed a significant main effect of frequency compression on ripple discrimination threshold ($F(4,36)=51.093$, $p < 0.001$). A Mauchly test revealed that the repeated measures data significantly departed from the sphericity assumption of the ANOVA ($\varepsilon = 0.39746$); therefore, the reported p-value of the repeated measures F-test has been modified using a Greenhouse-Geisser correction to control for possible inflation of the Type 1 error rate. Post-hoc paired t-tests revealed that ripple discrimination threshold was significantly poorer in the 1.5:1 ($t(9) = 4.5914$, p-value = 0.001307), 2:1 ($t(9) = 5.9716$, p-value < 0.001), 3:1 ($t(9) = 7.3274$, p-value < 0.001), and 4:1 ($t(9) = 8.4456$, p-value < 0.001) frequency compression conditions relative to the baseline no compression condition. This result is consistent with our hypothesis that behavioral ripple threshold would decrease with increasing frequency compression ratio in normal hearing listeners.
Figure 9 shows grand mean waveforms for each frequency compression condition based on the ten subjects. Inspection of the mean waveforms allows for visualization of the mean trend in CAEP amplitudes. As compression ratio increases, onset N1-P2 amplitude is stable, but ACC N1-P2 amplitude is noticeably smaller. ACC N1-P2 latency also increases slightly at the higher (3:1 and 4:1) compression ratios. This trend of smaller ACC amplitudes at higher compression ratios, with no effect for mild compression ratios (e.g. 1.5:1), is consistent with our hypothesis for the ACC measure. Analysis of the ACC response data revealed a significant main effect of frequency compression on normalized ACC amplitude (\(F(4,36)=8.1611, p=8.610e^{-05}\)). Post-hoc paired t-tests revealed that normalized ACC amplitude was significantly smaller in the 2:1 (\(t(9) = 2.9376, p\text{-value} = 0.0331\)), 3:1 (\(t(9) = 5.1026, p\text{-value} = 0.002572\)), and 4:1 (\(t(9) = 4.2327, p\text{-value} = 0.006594\)) frequency compression conditions. Figure 10 shows the distributions of normalized ACC N1-P2 amplitudes for this measure. The physiological data show an effect analogous to that of the behavioral ripple discrimination task with increasing frequency compression ratio.

4.2 Experiment 2 Results

As in Experiment 1, planned analyses of the data in Experiment 2 included univariate repeated measures ANOVAs with post-hoc paired t-tests with frequency compression ratio as the within subjects main effect of interest. P-values were adjusted for multiple comparisons according the Holm-Bonferroni method. Results for each frequency compression condition were compared to the non-compressed condition as baseline performance on each task.

Analysis of the vowel perception data revealed a significant main effect of frequency compression ratio on identification score (\(F(4,36)= 5.7493, p=0.02085\)). A Mauchly test revealed that the repeated measures data significantly departed from the sphericity assumption of the ANOVA (\(\varepsilon = 0.38196\)); therefore, the reported p-value of
the repeated measures F-test was modified using a Greenhouse-Geisser correction to control for possible inflation of the Type 1 error rate. **Figure 11** shows group results for vowel identification score, expressed as percent correct. As with the normal hearing listeners, vowel identification scores were excellent in the baseline (no frequency compression condition). Unlike the data for the normal hearing listeners in Experiment 1, post-hoc paired t-tests indicated that vowel identification score was significantly poorer in the 3:1 (t(9) = 2.9858, p-value = 0.04848) and 4:1 (t(9) = 3.1299, p-value = 0.04848) conditions compared to baseline for the hearing impaired listeners. It was hypothesized that vowel identification would be poorer at the highest frequency compression ratios, and these results were consistent with that hypothesis.

Analysis of the consonant perception data revealed a significant main effect of frequency compression condition on identification score (F(4,36)= 11.991, p < 0.001). **Figure 12** shows group results, expressed as percent correct, for the consonant identification task. Similar to the normal hearing listeners, baseline performance on the consonant identification task was excellent (the group mean score was slightly less than 90%). Post-hoc paired t-tests indicated that consonant identification score was significantly poorer in the 2:1 (t(9) =2.9672, p-value =0.03154), 3:1 (t(9) =7.3424, p < 0.001) and 4:1 (t(9) =8.5253, p < 0.001) conditions. It was hypothesized that consonant identification would be significantly poorer in the highest frequency compression ratio conditions in the hearing impaired subjects, and these results are consistent with that hypothesis. As with the normal hearing participants in Experiment 1, derived consonant confusion matrices for select compression conditions (Appendix A) indicated it was primarily the voiceless fricatives that were incorrectly identified.

Analysis of the QuickSIN data revealed a significant main effect of frequency compression condition on estimates of SNR-50 (F(4,36)= 6.2951, p < 0.001). **Figure 13** shows group results for this measure. However, post-hoc paired t-tests indicated that SNR-50 was not significantly different from baseline in any of the non-trivial frequency
compression conditions at an $\alpha=0.5$ level following correction for multiple comparisons. Comparison of these results with the normal hearing data (Figure 7) reveals greater variance across individuals in the hearing impaired listeners than was the case in the normal hearing group. Mean SNR-50 appears to be poorer in the no-compression condition compared to the normal hearing listeners. While speech in noise performance appeared not to differ across conditions significantly at the group level, decrements in SNR-50 from baseline in individual scores exceeded the 2.7 dB 95% critical difference criterion of the QuickSIN (based on two lists per condition) for one subject in the 2:1 compression conditions, three subjects in the 3:1 condition, and five subjects in the 4:1 compression condition.

Analysis of the behavioral ripple data revealed a significant main effect of frequency compression on ripple discrimination threshold ($F(4,36)= 8.1611, p<0.001$). Figure 14 shows group results for this measure, expressed as RPO threshold. Post-hoc paired t-tests revealed that ripple discrimination threshold was significantly poorer in the 2:1 ($t(9) =2.9376, p$-value $=0.0331$), 3:1 ($t(9) =5.1026, p$-value $=0.0026$), and 4:1 ($t(9) =4.2327, p$-value $=0.0066$) frequency compression conditions. These results were similar to those of the normal hearing listeners in Experiment 1 and consistent with our hypothesis that ripple discrimination would be significantly poorer at the highest frequency compression conditions.

Figure 15 shows grand mean waveforms for each frequency compression condition, demonstrating the trend across frequency compression condition. As was the case with the normal hearing listeners in Experiment 1, onset N1-P2 amplitude is stable with increases in frequency compression ratio, but ACC N1-P2 amplitude appears to become smaller. Slight increases in ACC N1-P2 latency were again evident at the higher compression ratio conditions. Analysis of the ACC response data revealed a significant main effect of frequency compression on normalized ACC amplitude ($F(4,36)= 4.4192, p=0.005222$). Post-hoc paired t-tests revealed that normalized ACC amplitude was
significantly smaller in the 3:1 ($t(9) = 3.3928$, $p$-value = 0.031856) but not the 4:1 ($t(9) = 2.4452$, $p$-value = 0.11115) frequency compression condition. Figure 16 shows the distributions of normalized individual ACC amplitudes for this measure. As with the normal hearing listeners in Experiment 1, the trend of smaller ACC amplitudes in the hearing impaired listeners at higher compression ratios is consistent with our hypotheses for the ACC measure.

While the previous figures are useful for identifying mean trends in results, inspection of individual data revealed differing patterns in the ACC response. Figure 17 shows ACC waveforms recorded from two hearing impaired participants HI2 (panel A), and HI4 (panel B). HI2 has a fairly gently sloping high frequency sensorineural hearing loss (see Figure 3). This subject’s AAC responses decrease more or less monotonically with increasing frequency compression ratio. Study participant HI4 has normal hearing in the low frequencies but an audiogram that slopes steeply to a moderate loss at 2 kHz. AAC responses recorded from this study participant showed a non-monotonic pattern with larger ACC amplitude in the 1.5:1 and 2:1 compression conditions (compared to 1:1) followed by decrements in the 3:1 and 4:1 conditions. Figure 18 shows results of behavioral testing from these two individual study participants. HI4 made small improvements over baseline (1:1) in the 1.5:1 compression ratio condition for speech in noise (panel C) and ripple (panel D), but not for vowel identification (panel A) and consonant identification (Panel B). In contrast, HI2 appears to have negligible improvements in speech perception in noise and decrements in ripple discrimination in the 1.5:1 and 2:1 compression conditions.

The inclusion of both behavioral and physiological measures in these experiments allows for their comparison across compression conditions. Figure 19 show scatter plots constructed using normalized ACC amplitudes recorded from individual study participants compared with their individual consonant identification scores (A and B), speech perception in noise results (C and D), and ripple discrimination thresholds (E and
Panels A, C, and E show results obtained from the normal hearing listeners who participated in Experiment 1. Panels B, D and F show results obtained from the hearing impaired listeners who participated in Experiment 2. A statistically significant linear correlation was found between ripple discrimination and normalized ACC amplitude in the normal hearing listeners (E). None of the other correlations were found to be statistically significant at an $\alpha=0.05$ level, even when excluding the outlier in the hearing impaired group (this particular individual had relatively small onset amplitudes, and noise in the 2:1 compression condition may have artificially inflated the resultant normalized amplitude).
Figure 5. Normal hearing group vowel identification scores (N=10). No significant differences in vowel identification score were found across frequency compression conditions. Bold lines represent mean scores; fine lines represent median scores. Whiskers represent 95/5 percentiles.
Figure 6. Normal hearing group consonant identification scores (N=10). Scores were significantly lower in the 3:1 and 4:1 compression conditions. Bold lines represent mean scores; fine lines represent median scores. Whiskers represent 95/5 percentiles.
Figure 7. Normal hearing group speech perception in noise scores (N=10). Scores were significantly poorer in the 2:1, 3:1, and 4:1 compression conditions. Bold lines represent mean scores; fine lines represent median scores. Whiskers represent 95/5 percentiles.
Figure 8. Normal hearing group behavioral ripple discrimination scores (N=10). Scores were significantly poorer in the 1.5:1, 2:1, 3:1, and 4:1 compression conditions. Bold lines represent mean scores; fine lines represent median scores. Whiskers represent 95/5 percentiles.
**Figure 9.** Normal hearing group grand mean waveforms (N=10). N1 and P2 peaks are marked for the onset and ACC response in the 1:1 condition. Decrements in ACC amplitude, and slight increases in ACC N1 latency, may be observed as frequency compression ratio increases.
Figure 10. Normal hearing group normalized ACC response amplitudes (N=10). ACC responses were significantly smaller in the 2:1, 3:1, and 4:1 compression conditions. Bold lines represent mean normalized amplitudes; fine lines represent median normalized amplitudes. Whiskers represent 95/5 percentiles.
Figure 11. Hearing impaired group vowel identification scores (N=10). Vowel identification was significantly poorer in the 3:1 and 4:1 compression conditions. Bold lines represent mean scores; fine lines represent median scores. Whiskers represent 95/5 percentiles.
Figure 12. Hearing impaired group consonant identification scores (N=10). Scores were significantly lower in the 2:1, 3:1, and 4:1 compression conditions. Bold lines represent mean scores; fine lines represent median scores. Whiskers represent 95/5 percentiles.
Figure 13. Hearing impaired group speech perception in noise scores (N=10). No significant differences in performance were found across frequency compression conditions. Bold lines represent mean scores; fine lines represent median scores. Whiskers represent 95/5 percentiles.
Figure 14. Hearing impaired group behavioral ripple discrimination scores (N=10). Scores were significantly poorer in the 2:1, 3:1, and 4:1 compression conditions. Bold lines represent mean scores; fine lines represent median scores. Whiskers represent 95/5 percentiles.
Figure 15. Hearing impaired group grand mean waveforms for ripple contrast stimuli (N=10). N1 and P2 peaks are marked for the onset and ACC response in the 1:1 condition. Decrements in ACC amplitude, and slight increases in ACC N1 latency, may be observed as frequency compression ratio increases.
Figure 16. Hearing impaired group normalized ACC response amplitudes for ripple contrast stimuli (N=10). ACC responses were significantly smaller in the 3:1 compression conditions. Bold lines represent mean normalized amplitudes; fine lines represent median normalized amplitudes. Whiskers represent 95/5 percentiles.
Figure 17. Waveforms for hearing impaired participants HI2 (A) and HI4 (B). N1 and P2 peaks are marked for the onset and ACC response in the 1:1 condition. HI2, who has a gently sloping mild-moderate loss, shows a monotonic decrease in ACC amplitude with increasing compression ratio. HI4, who has normal hearing in the low frequencies steeply sloping to a moderate loss at 2 kHz and above, shows a non-monotonic pattern of larger ACC amplitude in the 1.5:1 and 2:1 compression conditions (compared to 1:1) followed by decrements in the 3:1 and 4:1 conditions.
Figure 18. Comparison of individual scores on behavioral measures for HI2 and HI4. The panels show vowel identification (A), consonant identification (B), speech perception in noise (C), and ripple discrimination (D). Filled circles show scores for participant HI2; open circles show scores for participant HI4.
Figure 19. Relationships of ACC amplitude to behavioral measures. Consonant identification score (A and B), speech perception in noise (C and D), and ripple discrimination threshold (E and F) are shown separately for normal hearing (left column) and hearing impaired listeners (right column). A statistically significant linear correlation was found between ripple discrimination and normalized ACC amplitude in the normal hearing listeners (E).
CHAPTER 5
DISCUSSION

It was a goal of these experiments to investigate whether CAEPs, in particular the ACC, are sensitive to changes in nonlinear frequency compression ratio. Our experimental conditions applied to the stimuli used in this task consisted of five frequency compression ratios (1:1, 1.5:1, 2:1, 3:1, and 4:1) designed to span that which might be implemented in a wearable personal hearing aid. It was not a goal of this study to select or define optimal frequency compression settings on an individual basis, but rather to determine performance over a wide range of settings. Our selection of parameters for our experimental conditions also differs from previous investigations of the effects of nonlinear frequency compression in that typically only a single frequency compression setting was compared to conventional processing. A second goal was to determine whether performance on measures of speech perception (vowels, consonants, and sentences in noise) and behavioral discrimination of spectrally complex non-speech sounds are also affected in these same frequency compression conditions.

Previous work has demonstrated that amplitude of ACC responses elicited by ripple contrast stimuli is reduced in signal processing conditions that reduce spectral contrast, such as in cochlear implant vocoder simulations (Won et al. 2011) with varying numbers of simulated channels or with frequency allocation manipulations in cochlear implant mapping (Scheperle and Abbas, 2014a and 2014b, in Review). In these experiments we demonstrated significant negative effects of increasing frequency compression ratio on ACC amplitude in normal hearing and hearing impaired listeners, consistent with diminished spectral contrast with increasing frequency compression. That loss of spectral contrast is a primary, if not the sole cause of these decrements in ACC amplitude is supported by our finding that behavioral ripple discrimination thresholds were likewise negatively affected by increases in frequency compression in both normal
hearing and hearing impaired listeners. Ripple contrast stimuli like the ones used in this study are appealing in that they can readily be modified to have start frequencies matched to different knee points of compression (1500 Hz was used for these experiments). While the 0.5 RPO density stimulus was sufficient to elicit ACC responses from both normal hearing and hearing impaired listeners across a range of frequency compression settings, it may be the case alternate densities (e.g. 1 RPO, 2 RPO, etc.) or use of a series ripple densities to determine physiologic ripple \textit{threshold} (similar to Won et al. 2011) may also be applied to evaluation of the effects of different frequency compression parameters. It may be that physiological \textit{threshold} for the ACC response to ripple contrasts would serve as a better predictor of performance and would be more strongly correlated with behavioral tasks across frequency compression conditions. This approach would be more time-consuming than one using a single density contrast stimulus, however. There is, therefore, need for further exploration of optimal stimulus characteristics for the purpose of using the ACC response to evaluate nonlinear frequency compression parameters.

Results of the vowel identification task with our hearing impaired participants, which indicated negative effects of combined low start frequency and high compression ratio on performance, are consistent with the findings of Perreau, Bentler, and Tyler (2013), who found significant negative effects of frequency compression (implemented in a commercial hearing aid) on vowel identification in hearing impaired listeners. As stated previously, many of the participants in the Perreau, Bentler, and Tyler (2013) study were fit with frequency compression settings, selected using SoundRecover Fitting Assistant, nearly identical to the maximum frequency compression condition in this study (1.5 kHz knee point, 4:1 compression ratio). This would suggest that those conditions necessary to maximize audibility of high frequency consonant sounds may contribute to deleterious effects on vowel identification in some users. It is, however, noteworthy that similar reduction in vowel identification scores using these same frequency compression conditions was not observed in the normal hearing listeners. This differential effect may
be due to broadening of spectral filters in the periphery in hearing impaired listeners. It has been demonstrated that hearing impaired listeners require greater peak to trough amplitude and/or greater separation of peaks in order to accurately discriminate between different formant distributions and that they need relatively more contrast to perform well as degree of hearing loss increases (Leek, Dorman, and Summerfield, 1987).

Negative effects on consonant identification were found for both normal hearing and hearing impaired listeners at the two highest frequency compression conditions. Other studies (Simpson, Hersbach, and McDermott, 2006) have shown negative effects on consonant identification scores in a subset of listeners where frequency compression settings were adjusted on an individual basis prior to testing. Our experimental design did not include individual adjustment of these parameters. It may be the case that our highest frequency compression conditions (3:1 and 4:1 ratios) were more aggressive than would have been typically prescribed for these listeners (using either the audiogram or live-voice identification of sibilant sounds) contributing to these decrements. This highlights the negative effects on consonant identification of excessive nonlinear frequency compression settings. Clearly, great care should be taken in selecting these setting in young hearing aid users given that “optimal” and “excessive” frequency compression parameters may be more difficult to define in this population.

Significant negative effects on speech recognition in noise were also observed with increasing frequency compression ratio. This result is consistent with the findings of Ellis and Munro (2013), who found significant decrements in performance using similar materials (IEEE sentences) with ratios greater than or equal to 2:1 and a similarly low knee point of frequency compression (1.6 kHz). Though a significant negative main effect on speech in noise performance was observed in our hearing-impaired cohort, differences relative to the baseline (1:1) condition were not significant on the group level. On an individual basis, an increasing number of the hearing impaired listeners tested did show significant decrements in performance as compression ratio increased. This finding
suggests that clinical assessment of speech identification in noise (using a test such as the QuickSIN) is indicated in the evaluation of frequency compression in adult users. It is suggested that using a greater number of lists per frequency compression condition (two lists per condition were used in this study) may improve the sensitivity of this test to differences in listener performance with and without frequency compression or between multiple frequency compression settings.

Inspection of results of selected hearing impaired individuals (HI2 and HI4) revealed possible differences in relationships of ACC amplitude across frequency compression conditions amongst individuals with differing degrees and configurations of hearing loss. HI2, with a comparatively flatter hearing loss, showed only declines in ACC amplitude with increasing compression ratio, suggesting whatever improvements in audibility with increasing compression ratio were offset by loss of spectral contrast. Conversely, HI4, who had normal hearing in the frequencies up to 2 kHz and steeply sloping hearing loss above, showed increases in ACC amplitude with modest frequency compression (1.5:1 and 2:1), consistent with increases in audibility offsetting loss of spectral contrast. While these individual results are not wholly compelling relative to the group results, they are at least consistent with expectations concerning the influence of the audiogram on this measure.

In conclusion, the cumulative results of these experiments demonstrate the following: 1) as hypothesized, the amplitude of ACC response elicited by spectral ripple contrast stimuli is sensitive to frequency compression settings, and 2) that performance on speech perception and ripple discrimination may also be negatively affected as frequency compression ratios increase. While these findings demonstrate a novel application of the ACC response recording to evaluation and (possibly) the selection of nonlinear frequency compression characteristics, they also demonstrate significant negative perceptual consequences of excessive frequency compression parameters, which
is of particular concern as clinicians consider the user of this signal processing option in young hearing aid users.
APPENDIX A: CONSONANT CONFUSION MATRICES

The following are group consonant confusion matrices constructed using the stimulus and response logs of the consonant identification task for the normal hearing and hearing and hearing impaired participants. The stimulus presented constitutes the x coordinate and the subject response the y coordinate. Correct responses fall on the diagonal axis; incorrect responses are placed off this access. The size of the point at each coordinate is proportional to the count for that pairing of stimulus and response. Figure A1 shows results for the normal hearing listeners in the 1:1 compression condition, with errors consisting primarily of confusions for the consonants /f/, /s/, /θ/ (shown as ‘th’), and /ð/. In the 3:1 and 4:1 conditions (Figure A2 and Figure A3) confusions of these consonant sounds and also the voiced fricatives /z/ and /Ʒ/ (shown as ‘zh’) increase. Similar overall patterns of errors are observed in the hearing impaired group in the 1:1 (Figure A4), 2:1 (Figure A5), 3:1 (Figure A6), and 4:1 (Figure A7) compression conditions. Some asymmetries in errors are notable. For instance, /ð/ is incorrectly identified as /v/ more often than /v/ is incorrectly identified as /ð/.
Figure A1. Consonant confusions for normal hearing listeners, 1:1 compression ratio.
Figure A2. Consonant confusions for normal hearing listeners, 3:1 compression ratio
Figure A3. Consonant confusions for normal hearing listeners, 4:1 compression ratio
Figure A4. Consonant confusions for hearing impaired listeners, 1:1 compression ratio.
Figure A5. Consonant confusions for hearing impaired listeners, 2:1 compression ratio.
Figure A6. Consonant confusions for hearing impaired listeners, 3:1 compression ratio.
Figure A7. Consonant confusions for hearing impaired listeners, 4:1 compression ratio.
REFERENCES


