Spring 2011

The contribution of a frequency-compression hearing aid to contralateral cochlear implant performance

Ann Elizabeth Perreau
University of Iowa

Follow this and additional works at: https://ir.uiowa.edu/etd
Part of the Speech Pathology and Audiology Commons

THE CONTRIBUTION OF A FREQUENCY-COMPRESSION HEARING AID
TO CONTRALATERAL COCHLEAR IMPLANT PERFORMANCE

by

Ann Elizabeth Perreau

An Abstract

Of 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 2011

Thesis Supervisors: Professor Ruth Bentler
Professor Richard Tyler

ABSTRACT

Frequency-lowering signal processing in hearing aids has re-emerged as an option to improve audibility of the high frequencies by expanding the input bandwidth. However, few studies have investigated the usefulness of the scheme as a bimodal option for cochlear implant users. In this study, that question was posed. It was hypothesized that, following fitting and a period of adjustment to a frequency-compression hearing aid, sound localization and speech perception would be improved compared to conventional amplification. More specifically, more high-frequency cues would be perceived in the hearing aid ear using frequency compression, thereby providing better sensitivity to interaural level differences when a cochlear implant is used contralaterally.

There were two experiments in this study. In the first experiment, the goal was to determine if frequency compression was a better bimodal option than conventional amplification. Performance was assessed on tests of sound localization, speech perception in a background of noise, and using questionnaires. Ten subjects with a cochlear implant plus hearing aid participated in experiment one. In the second experiment, the goal was to determine the impact of frequency compression on speech perception in quiet. Consonant and vowel perception in quiet was assessed using the frequency-compression and conventional hearing aid. Seventeen adult subjects participated in the second experiment.

In both experiments, subjects alternated daily between a frequency-compression and conventional hearing aid for two months. The parameters of frequency compression were set individually for each subject and audibility was measured for the frequency compression and conventional hearing aid programs by comparing estimations of the Speech Intelligibility Index (SII) using a modified algorithm (Bentler, R., Cole, B., Wu, Y-H. (2011, March). *Deriving an audibility index for frequency-lowered hearing aids.* Poster session presented at the meeting of the American Auditory Society, Scottsdale, AZ). In both experiments, the outcome measures were administered following the hearing aid fitting to establish baseline performance and after two months of use.

Results revealed no significant difference between the frequency-compression and conventional hearing aid on tests of localization and consonant recognition. Spondee-in-noise and vowel perception scores were significantly higher with the conventional hearing aid compared to the frequency-compression hearing aid after two months of use. These results suggest that, for the subjects in this study, frequency compression is not a better bimodal option than conventional amplification. In addition, speech perception may be negatively influenced by frequency compression because formant frequencies are too severely compressed and can no longer be distinguished.

Abstract Approved: ______________________________

Thesis Supervisor

____

Title and Department

____

Date

____

Thesis Supervisor

____

Title and Department

____

Date

THE CONTRIBUTION OF A FREQUENCY-COMPRESSION HEARING AID TO CONTRALATERAL COCHLEAR IMPLANT PERFORMANCE

by

Ann Elizabeth Perreau

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 2011

Thesis Supervisors: Professor Ruth Bentler
Professor Richard Tyler

Graduate College
The University of Iowa
Iowa City, Iowa

CERTIFICATE OF APPROVAL

PH.D. THESIS

This is to certify that the Ph.D. thesis of

Ann Elizabeth Perreau

has been approved by the Examining Committee
for the thesis requirement for the Doctor of Philosophy
degree in Speech and Hearing Science at the May 2011 graduation.

Thesis Committee: ___________________________________
Ruth A. Bentler, Thesis Supervisor

____
Richard S. Tyler, Thesis Supervisor

____
Paul J. Abbas

____
Carolyn J. Brown

____
Jacob J. Oleson

____
Christopher W. Turner

To my parents for coming to Iowa and staying long enough for me to love it
To my sister and brother-in-law for showing that a terminal degree is possible
To my friends and family for their support and good humor when I needed it
To my husband for his love, dedication, and wonderful presence in my life

ACKNOWLEDGEMENTS

I would like to thank all the people that made this dissertation possible and assisted in its completion. First of all, thank you to my advisors, Drs. Ruth Bentler and Richard Tyler, for their teaching, words of wisdom, and patience along the way. Ruth, you became the driving force when I needed it most. And Rich, I am a more independent and critical thinker now because of your approach. I know that I am well prepared for the opportunities that lay ahead because of the knowledge you have both imparted.

I also want to thank my committee members for their contributions to this project. Throughout this process, you have provided me with constructive criticism and excellent feedback that has helped shape this project and led to its success. I thank all of you for challenging my thinking skills just when I thought I had it all figured out.

Thank you to my colleagues at the University of Iowa, including students, faculty, and staff for their continued support and assistance as well as to the many patients that participated in this study from the University of Iowa Cochlear Implant Program. For financial contributions to this project, I wish to thank the Graduate College at the University of Iowa as well as the Department of Communication Sciences and Disorders. I especially want to thank Dr. Gantz for his support of this project during its completion.

I also want to extend thanks to Dr. Joshua Alexander from Purdue University. Your guidance and support of this project in its infancy stages was beyond helpful. And at the end stages of the project, thanks to Drs. Ruth Litovsky and Yi Zheng from University of Wisconsin-Madison for their input and suggestions and to Andrew Keiner for assisting with some very important, last minute calculations. For providing the devices for this study, thank you as well to Phonak AG, especially Myriel Nyffeler for her assistance.

To my friends and family in Iowa, Chicago, Nebraska, Utah, and all over, I am so grateful for your love and encouragement. I am so blessed to know each of you.

Mom and dad, thanks for helping me to strive for this final goal. Without your wish for higher education, I am not sure I would have persisted this long.

Finally, to my husband, Kirk. Thank you for putting up with all my crazed behavior these last few months (years?). My crabbiness at times, my up-and-down emotions, and my obsessions that the house was too dirty. You have been so patient. I could not have finished this project without you.

TABLE OF CONTENTS

LIST OF TABLES

Table

LIST OF FIGURES

Figure

CHAPTER 1

INTRODUCTION

Cochlear implants (CI) have become an effective means of restoring hearing and improving speech understanding for persons with severe-to-profound hearing loss. The benefits of cochlear implantation have been well documented and include better speech perception in quiet and noise (e.g., Parkinson et al., 2002; Tyler et al., 1995, 2002; Waltzman et al., 1999) and an improved quality of life (e.g., Maillet et al., 1995; Summerfield et al., 2006; Vermeire et al., 2005). Research also suggests that bilateral device use through two cochlear implants (CI+CI) or a cochlear implant plus hearing aid in opposite ears (CI+HA) provides important auditory cues to both ears, thus allowing for potential binaural hearing advantages (Armstrong et al., 1997; Ching et al., 2001, 2004, 2005; Dorman et al., 2004; Dunn et al., 2005; Litovsky et al., 2006; Luntz et al., 2005; Laszig et al., 2004; Morera et al., 2005; Shallop et al., 1992; Tyler et al., 2002, 2006, 2007; van Hoesel & Tyler, 2003; Waltzman et al. 1992). As shown by Zurek (1993), the advantages of hearing with two ears include binaural summation (or combining the intensity or loudness a signal when presented bilaterally versus monaurally), binaural squelch (or combining inputs from the two ears to improve the overall signal-to-noise ratio), and localization (using differences in time, intensity, and spectrum across ears to compare the location of a sound source in space). However, the extent that a CI recipient can benefit from these advantages depends on how well the CI preserves important auditory cues.

The advantages of listening with two ears depend upon the importance of timing, level, and spectral differences across ears for determining the location of a sound source as well as for improved speech perception abilities through binaural summation and squelch effects. A discussion of interaural timing and level cues will be presented first.

1.1 Interaural Time Differences

Data show that when a signal is presented to one ear at 90° azimuth relative to a listener's head, there is a time delay of 650μs for the sound to reach the other ear (Feddersen et al., 1957; Shaw, 1974). This is referred to as the interaural time difference (ITD). In contrast, there is no ITD when a signal is presented at 0° azimuth because the signal arrives at both ears simultaneously. ITDs are useful for determining the location of a sound source, assuming that the wavelength of the signal is larger than the distance the signal must travel from the near to the far ear (Gelfand; Middlebrooks & Green, 1991). Researchers have shown that listeners with normal hearing are sensitive to ITDs of 10 to 15μs (Firszt et al., 2008). However, studies have reported relatively poor sensitivity to ITDs for CI+CI users (Lawson et al., 1998; Long et al., 2003; Seeber & Fastl, 2008; Senn et al., 2005; van Hoesel et al., 2002; van Hoesel & Tyler, 2003) and CI+HA users (Francart et al., 2008a).

ITDs have been classified into three types: 1) envelope ITDs, or low rate temporal fluctuations of the stimulus, 2) periodicity or temporal information relating to voice fundamental frequency, and 3) ongoing fine-structure ITDs, or those in the cycle-by-cycle structure of the stimulus (Rosen, 1992). Research has shown that listeners with normal hearing and hearing impairment are sensitive to all types of ITDs, whereas CI users are sensitive to envelope and periodicity cues only (Francart et al., 2008a; Grantham et al., 2007; Rosen, 1992). Furthermore, among CI+CI users, ITD sensitivity has been shown to vary considerably from 50μs (Lawson et al., 1998) to 1000μs (Grantham et al., 2008; van Hoesel & Tyler, 2003).

There are several reasons for poor sensitivity to fine-structure ITDs for CI users. For example, the signal processing strategy of the CI may not adequately convey fine-structure information (Moore, 2003). This is because most strategies use band-pass filters to extract the envelope of the signal from multiple bands along the electrode array and, in so doing, lose all fine-structure information of the original stimulus (van Hoesel & Tyler, 2003). In addition, processing time differences between devices or the use of directional microphones on one or both devices may also be factors. For CI+HA users, digital processing delays (or 'group' delay; e.g., Dillon et al., 2003) and different compression and microphone characteristics (e.g., Ching et al., 2007; Musa-Shufani et al., 2006) can result in time delays and have the potential to negatively affect speech understanding and localization.

1.2 Interaural Level Differences

Differences in loudness or the intensity of a signal can also be used to determine the location of a sound source, provided that the wavelength of the signal is shorter than the dimensions of the head (Gelfand, 2004; Middlebrooks & Green, 1991). When a signal is presented to one side of a listener, a difference in level emerges as the head acts as a barrier of sound, creating a head shadow effect by attenuating sound to the ear farthest from the signal, particularly in the higher frequency range. Consistent with the 'duplex theory' introduced by Rayleigh (1907), researchers have found that ITDs are used for discriminating low-frequency sounds below 1500Hz and ILDs are used for high-frequency sounds above 1500Hz. Research has also shown that listeners with normal hearing have ILDs ranging from 0.5 to 2.5 dB, depending on the intensity and frequency of the incoming signal (Yost & Dye, 1988).

Data from several studies have indicated relatively good sensitivity to ILD cues for CI+CI users (Laback et al., 2004; Lawson et al., 1998; Senn et al., 2005; van Hoesel & Tyler, 2003) and CI+HA users (Francart et al., 2008b). For instance, van Hoesel & Tyler (2003) demonstrated good sensitivity to binaural level cues up to 2.5 dB for four of five subjects even though sensitivity to timing cues was very weak (i.e., up to 100μs). Moreover, two subjects in that study showed ILDs of 0.17 dB, similar or better to that of listeners with normal hearing. Because CI users are more sensitive to ILDs than ITDs, ILDs have been considered the primary cue for sound source localization (Firszt et al., 2008).

Despite the potential limitations of CIs in preserving important auditory cues, studies have shown that fitting a hearing aid to the opposite ear of the implant can contribute to binaural hearing. Compared to use of a CI alone, CI+HA use has been shown to provide better speech understanding in quiet and in noise (Armstrong et al., 1997; Ching et al., 2004, 2005; Dunn et al., 2005; Flynn & Schmidtke, 2004; Kileny et. al, 2004; Luntz et al., 2005; Mok et al., 2006; Tyler et al., 2002), improved sound source localization (Ching et al., 2004; Dunn et al., 2005; Potts et al., 2009; Seeber et al., 2004; Tyler et al., 2002), and improved music recognition and pitch perception (Gfeller et al., 2006, 2007; Kong et al., 2005).

Furthermore, evidence from a meta-analysis of 14 studies suggests that CI+HA use provides binaural advantages for speech perception, resulting from the combined effects of binaural summation, binaural squelch, and head shadow (Ching et al., 2007). The binaural summation (or redundancy) effect is described as the improvement in speech perception when redundant information is processed simultaneously by the two ears. This effect can be measured for speech presented in quiet or in noise with the target(s) originating from the front of the listener. Binaural squelch effect refers to the improvement in speech perception when speech is presented from a location that is spatially separate from the noise and the listener can selectively attend to the ear with the better signal-to-noise ratio as compared to the sound at the ear with the poorer signal-to-noise ratio. Lastly, head shadow or head diffraction is a physical phenomenon that arises from ILDs cues, enabling the listener to attend to the ear with the better signal-to-noise ratio. In the report by Ching et al. (2007), binaural benefit was measured by comparing performance for the bilateral mode (CI+HA) to the unilateral modes (CI only and HA only) on speech-in-noise tasks using spatially separate target and noise sources. Results of the meta-analysis revealed a binaural advantage of 1-3 dB for speech perception

(Ching et al., 2007). Although this is smaller than the advantage for listeners with normal hearing (i.e., 5-7 dB from Zurek, 1993), these data indicate that CI+HA use can provide important binaural hearing advantages.

The use of a CI+HA in opposite ears requires bilateral integration of two different processing modes via electric and acoustic stimulation. More specifically, the CI provides electrical stimulation in one ear whereas the hearing aid provides acoustic stimulation to the other ear. Because the input to the two ears is inherently different, abnormal differences in timing and level cues may result. In fact, it has been reported that CI+HA users can hear separate and different sounds on each side. Tyler et al. (2002) described one individual where the implant side was louder and often times heard as an echo, “just a split second behind the hearing aid side”.

On the other hand, despite the asymmetry across ears, Potts et al. (2009) suggested that because listeners with normal hearing and hearing impairment are able to take advantage of atypical time and level cues overtime, it is likely that CI+HA users would also benefit from both devices. It has been shown that low-frequency residual hearing complements high-frequency electrical stimulation through a CI, thereby providing access to a wider frequency range than either the hearing aid or CI alone (Blamey & Saunders, 2008). Additionally, Kong et al. (2005) reported that the addition of a hearing aid to contralateral CI use significantly improved performance for four individuals when parsing out a male talker from a competing female talker. When the subjects utilized the CI alone, a female and male talker equally masked the target male talker, suggesting that segregation of the voices was made possible by the low-frequency audibility afforded by the hearing aid (Kong et al., 2005). Therefore, combining acoustic plus electric stimulation in opposite ears has the potential to provide important binaural hearing advantages for many listeners, but differences in signal processing and timing and level cues must be overcome to make this possible.

To conceptualize the spectral differences that occur between the two ears when a sound source is presented bilaterally, it is necessary to examine head-related transfer functions (HRTFs). The HRTF refers to the transformation of sound pressure level (in dB) from the free field to the eardrum as a function of frequency (in Hz) and azimuth (θ in °). The output or gain from the HRTF is due, in part, to the natural resonances of the concha and ear canal and the contours of the pinna. At +90° azimuth, there is an increase in sound energy of approximately 17 to 20 dB for frequencies from 2000 to 6000 Hz. For low-frequency sounds below 1500 Hz, there is a negligible effect which confirms previous findings that ILDs are not useful for localizing low-frequency sounds. At -90° (or +270°) azimuth, the same high frequencies of 2000 to 3000 Hz provide important auditory cues for localization and lower frequencies do not contribute to ILDs; however, the amount of gain measured at 2000 to 3000 Hz is about 8 dB, about 10 dB less than that for a +90° sound source. In sum, the HTRFs indicate that there are useable ILD cues from 2000 to 6000 Hz and that these frequencies should be coded by both the CI+HA for adequate sound source localization. As stated previously, ITDs would be used to localize low-frequency sounds below 1500 Hz. However, it should be noted that listeners might still be able to detect ITDs at high frequencies using envelope cues. Middlebrooks & Green (1991) reported that localization is possible using envelope cues for high-pass transients, band-pass noise, two-tone complexes, and sinuoidally amplitude modulated tones (for carriers up to 4000 Hz and modulated tones up to 500 Hz). Because CIs preserve envelope cues to a certain extent, it is possible that high-frequency ITDs would be processed via envelope cues.

Previous studies of CI+HA users have demonstrated improved localization when using both devices compared to use of either the CI or hearing aid alone (Ching et al., 2004; Dunn et al., 2005; Potts et al., 2009; Seeber et al., 2004; Tyler et al., 2002). Additionally, studies have examined ways to enhance the contribution of the contralateral hearing aid to localization performance. Ching et al. (2004) studied the influence of different hearing aid fitting methods on localization for 21 CI+HA users. They adjusted the hearing aid using a) a paired-comparisons task to identify the preferred frequency response for speech understanding and b) a loudness-balancing task to equate loudness of speech between the CI+HA. Although they found significantly better localization for 12 of 18 users with the CI+HA over the CI or hearing aid alone, localization abilities were not consistently improved for all subjects and many localization errors remained (Ching et al., 2004). In fact, research has shown that loudness balancing between a hearing aid and CI is a complex task (Dunn et al., 2005), as loudness differences may vary as a function of frequency *and* level between the two devices (Armstrong et al., 1997). Furthermore, sensitivity to ILDs necessitates equating loudness between the ears (Ching et al., 2007) and if a loudness imbalance occurs, localization of sound and speech perception with spatially separated noise sources would be compromised. It is possible that timing and level differences between the CI+HA explains why binaural advantages have not been observed across all studies or across all subjects with CI+HA (see Ching et al., 2007; Firszt et al, 2008).

Frequency-lowering signal processing in hearing aids has been touted as a unique way to improve audibility for high frequencies. This presents an opportunity to advance the fitting practices of CIs and hearing aids. By making high-frequency cues audible in the hearing aid through frequency lowering, it is possible that binaural sensitivity to ILD cues might also improve, provided that the CI is able to code level cues adequately. As a result, sound localization using a CI+HA may be improved if level differences between the CI and hearing aid are better resolved. However, frequency lowering also has the potential to distort the acoustic signal in ways that negatively impact both speech perception and localization abilities because of the shift in spectral components of the input signal. As a general rule, these frequency-lowering schemes alter the bandwidth of high-frequency input signals that would normally be stimulated in the basal region of the cochlea to the mid-to-low frequencies stimulated more apically in the cochlea. This is achieved by either compression or transposition of that high-frequency energy (and will be discussed in more detail in chapter 2).

Figure 1.1 (adapted from Simpson et al., 2005) illustrates how signals are processed in a frequency-compression hearing aid similar to the one used in this study. The input signal from the microphone is separated into a low- and a high-pass band as determined by the cutoff frequency programmed by the fitting software. The low-pass band is created using a finite impulse response (FIR) filter with a slope of -35dB/octave. Inputs in the low-pass band undergo a delay of 6.2 ms and no additional processing. The high-pass band is created by processing of overlapping data blocks, each consisting of 256 samples that are converted into 128-point sequences and processed through a fast Fourier transform (FFT). Magnitude and phase information are extracted from the FFT, which is executed after every 32 input samples or at intervals of 2.2 ms. As Figure 1.1 shows, the first 24 FFT bins above the cutoff frequency are selected and assigned to 24 oscillators. The outputs of the oscillators are summed and this composite signal is summed with the low-pass signal to generate the output signal at the receiver.

The amplitude from the FFT is modified to provide the necessary gain in each frequency range as determined by the hearing loss and amplification strategy used. The relationship between output and input frequency can be expressed by the following:

$$F_{out} = F_{in},\ F_{LPF}^{\ 1-p} * F^{\,p}_{in},$$

where F_{in}= input frequency, F_{out} = output frequency, F_{LPF} = cutoff frequency, p = compression exponent. Within the bandwidth to be compressed, the compression exponent is ≠ the ratio of the input/output. Instead, this compression ratio is = input bandwidth (above the cutoff frequency, in ERB)/output bandwidth (above the cutoff frequency, in ERB). The ERB is a log-based frequency scale related to the equivalent rectangular bandwidth of an auditory filter for normal hearing listeners (Glasberg & Moore, 1990).

A potential advantage of frequency compression over conventional amplification for persons with moderate-to-severe hearing loss is that the input bandwidth of the hearing aid can be extended to process more high-frequency sounds. For example, the useable input bandwidth for a given individual could range from 200 Hz, where the hearing aid begins to amplify, to 3200 Hz, where the hearing aid can no longer provide useful amplification due to the severity of loss. Using a frequency-compression hearing aid and a cutoff frequency of 2100 Hz, the input bandwidth would be extended to approximately 6600 Hz (the input bandwidth of compression = 4500 Hz). For this individual, the input bandwidth was effectively extended from 3200 Hz to 6600 Hz using frequency compression, giving the user the potential to improve perception of high-frequency cues necessary for speech perception or sound localization.

The primary goal of this study is to investigate the contribution of a frequency-compression hearing aid to contralateral CI performance. The first research question in experiment one will investigate sound localization using the CI+HA, comparing the frequency-compression hearing aid to a conventional hearing aid. It is hypothesized that, following fitting and a period of adjustment to a frequency-compression hearing aid, sound localization will be improved using frequency compression compared to conventional amplification because more high-frequency cues would be perceived in the hearing aid ear, thereby providing better sensitivity to ILDs when the CI is used contralaterally.

The second research question in experiment one will investigate the contribution of frequency compression for speech perception in a background of noise, using spatially-separate target and noise sources. Here, the impact of frequency compression on speech perception in background noise will be evaluated using both devices (CI+HA) and each device alone (CI only and hearing aid only) compared to conventional amplification. Because performance for speech tasks using spatially-separate target and noise sources depends upon the ability to process important auditory cues from both ears, it is hypothesized that frequency compression will provide improved performance over conventional amplification when a CI is used contralaterally, due to improved sensitivity to high-frequency cues (which may improve speech recognition in general and also spectral location cues).

A secondary goal of the study is to determine the impact of frequency-compression on speech perception in quiet. In experiment two, perception of consonants and vowels will be evaluated for the frequency-compression and conventional hearing aids. Because this particular frequency compression circuit does not affect inputs below 1500 Hz, it would be expected that low-frequency stimuli such as vowels would not be affected by frequency compression. However, frequency compression may improve the audibility of high frequencies and the perception of consonants because these high-frequency signals are subject to compression above the cutoff frequency. Therefore, it is hypothesized that although vowel perception will be unaltered by use of frequency compression, consonant perception will be enhanced because the frequency compression will provide access to those higher frequency cues.

In this study, audibility will be measured and quantified by comparing estimations of the Speech Intelligibility Index (SII) (ANSI S3.5-1997) obtained with the frequency-compression hearing aid to that of the conventional hearing aid (see Figure 1.2 for an example). It is anticipated that the frequency-compression hearing aid would provide better audibility because sound is processed in the mid-frequency range and not limited by the severity of the hearing loss in the high frequencies, as it could with conventional amplification. As a result, the SII for frequency compression may be increased compared to conventional amplification.

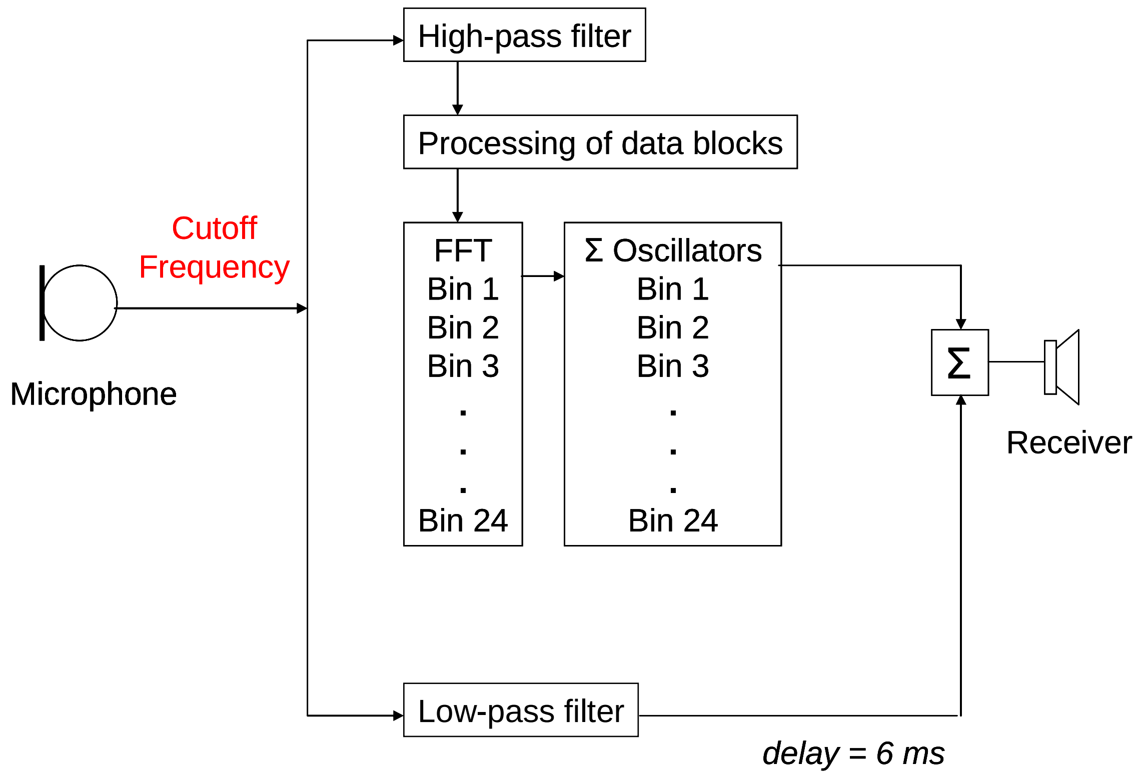


Figure 1.1. A block diagram of the signal processing of a frequency-compression hearing aid (adapted from Simpson et al., 2005).

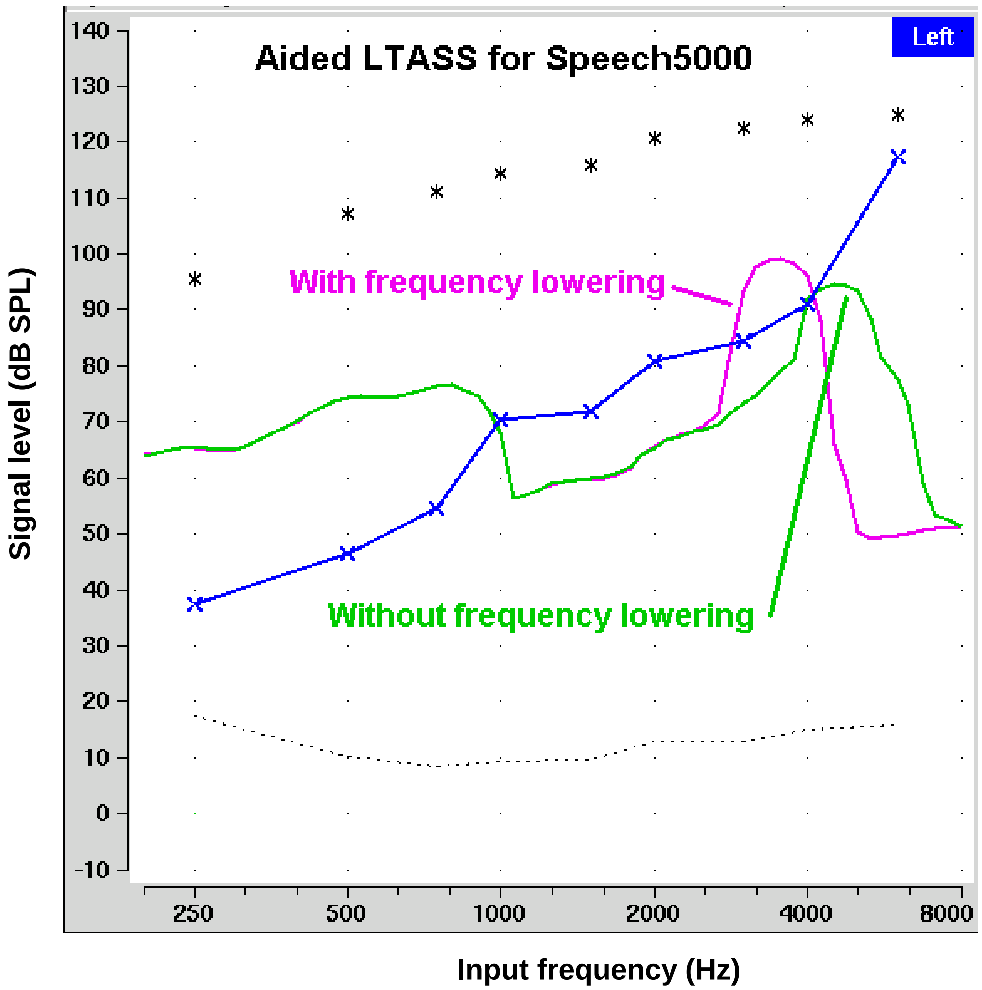


Figure 1.2. Aided long-term average spectrum of speech (LTASS) for a 5000 Hz one-third octave band. The x-axis displays the input frequency (in Hz) and the y-axis displays signal level (in dB SPL). Unaided hearing thresholds are plotted by the x's and the audibility of the isolated 5000 Hz one-third octave band of speech is shown with the frequency-lowering algorithm activated and not activated. The asterisks represent the estimated uncomfortable loudness levels (UCLs) and the dashed lower line represents normal hearing threshold.

CHAPTER 2

LITERATURE REVIEW

In the last chapter, previous studies were summarized on CI+HA use and a brief explanation of how frequency-compression hearing aids could potentially provide better audibility of high-frequency sounds, and therefore improve binaural cues with use of a CI contralaterally, was provided. The purposes of the study were reviewed and the hypotheses to be tested were outlined. In the current section, I will expand on this by reviewing the history of frequency-lowering technology and also describe the different types of frequency-lowering systems used in today's hearing aids. Results from recent investigations with frequency-lowering devices will be provided.

2.1 History of Frequency-lowering Technology

2.1.1 Review of Early Research

Since the 1950s, frequency-lowering technology has been explored as a way to improve hearing and speech perception for listeners with high-frequency hearing loss. Early on, researchers created a variety of frequency-lowering devices using frequency transposers and vocoders and studied their effect on intelligibility for listeners both with normal hearing and hearing loss. Braida et al. (1979) categorized these and other early efforts into six sections according to the type of frequency-lowering: a) transposition, which transposes high-frequency components to lower frequencies by adding the transposed signal to unprocessed low-frequency sounds, b) channel vocoders, where the input signal is divided into various frequency bands and the output envelopes of these bands are detected and applied to low-frequency carriers, c) zero crossing-rate division, where speech is filtered into four bands, each equal to the spacing between formant frequencies, and processed to generate a band reduction by a certain factor, d) slow-playback, where sound is recorded and replayed at a rate slower than the original signal, resulting in an output signal longer in time and lower in frequency than the input, e) time-compressed slow-playback, which is similar to slow-playback, but certain temporal components are deleted periodically to maintain the temporal characteristics of speech, and f) frequency shifting, where all spectral components in a given band are shifted by a fixed factor. A discussion of these devices as they originated and their impact on speech perception follows.

Johansson (1961) developed the first frequency transposition device using a body-worn hearing aid, called the Oticon TP-72, where frequencies between 3000 to 6000 Hz were processed via a nonlinear modulator and the input signals were converted into broad spectrum signals. The processed signals were amplified using a compression amplifier and mixed with unprocessed signals whose frequencies were below 1500 Hz. After training with the frequency transposition device, results revealed improvements in word recognition scores (Wedenberg, 1959; Johansson, 1966) and syllable and word identification (Johansson & Sjogren, 1965) compared to conventional amplification for some, but not all subjects. However, other studies have reported less favorable results including better detection of the phoneme /s/ using frequency transposition, and in some cases, poorer word recognition than conventional amplification (Foust & Gengel, 1973).

A study comparing the Johansson frequency transposition hearing aid device to linear amplification found no benefit in word recognition scores following training for a group of eight children (Ling, 1968). Performance using a six-channel vocoder was also evaluated, where sounds from 2000 to 3000 Hz were filtered and the output envelopes modulated amplitudes of five sine waves from 750 to 1000 Hz. These signals were linearly amplified with sounds in the 70 to 700 Hz range and all other sounds were discarded via filtering. An early study demonstrated improvement in speech understanding with the vocoder system after training, albeit at the same rate as with conventional amplification (Ling & Druz, 1967). In this study, however, Ling (1968) found no significant differences in word recognition scores between the Johansson transposition hearing aids and the vocoder system.

Velmans (1971, 1974) created a similar frequency transposition device to the Johansson model that shifted, rather than compressed, high frequencies to a lower frequency region. In that device, the incoming signals were separated into a low- and a high-pass band using a cutoff frequency of 4000 Hz. Signals in the high-pass band were lowered by 4000 Hz and mixed with the unprocessed signals between 0 and 4000 Hz. Velmans (1973, 1975) reported significantly improved detection of high-frequency speech sounds, especially for cues of manner and place of articulation. However, Velmans (1975) also found that word recognition abilities did not improve with the transposition device following training for six children with sensorineural hearing loss.

Other early developments in frequency-lowering systems were made by Guttman & Nelson (1968) who developed a zero-crossing-rate division system that lowered voiceless sounds from 6000 to 15,000 Hz using zero-crossing synchronous pulses and low-pass filtering. These processed signals were added to the unprocessed signals or voiced components of speech from 100 to 2000 Hz. They found improvements with short duration stimuli such as single words, but no improvement for longer duration stimuli such as sentences. Later on, Guttman et al. (1970) evaluated the benefits of the zero-crossing-rate division device for six children with severe hearing impairment. They compared changes in articulation and /s/ vs. /sh/ discrimination using their frequency-lowering device to a control group using conventional amplification. Following multiple training sessions, they found improvements in articulation of /s/ and /sh/ for children using the frequency-lowering device. For the control group, they noted similar improvements in articulation, thus suggesting that training proved more useful than the utility of frequency lowering with the zero-crossing-rate division system.

Finally, several early studies employed slow-playback speech, where speech is recorded and replayed at a slower rate, in effect, lowering the speech spectrum by a multiplicative factor equal to the slowdown factor (Braida et al., 1979). A potential downfall of this type of slow-playback frequency-lowering is that speech is distorted both spectrally and temporally and therefore, is not presented in ‘real-time’ to the listener. Despite this limitation, Bennett & Byers (1967) found improved word recognition scores for 15 individuals with high-frequency hearing loss when the playback speed was equal to 80% of the original recording speed. However, further reductions in the bandwidth degraded speech intelligibility significantly. Other research efforts on slow-playback frequency-lowering systems have tried to maintain the temporal aspects of speech by deleting segments of speech or compressing in the time domain. Beasley et al. (1976) compared time-compressed slow-playback speech to linear amplification for 18 children with hearing loss. The children were separated into two groups and training was provided for either time-compressed slow-playback speech or linear hearing aids. Results revealed that the children trained using the frequency-lowered speech obtained significantly more increases in word scores versus the children without frequency lowering. However, word recognition scores at the conclusion of the study were no different between time-compressed slow-playback and linear hearing aids.

Across all studies, more difficulties in speech intelligibility are reported by listeners when greater reductions occur to the bandwidth or when spectral components are more severely lowered (Bennett & Byers, 1967; Redden, 1973; Takefuta & Swigart, 1968). Turner and Hurtig (1999) reported on a frequency-shifting scheme to proportionally lower all frequencies by a fixed multiplicative factor; they found that the most severe frequency-shifting conditions resulted in poorer performance. In their study, speech perception was measured using nonsense syllables in quiet for both male and female talkers for 15 subjects (16 ears) with hearing impairment and 3 subjects with normal hearing. For the subjects with hearing impairment, high-frequency sounds were amplified to provide better audibility. Multiple frequency-shifting ratios (0.5, 0.6, 0.7, 0.8, 0.9, and 1.0) were studied and all subjects were trained in each condition to achieve a plateau or stable performance before testing was initiated. For listeners with normal hearing, the frequency-shifting conditions of 0.5 and 0.6 had the greatest affect on speech perception whereas conditions of less frequency-shifting (0.7-1.0) did not significantly affect scores. For the listeners with hearing impairment, significant improvements in speech recognition scores were obtained using the 0.8 condition for 7 out of 16 subjects with the female talker and 3 out of 15 subjects with the male talker. Only one subject demonstrated a significant decrement using frequency-shifting for the female talker and two subjects for the male talker.

Finally, there is evidence from the frequency-shifting literature that listeners with severe-to-profound deafness successfully ignore aliasing effects, where high- and low-frequency components are overlaid or ‘aliased’ during signal processing (Biondi & Biondi, 1968). In that study, frequency-shifting was accomplished using the aliasing technique and deemed very bothersome for listeners with normal hearing and moderate hearing loss, but not for listeners with severe-to-profound hearing loss. Despite this notion, speech perception abilities using frequency-lowering for listeners with severe-to-profound hearing loss remain only slightly better than conventional amplification after extensive training (Braida et al., 1979).

2.1.2 Limitations of Early Studies

Research of the early frequency-lowering systems suggests that no or only slight improvement (e.g., high-frequency sound detection for /sh/ and /s/) has been found for listeners with hearing impairment compared to conventional amplification (Braida et al., 1979). For example, several studies of time-compressed slow-playback speech reported improvements in word recognition scores overtime with extensive training for adults and children, but the best scores continued to be worse than those with conventional amplification (Oeken, 1963; Haspiel, 1969; Zemlin, 1966). In studies where frequency-lowering was found more effective compared to conventional amplification, the margin of improvement was very small (Braida, et al., 1979). Additionally, studies on frequency-lowering have traditionally been conducted on subjects with profound hearing loss in the high frequencies. It is possible that the persons with more severe hearing losses may not receive as much benefit from frequency lowering as those with better hearing thresholds (Simpson, 2009). Secondly, many of the earliest frequency-lowering algorithms altered the speech signal in ways that negatively affected important characteristics of speech, such as temporal and rhythmic patterns, fundamental frequency information, and vowel and syllable durations, and did not preserve the necessary contrasts for distinguishing acoustic cues, such as unvoiced and voiced sounds (Braida et al., 1979).

2.2 Recent Studies

In the past decade, frequency-lowering signal processing has re-emerged as a strategy to improve salient aspects of the speech signal. There are several digital hearing aids that now offer a frequency-lowering signal processing option; however, the methods of frequency-lowering vary considerably across the devices. We will discuss in detail three types of frequency-lowering schemes that have been studied extensively and are currently available in hearing aids. These include, slow-playback, frequency transposition, and frequency compression.

2.2.1 Slow-playback

2.2.1.1 How It Works

Slow-playback is described as sound that is recorded and replayed at a rate slower than the original input signal, resulting in an output signal that is longer in time and lower in frequency than the input signal. All inputs are proportionally lowered or shifted, which preserves the spectral relationships among the different frequency components. The disadvantage of slow-playback is that input signals are lengthened in time, thus causing distortion in the time domain.

AVR Sonovations has four different hearing aids with slow-playback frequency lowering: the ImpaCT XP behind-the-ear (BTE), the Nano XP BTE, the Logicom XP BTE, and previously, the TranSonic Model FT-40 MKII body aid. In all AVR

Sonovations hearing aids, frequency lowering occurs as the signal is sampled periodically, more than 56,000 Hz, and then transferred to an analog memory device where the samples are stored (McDermott et al., 1999). The samples are extracted from the device at a constant rate of 56,000 Hz and an output signal is constructed that is then amplified and delivered to the earpiece. Next, the input sampling rate is increased relative to the constant output signal by a predetermined factor = Z_C. This has the effect of lowering all frequencies by the same proportion. Because two different sampling rates are employed at the input and output of the device, a second memory device is used so that operation is alternated between the two devices automatically. To continue operation in 'real-time', incoming samples have to be deleted or discarded because the samples are stored in one device before delivered to the earpiece by the other device. Also, a delay of 9 ms is applied as signals are processed by the two devices. Input signals are frequency-lowered based on the dominant spectral contents of the incoming waveform, and specifically, based on the amount of high-frequency energy above 2500 Hz. If the dominant input is above 2500 Hz, then all frequencies above this cutoff (Z_C) are shifted proportionally. However, if the dominant input signal is lower than 2500 Hz, then the signal is amplified without frequency lowering. When frequency lowering is activated, an adjustable factor called dynamic consonant boost (DCB) can be applied that will increase gain for the frequencies being lowered. The hearing aid models differ slightly with respect to how the DCB occurs or how additional gain is applied to the frequency-lowered signal.

2.2.1.2 Studies

Several studies have been conducted with adults and children using slow-playback technology via the AVR Sonovations hearing aids. For children, improved detection of environmental and Ling sounds (Rosenhouse, 1990) and better aided thresholds (MacArdle et al., 2001; Miller-Hansen et al., 2003) have been reported with the AVR slow-playback devices over conventional hearing aids. Small, yet measureable improvements in word recognition have also been documented for children using the AVR slow-playback hearing aids. Miller-Hansen et al. (2003) demonstrated a 10% improvement in word scores for 8 of 16 children following one-month of use in a home trial with the ImpaCT BTE hearing aid compared to conventional hearing aids. Further analyses showed that the children with the poorest performance using conventional amplification had the best performance during the study with the frequency-lowering hearing aids.

For adults, Parent et al. (1998) fitted the TranSonic body aids bilaterally on four experienced hearing aid users and evaluated performance over four test sessions at two-week intervals. All subjects had severe-to-profound hearing loss. Similar to results from children (Miller-Hansen et al., 2003), there was an improvement for detection of high-frequency sounds for all subjects using the TranSonic hearing aids. For speech perception, two subjects showed significant improvements on the CID Everyday Sentence Test presented in quiet (from the cochlear implant test battery; Cochlear Corporation, 1995). The other two subjects showed no improvement with frequency lowering. One subject reportedly heard soft speech sounds better with the frequency-lowering hearing aids, but preferred his own conventional hearing aids due to the harsh quality of the frequency-lowered high-frequency sounds (i.e., /s/ and /sh/) even after the four two-week intervals of adjustment. Finally, it was reported by Parent et al. (1998) that those participants with significantly improved speech perception using the frequency-lowering hearing aids also preferred these devices over conventional amplification.

McDermott et al. (1999) reported improvements in sound detection and speech recognition for five adult subjects using the TranSonic slow-playback frequency-lowering hearing aid, but largely due to improvements in audibility compared to their own hearing aids. Performance was measured on tests of aided audiometrics and speech recognition using the frequency-lowering hearing aid versus their own hearing aid. The amount of frequency lowering, or Z_c value, was progressively increased over 12 weeks of use with the TranSonic hearing aid. Results showed significantly better aided thresholds with the TranSonic aids compared to the subjects' own aid for four of five subjects. Additionally, for four of five subjects, speech perception scores using the TranSonic aid were higher than the mean scores with their own aid using at least one of the Z_c or frequency-lowering settings. However, results suggested that better audibility in the lower frequencies using the TranSonic hearing aid may have contributed to the better performance observed on the speech perception tests compared to their own aids. Moreover, there were only two subjects that, in addition to improved low-frequency audibility, also demonstrated improved speech perception abilities using the TranSonic frequency-lowering hearing aid.

In a similar design, McDermott and Knight (2001) compared results with the ImpaCT frequency-lowering hearing aid to conventional amplification for three subjects following six weeks of use. There was no significant difference in scores for the Consonant/Nucleus/Consonant (CNC; Peterson & Lehiste, 1962) word recognition test and a consonant perception test in quiet compared to use of the subjects' conventional hearing aids. In comparison, for sentences in noise, performance was significantly worse with frequency lowering compared to conventional amplification. Reportedly changes to the high- and low-pass filters during programming of the devices produced an overall reduction in bandwidth, adversely affecting performance on the speech-in-noise task for these subjects.

More recently, Gifford et al. (2007) investigated the use of the AVR Nano frequency-lowering hearing aids for six subjects with steeply sloping losses that fit the criteria for the short electrode CI. Each subject wore the AVR Nano hearing aid for five weeks. Performance was assessed on the CNC word recognition test in quiet (Peterson & Lehiste, 1962), the City University of New York sentence recognition test (CUNY; Boothroyd, Hanin, & Hnath, 1985) in quiet and in noise (+10 dB SNR), and the AZBio sentence recognition test (Spahr & Dorman, 2004) in quiet and in noise (+10 dB SNR). There were no significant differences in mean scores between the subjects' conventional hearing aids and the frequency-lowering hearing aids for any measures tested. Individual results showed improved performance using the frequency-lowering hearing aid for two of six subjects across all test measures. In addition, three of six subjects showed improvements of 2-15% for the CUNY and AZBIO sentences in noise (statistical significance was not provided). Finally, subjective ratings were collected using the Abbreviated Profile of Hearing Aid Benefit (APHAB; Cox & Alexander, 1995; Cox, 1997) to compare the percentage of problems before and after fitting of the frequency-lowering hearing aids. Results on the APHAB revealed no significant difference in subjective performance using the frequency-lowering hearing aid. In sum, although individual differences existed, the results from this study suggested that frequency lowering did not provide better performance compared to conventional amplification.

As reported by Simpson (2009), results of these studies on slow-playback hearing aid technology show that 13 of 28 children and 6 of 18 adults reported significant improvements for speech understanding. The range of improvement was 10-20% for word and sentence recognition, which is not entirely different from the results reported by early studies on slow-playback technology (Braida et al., 1979).

2.2.2 Frequency Transposition

2.2.2.1 How It Works

Generally speaking, frequency transposition shifts high-frequency sounds to a lower frequency region by adding the transposed high-frequency signal to unprocessed low-frequency sounds. In recent years, two frequency transposition systems have been described in the literature, the Widex Audibility Extender (Kuk et al., 2006) and a frequency transposition scheme based on dead regions in the cochlea (Robinson et al., 2007). The Widex Audibility Extender uses two defined frequency regions as the basis for frequency transposition, the source octave and the target octave. The source octave describes the frequency region of the signal to be transposed whereas the target octave includes the region of both the high-frequency, transposed signals and low-frequency, unprocessed signals. Frequencies to be transposed are analyzed and a dominant energy peak is determined. The dominant peak within the source octave is shifted or lowered by one octave and all other frequencies are lowered linearly by the same amount. Also, the transposed signal is simultaneously band-pass filtered by one octave around the dominant peak to limit any masking effects that may occur from the frequency-transposed signal. The transposed signals are mixed with the low-frequency, unprocessed signals and amplification is applied as determined by the fitting software. In the Widex algorithm, a "start frequency" is used to determine the source octave for frequency transposition and frequencies can be transposed up to two octaves above this frequency.

Robinson, Baer, & Moore (2007) developed a similar frequency transposition scheme wherein the start frequency was defined by an individual's dead region, or the region of the cochlea with nonfunctioning inner hair cells (Thornton & Abbas, 1980). Prior to fitting the device, the edge frequency of the dead region (f_e) was determined by the TEN(HL) test (Moore et al., 2004) and psychophysical tuning curves (Kluk & Moore, 2005; Sek et al., 2005). Next, a target octave from f_e to $1.7f_e$ was determined as the destination of the transposed signals. This target octave was decidedly based upon on a study by Vickers et al. (2001), who found that individuals with dead regions can make use of amplification up to one octave above the f_e. For the source octave, the high-frequency sounds fell within the dead region from $2f_e$ to $2.7f_e$. Only frequencies within the source octave of $2f_e$ to $2.7f_e$ are lowered and frequencies below f_e are left unprocessed.

2.2.2.2 Studies

Kuk et al. (2007) studied the effect of frequency transposition via the Widex Audibility Extender on consonant and vowel perception and subjective preference ratings for 13 adults with high-frequency hearing loss. The subjects were fit with hearing aids bilaterally that were programmed with and without frequency-lowering in two memories. Subjects were kept blind to the programs. Twelve of the 13 subjects used a start frequency of 4000 Hz and one subject used a start frequency of 3200 Hz. In this study, information on verification that the transposed signal was audible was not provided. Following two weeks of hearing aid use, results revealed an average improvement in consonant scores of 12% at 30 dB HL and 3% at 50 dB HL, with little improvement or decrement in vowel scores. On average, subjective ratings indicated that the Audibility Extender frequency-lowering program was favored over the conventional program when listening to birds (about 70% of the time) and music (65%) and, to a lesser degree, female voices (30%).

A later study by Kuk et al. (2009) investigated the effect of frequency transposition on consonant perception during a home trial of two months. Eight adult participants with steeply sloping high-frequency hearing loss were recruited and fit bilaterally with hearing aids. The start frequency and gain of the transposed signal were verified using a recorded, interrupted /s/ presented through an audiometer at 30 dB HL. However, details on the fitting of frequency transposition were not further discussed. Performance for consonant recognition in quiet and in noise was assessed initially using conventional amplification and after three subsequent visits using frequency transposition. Results showed that discrimination of fricatives was significantly improved both in quiet and in noise using frequency transposition over the three visits compared to conventional amplification. All other phoneme classes showed no significant difference with or without frequency transposition.

Studies of frequency transposition based on dead regions have shown inconsistent findings. In the first study, Robinson et al. (2007) showed improvements for consonant perception especially for fricatives and affricates, compared to a control condition with no frequency lowering. Seven subjects with high-frequency dead regions were fit with the frequency transposition device and trained for eight hours prior to testing. Verification of the transposed signal was completed by presenting two VCV stimuli (i.e., /asa/ and /asha/) at different levels and adjusting amplification individually for each subject based on these results. Using the frequency transposition device, four of seven subjects demonstrated improved discrimination of affricate consonants and five of seven subjects showed improved identification of /s/ and /z/ in the final word position compared to the control condition.

Despite these improvements with the frequency-lowering scheme in the laboratory, a follow-up study implementing a field trial of five weeks (Robinson et al., 2009) found no significant improvement in speech perception. Here, the frequency-lowering scheme was added to digital hearing aids in one program and a control condition with no frequency lowering was programmed in the other. Five subjects wore the aids for five weeks. Similar to their initial study, hearing aid gain was prescribed for both the transposed and conventional programs using the ‘Cambridge’ formula, which aims to make all speech frequency bands from 0.5 to 5 kHz equally loud (Robinson et al., 2007). In this study, however, no formal verification of the transposed signal was done. Instead, a default level was chosen for each subject and adjusted only when required to make the transposed signal audible. Only one subject showed an improvement in consonant discrimination with the frequency-lowering device and two subjects showed significant decrements. For discrimination of final-word /s/ and /z/, only one subject showed a statistical improvement with frequency lowering over the conventional program. Finally, no subject preferred the frequency transposition program over the control condition following the home trial.

2.2.3 Frequency Compression

2.2.3.1 How It Works

Frequency compression occurs when the input signal bandwidth is compressed into a narrower output bandwidth. Previous studies have reported two types of frequency compression: 1) proportional (or ratio) shifting, whereby all frequencies are shifted by the same multiplicative factor (McDermott & Dean, 2000; Turner & Hurtig, 1999), or 2) nonlinear shifting, wherein certain frequencies above a predetermined cutoff are compressed and signals below the cutoff are not compressed (Simpson et al., 2005). At this time, frequency compression via linear shifting has yet to be implemented into a commercial hearing aid device (Simpson, 2009). Only one manufacturer is marketing nonlinear frequency compression in a hearing aid (Phonak AG). Consequently that algorithm will be the focus of remaining discussion.

As shown in the introduction, Figure 1 describes how signals are processed in Phonak's Sound Recover frequency compression algorithm. This diagram illustrates how the low- and high-pass bands process a given input signal independently before they are summed to create the output signal. There are two programmable parameters that define how frequency compression is applied to the input signal. These include the *cutoff frequency*, or the frequency that separates the low- and the high-pass bands, and the *compression ratio*, or the amount of frequency compression calculated by a ratio of the input over output bandwidth. The frequency compression algorithm is constantly active and there is no overlap in the output signal between the compressed frequencies in the high-pass band and the uncompressed frequencies in the low-pass band.

2.2.3.2 Studies

Simpson et al. (2005) initially investigated the benefits of nonlinear frequency compression for experienced hearing aid users with sloping, moderate-to-severe sensorineural hearing loss. In their study, a frequency compression exponent of 0.5 was set for all subjects and the cutoff frequency was selected based on the audiogram of each subject. Verification of the frequency-compression device was achieved using a loudness balancing procedure which equated loudness for frequencies 1.6, 2, 2.5, 3.15, 4, and 5 kHz. The stimuli were narrow-band noises presented in the sound field from a single loudspeaker at 0° and 1 m from the subject. Amplification of the compressed signal was adjusted by the clinician until that signal was equal in loudness to the reference, or non-compressed, signal. Each subject wore the experimental device for four to six weeks, and then reverted back to their own conventional hearing aid for two weeks. Results from the CNC word recognition test in quiet (Peterson & Lehiste, 1962) revealed that 8 out of 17 subjects obtained significantly better phoneme scores with the experimental device whereas one subject demonstrated a significant decrease in performance compared to their conventional hearing aids.

A follow-up study (Simpson et al., 2006) evaluated the impact of nonlinear frequency compression for subjects with steeply-sloping hearing losses. Using a similar research design and the same frequency compression algorithm as in Simpson et al. (2005), subjects wore bilateral conventional hearing aids for four to five weeks, then bilateral frequency-compression hearing aids for four to six weeks. The hearing aids were fit based on the NAL-NL1 prescriptive formula (Byrne et al., 2001) and the frequency compressed signals were verified for frequencies 1, 1.25, 1.6, 2, 2.5, 3.1, and 4 kHz using the same loudness balancing procedure as in Simpson et al. (2006). Performance with the conventional and frequency-compression hearing aids was measured using tests of word recognition in quiet, consonant recognition in quiet, and sentence recognition in noise. On average, results revealed no statistical differences between the conventional and frequency compression hearing aids on any of the test measures. However, for two subjects, significant improvements were found using the frequency-compression hearing aids over the conventional hearing aids. Finally, self reported benefit was assessed using the APHAB (Cox & Alexander, 1995) and revealed higher scores using the conventional hearing aids for four of six subjects. For the remaining two subjects, one subject reported better scores on the APHAB with frequency compression compared to conventional amplification and one subject reported no difference between the two devices.

Similarly, Glista et al. (2009) compared the benefits of frequency compression to conventional amplification for tests of speech perception and speech sound detection for 11 children and 13 adults with sloping high-frequency hearing loss. In their study, several phases were implemented, including: 1) a four-week home trial with conventional hearing aids, 2) a 10-week home trial with frequency compression, and 3) a six-week home trial where the user could select either conventional amplification or frequency compression using multiple memories. Initially, the conventional hearing aids were fit to DSL prescriptive targets (DSL v5: Bagatto et al., 2005; Scollie et al., 2005) and simulated real-ear measures were obtained to verify the accuracy of the fit incorporating individual real ear to coupler differences (RECDs). Next, the frequency-compression hearing aids were fit to provide audibility of high-frequency sounds (e.g., /s/ and /sh/) using listening checks and aided spectra performed by the fitter. This method allowed for the frequency compression to be set without subjective feedback. Testing was performed after the six and ten week trials on outcome measures of aided speech sound detection, consonant recognition in quiet, vowel recognition in quiet, and plural recognition in noise. Subjects also completed a diary during the six-week trial to document performance differences or preferences with and without frequency compression.

Based on group scores, results from Glista et al. (2009) revealed improved (lower) aided thresholds with frequency compression, significantly better consonant and plural recognition with frequency compression, and no significant change in vowel perception. Analyses of the individual data found significantly improved sound detection of /s/ or /sh/ using frequency compression for five children and five adults, but significantly worse detection for one child and one adult compared to conventional amplification. For speech perception, five adults and seven children demonstrated significantly improved performance on at least one test using frequency compression. There was one case of an adult subject that showed significantly better vowel perception using conventional amplification compared to frequency compression.

Additionally, multiple linear regression analyses were completed in this study to compare subject variables to the scores on the outcome measures (Glista et al., 2009). Three factors emerged as significant predictors of performance: age group, the magnitude of high-frequency hearing loss, and the audiometric drop-off (i.e., frequency at which thresholds $\geq$70 dB HL in the better ear). More specifically, subjects with more high-frequency hearing loss that occurred at higher frequencies demonstrated more benefit using frequency compression. Also, children with better plural recognition using frequency compression favored frequency compression compared to adults as evidenced by subjective preferences through the self-report measure.

Nyffeler (2008) reported improved speech perception and sound quality with frequency compression for 11 subjects with moderately severe-to-profound hearing loss. All subjects were experienced hearing aid users and wore the Phonak Naida UP frequency-compression hearing aids for two months. Performance was initially tested with their own conventional hearing aids and again after two months with the frequency-compression hearing aids. The tests consisted of 1) a German sentence-in-noise test where the noise was presented at 65 dB SPL and the speech stimuli were adaptively varied to determine the signal-to-noise threshold where the listener could identify 50% of the words in the sentence, and 2) a subjective questionnaire administered at intervals of 2 weeks, 1 month, and 2 months to determine the influence of frequency compression on sound quality judgments and overall speech perception. Results indicated no significant difference for sentence recognition in noise using the frequency-compression hearing aids and the conventional hearing aids. Results from the subjective questionnaire, however, found significantly higher satisfaction ratings using frequency compression compared to their own hearing aids for speech perception in quiet and in noise. Additionally, sound quality judgments (e.g., perception of fricatives) based on group data improved from one to two months using frequency compression, indicating that acclimatization to frequency compression took a relatively short time. In that study, no aided thresholds or real ear measures were obtained to confirm “goodness” of fit.

O’Brien et al. (2010) recently investigated the impact of frequency compression on horizontal localization and speech perception in a background of noise. Twenty-three hearing aid users with moderate-to-severe hearing losses were fit with bilateral Phonak Naida V SP hearing aids and followed for eight weeks with and without frequency compression. The hearing aids were fit using the NAL-NL1 prescription (Byrne et al., 2001) and real-ear measures were obtained to verify the goodness of fit. As reported by the authors, the frequency compression ratio was set in each ear to be “audible but not bothersome when listening to the sentence ‘She sells sea shells by the seashore’”. Horizontal localization performance was tested using a 360° array of 20 loudspeakers, each separated by 18°, using broadband, pulsed, pink noise stimuli presented at 70 dB Leq. Localization accuracy was judged based on the number of front/back confusions by comparing the difference in RMS error for the presentation versus response azimuth along the front/back dimension. For the speech perception task, the stimuli were 50 monosyllables presented from a loudspeaker at 0° azimuth, with an eight-talker babble noise presented loudspeakers at 0° and 180° azimuth. The noise stimuli were adaptively varied in 1-dB steps to find a signal-to-noise threshold where the listener could identify 50% of the monosyllabic words. All tests were performed at one, four, and eight weeks following fitting of the frequency-compression hearing aids. Finally, the Speech, Spatial and Qualities of Hearing scale (SSQ, Gatehouse & Noble, 2004) was collected at eight weeks post-fitting.

The results revealed no significant difference in localization performance along the front/back dimension using frequency compression compared to conventional amplification. It is important to note, however, that the effect neared significance, $F(2, 22) = 4.21$, $p = .052$, revealing that mean localization performance with the frequency-compression hearing aid was 2° poorer than with conventional amplification. For speech perception, no difference was found between frequency compression and conventional amplification when the speech and noise originated from 0° azimuth. With spatially-separated noise sources, frequency compression provided more benefit initially compared to conventional amplification. However, speech perception performance improved overtime with the conventional aids and, after eight weeks, frequency compression no longer provided an advantage over conventional amplification. Finally, results from the SSQ revealed no significant differences between frequency compression and conventional amplification on any of the SSQ subscales.

2.2.3.3 Summary

In summary, two studies reported improvements on speech sound detection and consonant perception using frequency compression compared to conventional amplification for listeners with sloping high-frequency hearing loss (Glista et al; 2009; Simpson et al., 2005). However, other studies have suggested no difference in performance using frequency compression for speech recognition in quiet and in noise compared to conventional amplification (Nyffeler, 2008; Simpson et al., 2006). Individual results from two studies have indicated decrements for at least one subject in speech perception and self-reported benefit using frequency compression compared to conventional amplification (Glista et al., 2009; Simpson et al., 2005; 2006). Finally, one study investigated sound localization for bilateral hearing aid users and results revealed no significant difference using frequency compression compared to conventional amplification (O’Brien et al., 2010).

2.3 Studies of Frequency Lowering for CI+HA

Despite the numerous studies of frequency-lowering signal processing, relatively few studies (Chute et al., 1995; McDermott & Henshall, 2010) have examined the benefits of frequency-lowering signal processing for users of a CI contralaterally.

Chute et al. (1995) studied the contribution of frequency-lowering signal processing to contralateral CI use for five Nucleus 22 implant recipients. All subjects had severe-to-profound hearing loss and were fit in the opposite ear with an AVR Sonovations slow-playback frequency-lowering body aid. For two months, subjects wore their CI plus the frequency-lowering body aid all day except for at least two hours/day where only the frequency-lowering body aid was used (no CI). In their study, the $Z_{C,}$ or lowering factor, was set individually based on the ability to detect narrow-band noise from 0.25 to 4 kHz at 40 dB in the sound field. No other details regarding the fit of the devices were described. Overall performance was assessed using tests of Glendonald Auditory Screening Procedure (GASP) phonemes, GASP word and stress identification, Minimal Pairs, NU-6 words and phonemes, and CUNY sentences presented via auditory only mode. Subjects were tested after one and two months of use and scores were averaged together. Of the five subjects in the study, one subject withdrew from the study because it reportedly made her tinnitus worse and was too bulky with the CI worn contralaterally. Of the remaining four subjects, performance was no different with the CI alone or the CI+HA on any of the test measures. With the frequency-lowering hearing aid alone, two subjects scored above chance on the GASP word and stress identification tests whereas the other two subjects showed no improvement on any test measures. Reportedly, there was also no decrement in performance for these four subjects using the CI+HA compared to the CI alone.

More recently, McDermott & Henshall (2010) examined the benefits of frequency compression for contralateral CI use for eight individuals with severe-to-profound high-frequency hearing loss. Subjects were fit with the Phonak Naida V UP hearing aid and used this for approximately 10 weeks in conjunction with their CI. Different frequency compression settings were evaluated during the 10-week trial. These included frequency compression disabled, frequency compression enabled with default settings from the clinical programming software, and frequency compression enabled at the maximum compression settings (or cutoff frequency = 1.5 kHz; compression ratio = 4:1.1). Hearing aid gain as a function of frequency was initially adjusted in the uncompressed mode to approximate NAL-RP targets (Byrne et al., 1990) using insertion-gain measurements. Then, frequency compression was activated using either default settings in the software based on the audiogram or using the maximum compression settings, with no additional verification. Speech perception was evaluated using two tests: a) consonant identification in quiet, 12 stimuli presented in an /aCa/ format by a female talker; and b) sentence recognition using CUNY sentences, presented in noise with a four-talker babble that adaptively varied to find a signal-to-noise threshold where the listener could identify 50% of the words in the sentence. All stimuli were presented from a single loudspeaker placed at 0° azimuth from the subject. Subjects completed both tests in seven conditions: a) their own hearing aid, b) the Phonak Naida with frequency compression on, c) the Phonak Naida with frequency compression off, d) their CI only, e) their CI plus their own hearing aid, f) their CI plus the Phonak Naida with frequency compression on, and g) their CI plus the Phonak Naida with frequency compression off.

On average, consonant recognition was significantly better when subjects used the CI alone or the CI+HA compared to their own hearing aid (by approximately 50%). There was no significant difference when subjects used the CI alone and the CI+HA, or between the hearing aid with and without frequency compression. In contrast, sentence recognition in background noise was significantly better with the CI+HA compared to the CI or hearing aid alone. Additionally, no significant difference was found on the sentence recognition tests with and without frequency compression. Individual results on the sentence recognition test revealed that when subjects used the conventional hearing aid only, a score could not be obtained for seven of eight subjects. In comparison, using the frequency compression hearing aid, a score could be obtained for all but two subjects. The results of this study indicated that use of frequency compression provided no significant improvement over conventional amplification when a CI was used contralaterally. The benefit of frequency compression over conventional amplification was limited to improvement in sentence recognition using the hearing aid alone and could potentially have been greater if the listeners had better high-frequency thresholds in the aided ear.

2.4 Summary

In conclusion, more recent studies of frequency-lowering signal processing have demonstrated small, yet significant improvements for speech perception and speech sound detection compared to conventional amplification. However, not all subjects have shown better performance using frequency compression and some individual results have indicated decrements in performance compared to conventional amplification. The results suggesting improved sound detection for high frequencies are encouraging. If high-frequency cues are better perceived using frequency compression in the hearing aid ear, it is possible that these listeners would have better sensitivity to ILDs when the CI is used contralaterally. By improving sensitivity to binaural cues, localization and speech perception performance with spatially separate target and noise sources could also improve. To date, only one study has examined the impact of frequency compression on localization performance (O'Brien et al., 2010) and results were analyzed for front-to-back errors only. Furthermore, participants in their study had an average age of 78.7 years, which may have limited generalization of the results.

Of the studies of frequency compression in CI+HA users, there appear to be some potential disadvantages, including the limited number of subjects (n=5, Chute et al., 1995; n=8, McDermott & Henshall, 2010) and relatively poor residual hearing of the subjects. Thus, it may be important to revisit the benefits of frequency-lowering signal processing for CI+HA users with better residual hearing. In addition, the methodology for fitting and verifying frequency-lowering hearing aids varies considerably across studies, and some studies provide no data or evidence that the hearing aids were verified at all.

Finally, it is important to consider new ways to optimize CI+HA fittings and this type of signal processing may provide additional benefits such as improved sound localization.

CHAPTER 3

METHODS

3.1 Experiment 1: Contribution of Frequency Compression to CI+HA Performance

3.1.1 Participants

Ten adult subjects (5 male, 5 female, M_{age} = 60.0 years, age range: 39-82 years) with a CI+HA and aidable residual hearing in the opposite ear from the implant participated in the experiment. Candidacy criteria were as follows: subjects 1) had a history of postlingual-deafness, 2) had been implanted with a full electrode array, 3) had been implanted monaurally with an Advanced Bionics or Cochlear Corporation device, 4) had a moderate-to-severe hearing loss in the contralateral ear to the implant, and 5) had not worn this type of frequency-lowering hearing aid previously. The participants' age of onset of hearing loss ranged from 5-57 years (M_{age} = 24.7 years) and all subjects were experienced hearing aid users (see Table 3.1 for detailed demographic information). All participants were recruited from the University of Iowa Cochlear Implant Center. The length of CI use for the participants varied from 0.5 years to 7.4 years with an average of 2.9 years. All participants were compensated $100 for their participation plus complementary parking during the two test sessions of the study.

3.1.2 Procedures

Subjects were fit with the Phonak Naida IX SP hearing aid contralaterally to the CI, and the hearing aid was programmed with and without frequency-compression signal processing activated. Two subjects were fit with a Phonak Naida V UP (ultra power) hearing aid because the Naida IX SP hearing aid did not provide sufficient gain. All subjects were required to alternate daily among the two hearing aid programs for two months, thus providing four weeks of wear time for each hearing aid condition. Existing literature indicates that listening time or experience with frequency-lowering signal processing is necessary for listeners to achieve better performance (Braida et al., 1979;

Simpson et al., 2009). On average, previous studies have provided four weeks of listening time for listeners to adjust to a new frequency-lowering signal processing strategy using a take-home device (Gifford et al., 2007; Glista et al., 2009; McDermott & Knight, 2001; Parent et al., 1997; Simpson et al., 2005, 2006). By giving each subject two months in the study while alternating daily among the two hearing aid programs, each signal processing (frequency-compressed and conventional) was used for approximately four weeks. Additionally, subjects were blinded to the programs in the hearing aid so as not to bias the results. Compliance with this method of alternating programs was judged based on a data logging feature which allowed for tracking the use of each of the two programs in the hearing aid on a daily basis. Programs were assigned in a counter-balanced manner across participants.

There were two data-gathering sessions in this experiment. In the first session, subjects consented to participate in the study, and unaided thresholds were measured at all octave and inter-octave frequencies from 125 to 8000 Hz in the non-implanted ear (refer to Figure 3.1). Earmold fittings were assessed for each subject and new earmolds were ordered if the subject's own earmold was not sufficient, or if they did not have an earmold. The CI and hearing aid were programmed and verified for each subject and in each hearing aid program (see sections 3.1.2.1 and 3.1.2.2. for details). After the hearing aid was programmed, subjects were tested on a battery of tests to establish baseline performance (refer to section 3.1.3. for test measures). All test measures were administered in a sound-treated IAC booth in the Audiology department of the University of Iowa Hospitals and Clinics. For all tests, the loudspeaker outputs were calibrated at 1.4 m from the sound source using a Larson-Davis 824 sound level meter. Calibration of all test measures was checked prior to each session.

Following two months of hearing aid experience, subjects returned for session two. At this session, testing was completed with the CI and hearing aid with and without frequency-compression signal processing. This concluded their participation in the study.

3.1.2.1 Cochlear Implant Programming

To eliminate any learning effects that could be introduced by altering the subjects' cochlear implant programs, the subjects' preferred listening program prior to the onset of the study was used. T- and C- or M-levels were measured using standard clinical procedures only when reprogramming of the cochlear implant speech processor was needed. If a patient utilized adaptive directional microphones, ADRO, Autosensitivity, or other noise reduction/cancellation features routinely, these were not active during the testing.

3.1.2.2 Hearing Aid Programming

The hearing aid was programmed using the Phonak I-PFG software v. 2.4 in combination with the Fitting Assistant software (Alexander, 2009) and NOAH hearing aid software. Unaided air conduction thresholds were obtained and these values were entered into the hearing aid software to begin the fitting. The hearing aid was programmed for two memories only, with and without frequency compression (i.e., enabled vs. disabled). Noise reduction, directional microphones, and wind noise management schemes were deactivated in the hearing aid for verification and during the home trials.

3.1.2.3 Hearing Aid Verification

The NAL-NL1 (Byrne et al., 2001) hearing aid prescription was used to set hearing aid gain. Given that the NAL prescription is fitting the hearing aid to an average response, this approach was utilized to provide sufficient and consistent gain recommendations for all subjects in the study. Real ear measurements were obtained using the Verifit real ear analyzer for speech inputs presented at 55, 65, and 75 dB SPL and a maximum power output (MPO) at 85 dB SPL. Subjects were seated at 0° azimuth to the loudspeaker for real ear measurements and gain adjustments were made to achieve the prescriptive targets. Verification of the conventional hearing aid fit was quantified by comparing the overall aided long-term average spectrum of speech (LTASS) to the NAL-NL1 target at 500, 1000, and 2000 Hz. Acceptable fitting error was defined as +/- 3 dB. If the output fell outside of the acceptable error range, gain was adjusted on the hearing aid as necessary to be within the RMS error of +/- 3 dB. Figure 3.2 illustrates an example of the aided LTASS relative to NAL targets for the conventional hearing aid. Next, audibility was calculated based on estimations of the Speech Intelligibility Index (SII) (ANSI S3.5-1997) for the conventional fitting and the frequency-compressed fitting. For the frequency-compressed scheme, a modified LTASS (Bentler, Cole, & Wu, 2011) was presented at 65 dB SPL and one-third octave speech bands centered at 3150, 4000, 5000 and 6300 Hz were used to quantify the lowered audibility. Figure 3.3 shows an example of the isolated one-third octave speech bands used in deriving the SII calculation for the frequency-compressed hearing aid. The derivation of the lowered SII was accomplished using software that has been modified for this purpose by the developer of the Verifit probe microphone system (refer to Appendix B for a more detailed description).

3.1.2.4 Setting the Cutoff Frequency and Compression Ratio

The parameters of frequency compression were set using the SoundRecover Fitting Assistant v. 1.05 software (Alexander, 2009). In this program, a fuzzy logic model is used to select the appropriate cutoff frequency setting for a given individual, based on four variables: 1) audible output bandwidth, or the maximum range of frequencies calculated from the output of the hearing aid fitting using conventional amplification, 2) audible input bandwidth, or the maximum amount of information in the audible frequency range, 3) the cutoff frequency, and 4) the compression ratio.

To begin using this program, the hearing aid model was initially selected. Next, the audible output bandwidth was measured when frequency compression was deactivated using the Verifit probe microphone system and this was entered into the software program. Frequency compression was activated in the hearing aid and the default frequency compression setting recommended by the manufacturer, consisting of the cutoff frequency and compression ratio, was noted. This setting was entered into the program and the maximum audible input bandwidth was calculated. Based on the results and interactions between the four variables, the program recommended a setting that fit the bandwidth of compression within the individual's region of audibility using the highest cutoff frequency and lowest compression ratio possible. To visually determine how the input could be maximized relative to the output bandwidth, the software provided a graph of each setting showing input versus output frequency.

Table 3.2 displays the unaided thresholds, SII values, and parameters of frequency compression for each subject. The appropriateness of these settings was verified prior to each subject beginning their at-home trial. That is, using subsequent probe-microphone measurements, it was determined that the frequency-compression scheme was providing adequate audibility for high-frequency inputs.

3.1.2.5 Verification of CI+HA Fitting

Verification of CI+HA programs was completed by obtaining aided sound field thresholds at all octave and inter-octave frequencies (in dB HL) using narrow-band noise presented from a clinical audiometer from 125 to 8000 Hz (see Figure C1, Appendix C). Due to time constraints while testing, aided sound field thresholds were not measured for subject S2. For subject S5, sound field thresholds using the frequency-compression hearing aid could not be reliably measured and, therefore, were not reported. Table 3.3 shows the change in sound field thresholds and SII value for the subjects in experiment one following fitting of the frequency-compression hearing aid as compared to the conventional hearing aid fit.

A loudness-matching method was also performed using the clinician's voice facing the participant at 0° azimuth to equate loudness bilaterally. Participants were asked to assign a category to the voice: cannot hear, very soft, soft, comfortable but slightly soft, comfortable, comfortable but slightly loud, loud, very loud, and too loud (Ricketts & Bentler, 1996).

3.1.3 Outcome Measures

3.1.3.1 Localization Test

Two test measures were used to evaluate the contribution of frequency compression to CI+HA performance. First, an 8-loudspeaker Everyday Sounds Localization test was presented in quiet (Dunn, Tyler, & Witt, 2005). Sixteen everyday sounds (i.e. glass breaking, a knock at the door, child laughing, etc.) were presented at 70 dB SPL(C) from eight loudspeakers, each 15.5° apart, forming an 108° arc in front of the subject. The sixteen stimuli were each presented six times for a total of 96 trials. Stimuli were presented from each loudspeaker 12 times throughout the test. Subjects could see all eight loudspeakers placed in the horizontal arc in front of them. The subjects were instructed to face forward and restrict head movement during the duration of the test. Each subject selected the speaker from which the sound originated using a touch screen monitor placed in front of them. No feedback was provided throughout the test. Scores were reported by the average total root mean square (RMS) error, a single number that represented the variability in localization performance among the eight loudspeakers. Lower scores on the test (e.g., RMS error of 10 to 20º) represented better localization abilities. The localization test was administered in the binaural condition, CI+HA, with and without frequency-compression signal processing (Table 3.4 describes the conditions tested).

3.1.3.2 Spondee-in-noise Test

The second test measure used to evaluate the contribution of frequency compression to CI+HA performance was a spondee-in-noise test. The Multiple Jammer spondee-in-noise test (Hawley et al., 1999) was administered using the same eight loudspeaker array as described for the localization test. Twelve spondee words (Turner et al., 2004), equal in difficulty level, were presented from one of two loudspeakers at +8 and -8 degrees from the midline. The adaptive masker noise consisted of a two-talker (one female and one male) voice repeating sentences and played from two of four loudspeakers which were placed at opposite sides of the subject. More specifically, the masker noise alternated between a right and left loudspeaker on opposite sides of the subject as the spondee word alternated between a right and left loudspeaker placed in front (i.e., from either +54° and -38.5° when presented from the loudspeaker at +8 degrees, or +38.5° and -54° when speech was presented from -8 loudspeaker). The presentation level of the spondee word was set to be identical for each subject across all listening conditions, at 55 or 65 dB SPL(C). The level of the masker noise varied with each trial. A 2-dB adaptive procedure with a total of 14 reversals was used to find the signal-to-noise ratio (SNR) that produces a 50% correct level for the spondee words. The final SNR was determined by taking an average of the SNR from the last two of three runs. Subjects were asked to select a spondee from the 12 possible choices using a touch screen monitor placed in front of them. Data were obtained for the Multiple Jammer test using five different conditions: CI+HA with and without frequency-compression signal processing, CI only, and the hearing aid only with and without frequency-compression signal processing (see Table 3.4).

3.1.3.3 Questionnaires

Two questionnaires, the Spatial Hearing Questionnaire and a Sound Quality Questionnaire, were administered to document self-reported outcomes. The Spatial Hearing Questionnaire is a 24 item subjective outcome measure (Tyler et al., 2009) used to evaluate perceived spatial hearing performance of listeners with CIs and hearing aids. The questionnaire utilizes eight subscales to assess voice perception (male, female, and child), music perception, source localization, and speech understanding (in quiet and noise). Subjects were asked to complete the questionnaire at the end of the study following the at-home trial to document performance with and without frequency-compression using the CI+HA. Scores on the Spatial Hearing Questionnaire were reported for the eight subscales as well as the total score.

The Sound Quality questionnaire consisted of six items and was used to assess the quality of sound with the CI and hearing aid, with and without frequency compression. The items included speech understanding, quality of sound, recognizing everyday sounds, distortion or noise in the hearing aid, music listening, and voice quality. Subjects were asked to rate each item on a scale of 0 to 100, where 0 indicated very poor and 100 indicated excellent. Subjects completed the questionnaire at the conclusion of the home trial using the CI+HA with and without frequency compression. This questionnaire was added to the test protocol after the first two subjects completed the study due to their subjective reports of sound quality differences between the frequency-compression and conventional hearing aids.

3.1.4 Statistical Analysis

For the localization test, data were analyzed using two paired *t*-tests to compare performance for the two listening conditions, CI+HA-conventional and CI+HA-frequency compressed. This was calculated by comparing the difference between the listening conditions after two months listening experience. A second analysis was used to compare the rate of change overtime between the two conditions, or amount of change overtime for the CI+HA-compressed condition relative to the CI+HA-conventional condition. This was calculated as follows: a) the difference in performance overtime (two month – baseline) was computed for each listening condition, and b) a paired t-test was performed between the two listening conditions.

Individual data on the localization test was compared for the CI+HA-conventional and CI+HA-frequency compressed conditions at two months using a paired *t*-test. This was calculated by comparing the means of the squared error between the stimulus and response loudspeakers, computed in degrees, between the two listening conditions. More specifically, the stimulus and response speaker numbers were initially converted to degrees. Next, the stimulus speaker degree was subtracted from the response speaker degree. The difference between speaker and response speaker was then squared. Each row in the data structure was matched for responses to the same speaker between the two conditions. Finally, the paired *t*-test was computed comparing the distribution of responses between the two listening conditions for each individual.

For the spondee-in-noise test, data similarly were analyzed using paired *t*-tests to compare performance at two months as well as the difference in rate of change from baseline to two months for the CI+HA-conventional and CI+HA-frequency-compressed conditions. In addition, performance was compared at two months using paired *t*-tests for all five listening conditions, including the CI+HA with and without frequency-compression signal processing, the CI only, and the hearing aid only with and without frequency-compression signal processing. Individual results on the spondee-in-noise test were also analyzed by comparing performance with the CI+HA-conventional and CI+HA-frequency-compressed conditions at two months. For each individual, a score was considered significant if the difference score was greater than 2.92 SNR, or the critical difference score calculated from pilot testing of four adult CI+CI users. The critical difference score for this test at a 95% confidence level was found by calculating the difference (or standard deviation) in SNR for the average of two of three runs following five repeated administrations.

Data from the two questionnaires, the Spatial Hearing Questionnaire and the Sound Quality Questionnaire, were analyzed using paired *t*-tests to compare performance for the CI+HA-conventional and CI+HA-frequency-compressed conditions. This comparison was computed based on scores after two months of use because the questionnaires were administered at the two month interval only.

The intended sample size for this study was approximately 15-20 subjects. This was determined based on the results of a power analysis (Statistical Analysis Software (SAS) Proc Power v. 9.2) completed prior to the onset of the study. Power was calculated based on the comparison of two frequency-lowering conditions and various mean differences were explored. A high correlation was predicted due to the close relationship among the two frequency-lowering conditions, thus a correlation of 0.6 to 0.7 was used. The predicted standard deviation was obtained from results on CNC phoneme scores from an existing study by Simpson et al., 2005. The resulting power calculation revealed 15-20 subjects would be required to obtain calculated power of 0.80. Because the actual sample size was smaller than the target, paired *t*-tests were used to compare performance differences rather than other types of analyses, such as a repeated measures ANOVA.

All statistical analyses were performed using the Statistical Package for the Social Sciences (SPSS) v. 17.0 and v. 19.0. Statistical significance was defined as $p < 0.05$ for all tests.

3.2 Experiment 2: Impact of Frequency Compression on Speech Perception

3.2.1 Participants

Nineteen adult subjects with a CI+HA or bilateral hearing aids (HA+HA) and aidable residual hearing were recruited for this study. Candidacy criteria were as follows: subjects 1) had a moderate or moderate-to-severe hearing loss in the test ear, and 2) had not worn this type of frequency-lowering hearing aid previously. All CI+HA subjects that participated in the first experiment also participated in experiment two. The CI+HA subjects were implanted monaurally with a standard length CI and had post-lingual onset of deafness. There were two subjects that withdrew from the study. Thus, participants included seventeen adult subjects (10 male, 7 female, $M_{\text{age}} = 61.8$ years, age range: 39-82 years) with the onset of hearing loss ranging from 5-74 years ($M_{\text{age}} = 34.9$ years). All subjects had hearing aid experience prior to the onset of the study (see Table 3.5 for subject demographic information). All participants were compensated $100 and provided complementary parking during the two test sessions of the study.

3.2.2 Procedures

As in experiment one, subjects were fit with the Phonak Naida IX SP hearing aid and the hearing aid was programmed with and without frequency compression. For HA+HA subjects, the Phonak Naida IX SP hearing aids were fit bilaterally. Two CI+HA subjects were fit with a Phonak Naida V UP hearing aid due to insufficient gain with the Naida IX SP model. All subjects were required to alternate daily among the two hearing aid programs for two months and compliance with alternating programs was judged based on a data logging feature which allowed for tracking the use of each of the two hearing aid programs. Subjects were kept blind to the hearing aid programs which were assigned in a counter-balanced manner across all subjects.

At the first study visit, subjects consented to participate in the study, and unaided thresholds were measured at all octave and inter-octave frequencies from 125 to 8000 Hz in the contralateral ear for CI+HA subjects, and bilaterally for HA+HA subjects (refer to Figure 3.5). Earmold fittings were assessed for each subject and new earmolds were ordered if the fit of the subject's own earmold was not sufficient, or if they did not have an earmold. The hearing aid(s) was programmed and verified for each subject and in each hearing aid program. After the hearing aid programs are fit, subjects were tested monaurally to establish baseline speech perception performance (see below for test measures). For CI+HA subjects, the CI was removed during testing. Likewise, one hearing aid was removed during testing of HA+HA subjects and an ear plug was placed in the ear. Data were collected in a sound-treated booth in the Audiology department of the University of Iowa Hospitals and Clinics. For all tests, the audiometer and loudspeaker outputs were calibrated at 1 m from the sound source using a Larson-Davis 824 sound level meter. Calibration of all test measures was checked prior to each session.

Following two months of hearing aid experience, subjects returned for session two. At this session, testing was completed with the hearing aid only, with and without frequency compression. For HA+HA subjects, the identical hearing aid that was removed at baseline was also removed during the final testing session. This concluded participation in the study.

3.2.2.1 Hearing Aid Programming

The hearing aid was programmed using the Phonak I-PFG software v. 2.4 in combination with the Fitting Assistant software (Alexander, 2009) and NOAH hearing aid software. Unaided air conduction thresholds were obtained and these values were entered into the hearing aid software to begin the fitting. The hearing aid(s) was programmed for two memories only, with and without frequency compression (i.e., enabled vs. disabled). Noise reduction, directional microphones, and wind noise management schemes were deactivated in the hearing aid for verification and during the home trials.

3.2.2.2 Hearing Aid Verification

Initially, the NAL-NL1 (Byrne et al., 2001) hearing aid prescription was used to set gain for the hearing aid. Real ear measurements were obtained using the Verifit real ear analyzer for speech inputs presented at 55, 65, and 75 dB SPL and MPO at 85 dB SPL. Subjects were seated at 0° azimuth to the loudspeaker for real ear measurements and gain adjustments were made to achieve the prescriptive targets. Verification of the conventional hearing aid fit was quantified by comparing the overall aided LTASS to the NAL target at 500, 1000, and 2000 Hz. Acceptable fitting error was defined as +/- 3 dB. If the output fell outside of the acceptable error range, gain was adjusted on the hearing aid as necessary to be within the RMS error of +/- 3 dB. Real ear measurements and gain adjustments were made bilaterally for the HA+HA subjects to approximate the NAL targets for each ear. As in experiment one, audibility was calculated based on estimations of the Speech Intelligibility Index (SII) (ANSI S3.5-1997) for the conventional fitting and the frequency-compressed fitting. For the frequency compression scheme, a modified LTASS (Bentler, Cole, & Wu, 2011) was presented at 65 dB SPL and one-third octave speech bands centered at 3150, 4000, 5000 and 6300 Hz were used to quantify the lowered audibility (see Figure 3.3). Appendix B provides a more detailed description of the modified LTASS and the calculation to verify the lowered audibility for frequency-compression hearing aids.

3.2.2.3 Setting the Cutoff Frequency and Compression Ratio

The parameters of frequency compression were set using the SoundRecover Fitting Assistant v. 1.05 software (Alexander, 2009), as described in experiment one. For HA+HA subjects, the cutoff frequency and compression ratio were set identical for both hearing aids. Table 3.6 displays the unaided thresholds, SII values, and programming parameters for frequency compression for each subject in experiment two.

3.2.2.4 Verification of HA fitting

Verification of the hearing aid programs was completed by obtaining aided sound field thresholds in the test ear at all octave and inter-octave frequencies (in dB HL) using narrow-band noise presented from a clinical audiometer from 125 to 8000 Hz (see Figure C2, Appendix C). Although aided sound field thresholds were collected bilaterally for HA+HA users in each hearing aid program, results in Figure C2 are shown for the test ear only. Due to time constraints while testing, aided sound field thresholds were not measured for subject S2. Additionally, data for subject S5 using the frequency-compression hearing aid could not be reliably measured and therefore, were not reported. Table 3.7 displays the change or improvement in sound field thresholds as well as in SII value for the subjects in experiment two following fitting of the frequency-compression hearing aid compared to the conventional hearing aid.

For all subjects, the loudness of the two hearing aid programs was assessed using the clinician's voice facing the participant at 0° azimuth. Participants were asked to assign a category to the voice: cannot hear, very soft, soft, comfortable but slightly soft, comfortable, comfortable but slightly loud, loud, very loud, and too loud (Ricketts & Bentler, 1996).

3.2.3 Outcome Measures

3.2.3.1 Iowa Consonant Recognition Test

The Iowa Consonant Recognition Test (Tyler et al., 1984) was used to assess consonant perception in a quiet background. This test incorporates 13 different consonants {d, v, k, g, n, f, s, sh, t, m, z, p, z} presented in an /iCi/ format from a front facing loudspeaker at a distance of 1m from the subject. Each consonant was presented six times for a total of 78 total items. Subjects were asked to select which of the 13 consonants they heard and enter their choice using a touch screen monitor. Two lists were presented, one using a male talker and another with a female talker. The stimuli were calibrated separately for each list to ensure a constant presentation level of 70 dB SPL(C). Prior to the start of the test, subjects were familiarized with the consonant tokens on the touch screen. The final consonant recognition score was determined by averaging the two scores from the male and female individual lists. This test was administered in the following test conditions: the hearing aid only, with and without frequency compression (see Table 3.8 that displays the conditions tested).

3.2.3.2 Iowa Vowel Recognition Test

The Iowa Vowel Recognition Test (Tyler et al., 1984) was used to assess each subject's vowel perception ability in a quiet background. In this test, nine vowel tokens were each presented six times in an /hVd/ format {had, hawd, head, herd, heed, hid, hood, hud, and who'd} from a front-facing loudspeaker at a distance of 1 m from the subject. The stimuli were presented at 70 dB SPL(C). Subjects were instructed to select the appropriate vowel token using a touch screen monitor placed in front of them. All participants were familiarized with the vowel combinations prior to testing. This test was administered in the following test conditions: the hearing aid only, with and without frequency compression (see Table 3.8 for more details).

3.2.4 Statistical Analysis

For the consonant and vowel perception tests, data were analyzed using paired *t*-tests, as discussed for the localization test in experiment one, to compare performance between the two listening conditions: hearing aid only, with and without frequency compression. This was calculated by comparing the difference in performance at two months for the two listening conditions as well as the difference in rate of change from baseline to two months between the listening conditions.

Individual scores on the consonant and vowel perception tests were analyzed to determine if scores were significantly different between the two listening conditions: hearing aid only, with and without frequency compression, after two months of listening experience. Critical difference scores at a 95% confidence level (Carney and Schlauch, 2007) were derived based a list length of n = 50 for the vowel test and n = 100 for the consonant test. The total items on the vowel and consonant tests in this study were n = 54 and n = 178, respectively. Statistical significance was defined at $p < .05$ for the consonant and vowel perception tests.

Table 3.1. Demographic and device use information for subjects in experiment one.

Subject ID	Gender	Age (yrs)	Age of onset (yrs)	Etiology	CI Type	Length of CI use (yrs)	Previous HA	Length of HA use (yrs)
S2	M	75	57.0	Ototoxicity, noise	1	0.8	ITE	25.0
S3	F	41	5.0	Unknown	2	0.8	ITE	34.0
S4	F	71	14.0	Hereditary	1	3.3	ITE	29.0
S5	M	58	18.0	Noise, Hereditary	3	7.4	BTE	11.0
S6	F	49	11.0	Unknown	2	1.1	ITE	37.0
S7	M	72	55.0	Infection, noise	2	1.3	ITE	17.0
S8	F	60	12.0	Unknown	4	6.9	BTE	46.0
S10	M	39	16.0	Unknown	5	0.6	BTE	15.0
S15	F	53	24.0	Unknown	5	0.5	BTE	29.0
S16	M	82	35.0	Meniere's disease	1	6.8	BTE	50.0
Average		60	24.7			2.9		29.3

Note: Age of onset refers to the age at which subject first noticed hearing loss, Etiology = cause of hearing loss; Noise = noise-induced hearing loss; CI Type, 1 = Advanced Bionics 90K, 2 = Nucleus CI24RE, 3 = Nucleus CI24R, 4 = Nucleus RP8, 5 = Nucleus CI512; Previous HA = type of hearing aid used by the subject prior to the study; ITE = in-the-ear hearing aid; BTE = behind-the-ear hearing aid; Length of HA use = length in years of hearing aid use prior to onset of study.

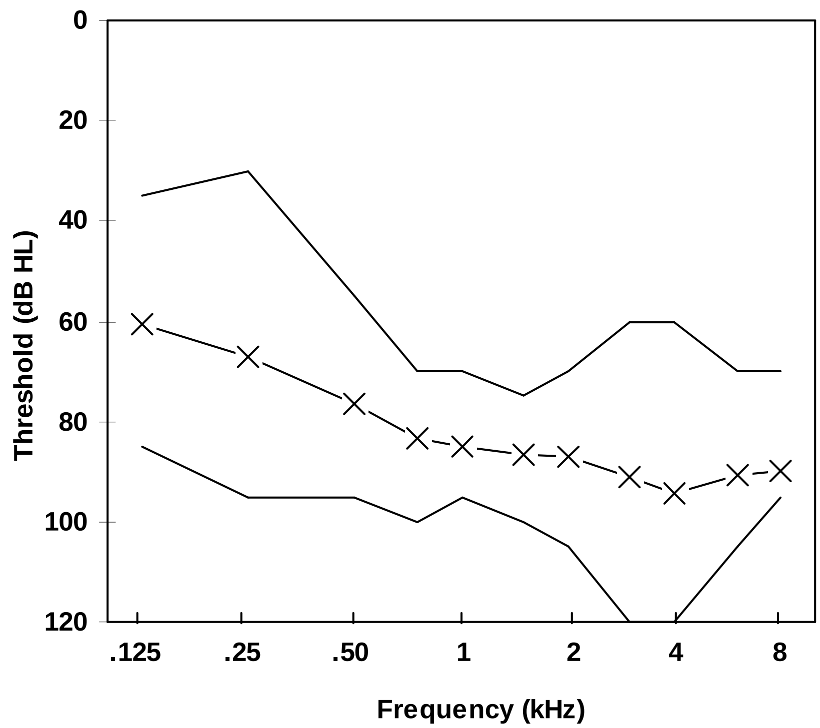


Figure 3.1. Unaided audiometric thresholds for CI+HA participants for the contralateral ear to the implant. Frequency is shown along the x-axis from 0.125 to 8.0 kHz and threshold is shown (in dB HL) along the y-axis. Average audiometric thresholds are indicated by the x's and the upper and lower range of thresholds are shown by the lines.

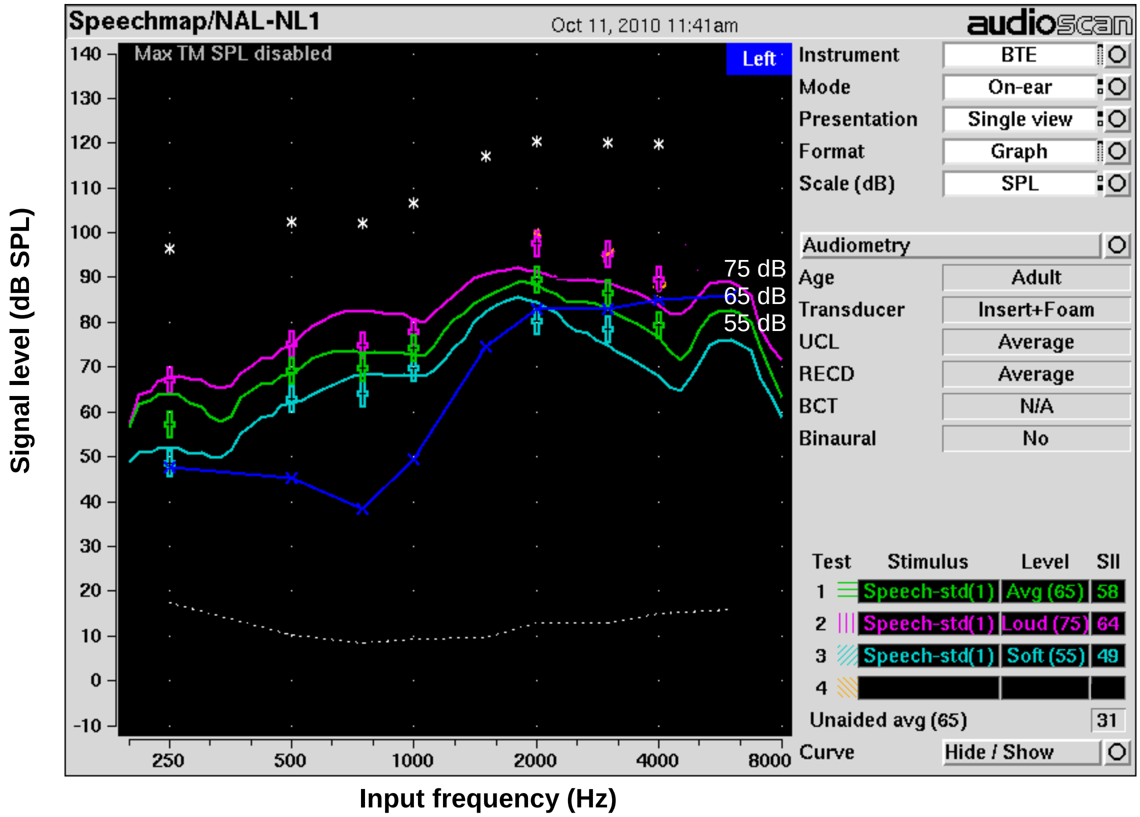


Figure 3.2. Aided LTASS for the conventional hearing aid using various speech inputs. The x-axis displays the input frequency (in Hz) and the y-axis displays signal level (in dB SPL). Unaided hearing thresholds are plotted by the x's and the output of speech for the conventional hearing aid for input levels of 55 dB (lower line), 65 dB (middle line), and 75 dB (upper line) are shown.

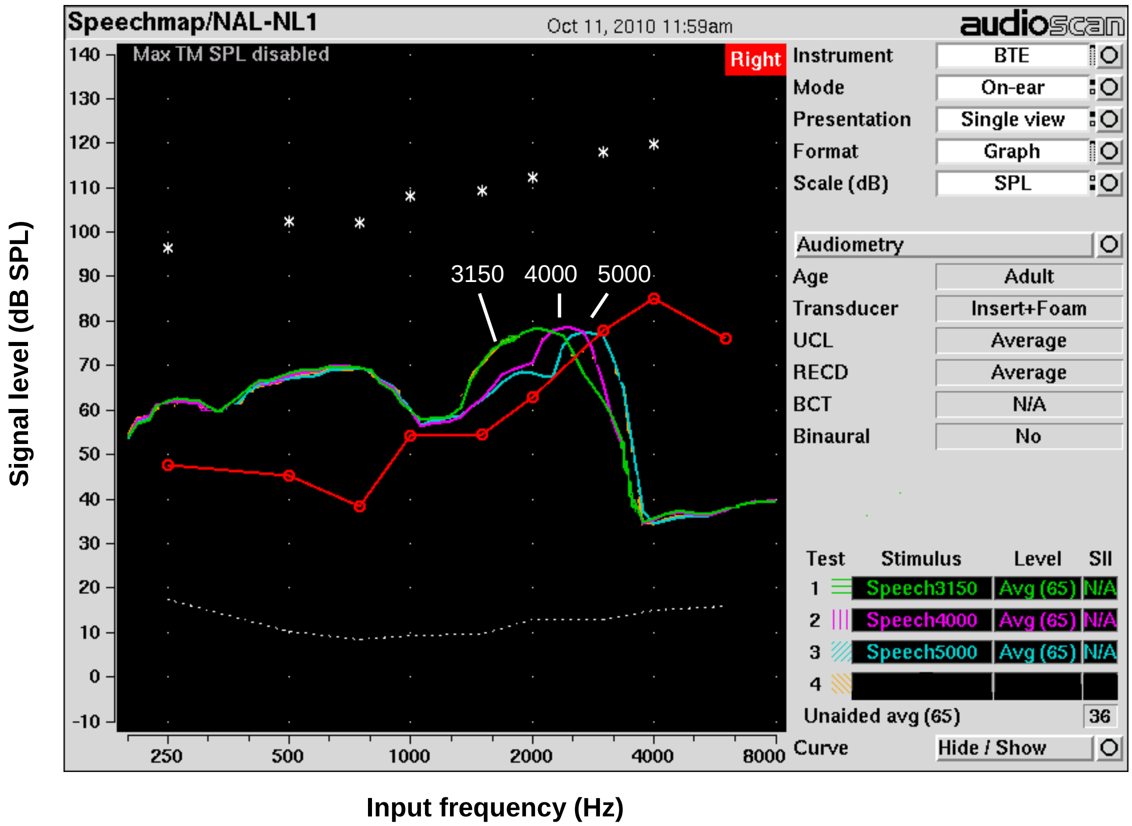


Figure 3.3. Aided LTASS for the frequency-compression hearing aid using three isolated one-third octave bands. The x-axis displays the input frequency (in Hz) and the y-axis displays signal level (in dB SPL). Unaided hearing thresholds are plotted in the circles and the output relative to those thresholds of the isolated 3150 Hz, 4000 Hz, and 5000 Hz one-third-octave speech bands are shown with the frequency-lowering algorithm activated.

Table 3.2. Unaided audiometric thresholds, SII calculations, and compression settings for all subjects.

	Unaided thresholds (in dB HL)					SII – Conv	SII— Comp.	Default CF	Default CR	Fitting assistant CF	Fitting assistant CR
	500	1000	1500	2000	4000			(Hz)		(Hz)	
S2	55	85	90	105	105	0.246	0.247	2.1	1.5	1.5	4.1
S3	65	75	80	80	90	0.379	0.408	2.8	1.9	1.5	1.6
S4	75	70	75	70	60	0.465	0.470	3.9	2.5	4.8	2.6
S5	85	80	75	70	75	0.373	0.414	3.4	2.2	1.6	1.7
S6	90	90	90	90	90	0.228	0.260	2.1	1.6	1.5	3.4
S7	60	80	90	95	110	0.273	0.273	2.1	1.5	1.5	4.1
S8	85	85	80	80	85	0.213	0.240	2.7	1.8	1.5	2.6
S10	75	95	100	100	120	0.229	0.229	2.0	1.5	1.5	4.1
S15	80	95	95	90	90	0.295	0.331	2.2	1.6	1.5	3.2
S16	95	95	90	90	120	0.180	0.208	1.8	1.5	1.5	3.4

Note: SII—Conv = SII estimations for the conventional hearing aid; CII—Comp. = SII estimations for the frequency compressed hearing aid.

The compression settings are reported as: default cutoff frequency (CF), compression ratio (CR), Fitting assistant cutoff frequency, and Fitting assistant compression ratio.

Table 3.3. Change in SII value and sound field thresholds following fitting of the frequency-compression hearing aid with reference to the conventional hearing aid for subjects in experiment one.

Subject ID	Improvement in SII	Amount	Improvement in SF Thresholds	Change in SF Threshold per Frequency (in dB)				
				2000	3000	4000	6000	8000
S2	X	.001	DNT					
S3	✓	.029	✓	0	5	20	10	0
S4	✓	.005	✓	0	5	5	5	25
S5	✓	.041	DNT					
S6	✓	.032	✓	-10	0	5	15	20
S7	X	0	✓	5	10	15	0	0
S8	✓	.027	✓	0	0	5	25	15
S10	X	0	✓	-5	10	25	30	5
S15	✓	.036	✓	0	5	10	0	0
S16	✓	.028	✓	0	20	35	30	10

Note: ✓ = An improvement in SII or sound field threshold for the frequency-compression hearing aid compared to the conventional hearing aid; X = no change in SII or sound field threshold for the frequency-compression hearing aid; SF = sound field; DNT = data was not obtained for this condition or individual.

Table 3.4. Listening conditions evaluated for each test in experiment one.

	Listening Condition				
Tests	CI+HA-conventional	CI+HA-compressed	CI only	HA only-conventional	HA only-compressed
Localization	✓	✓			
Spondee-in-noise	✓	✓	✓	✓	✓
Spatial hearing questionnaire	✓	✓			
Sound quality questionnaire	✓	✓			

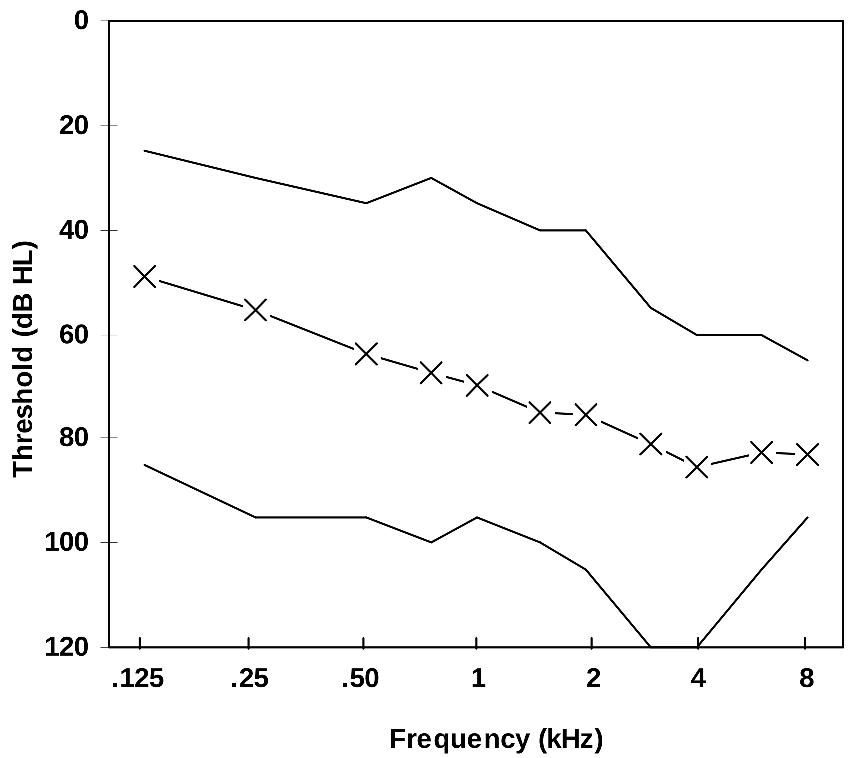


Figure 3.5. Unaided audiometric thresholds for participants in experiment two for the test ear. Frequency is shown along the x-axis from 0.125 to 8.0 kHz and threshold is shown (in dB HL) along the y-axis. Average audiometric thresholds are indicated by the x's and the range of thresholds are shown by the upper and lower lines.

Table 3.5. Demographic and device use information for subjects in experiment two.

Subject Number	Gender	Age (yrs)	Age of onset (yrs)	Etiology	CI Type	Previous HA	Length of HA use (yrs)
S2	M	75	57	Ototoxicity, noise	1	ITE	25.0
S3	F	41	5	Unknown	2	ITE	34.0
S4	F	71	14	Hereditary	1	ITE	29.0
S5	M	58	18	Noise, Hereditary	3	BTE	11.0
S6	F	49	11	Unknown	2	ITE	37.0
S7	M	72	55	Infection, noise	2	ITE	17.0
S8	F	60	12	Unknown	4	BTE	46.0
S9	F	41	17	Hereditary	None	BTE	19.0
S10	M	39	16	Unknown	5	BTE	15.0
S11	M	57	54	Unknown	None	ITE	3.0
S12	M	64	62	Noise	None	BTE	1.5
S14	M	72	48	Unknown	None	ITE	10.0
S15	F	53	24	Unknown	5	BTE	29.0
S16	M	82	35	Meniere's disease	1	BTE	50.0
S18	M	68	58	Noise	None	ITE	6.0
S19	F	77	74	Hereditary	None	BTE	2.0
S20	M	72	33	Unknown	None	ITE	5.0
Average		61.8	34.9				20.0

Note: Age of onset refers to the age at which subject first noticed hearing loss, Etiology = cause of hearing loss; Noise = noise-induced hearing loss; CI Type, 1 = Advanced Bionics 90K, 2 = Nucleus CI24RE, 3 = Nucleus CI24R, 4 = Nucleus RP8, 5 = Nucleus CI512; Previous HA = type of hearing aid used by the subject prior to the study; ITE = in-the-ear hearing aid; BTE = behind-the-ear hearing aid; Length of HA use = length in years of hearing aid use prior to onset of study.

Table 3.6. Unaided audiometric thresholds, SII calculations, and compression settings for subjects in experiment two.

	Unaided thresholds (in dB HL)					SII-Conv	SII-Comp.	Default CF (Hz)	Default CR	Fitting assistant CF (Hz)	Fitting assistant CR
	500	1000	1500	2000	4000						
S2	55	85	90	105	105	0.246	0.247	2.1	1.5	1.5	4.1
S3	65	75	80	80	90	0.379	0.408	2.8	1.9	1.5	1.6
S4	75	70	75	70	60	0.465	0.470	3.9	2.5	4.8	2.6
S5	85	80	75	70	75	0.373	0.414	3.4	2.2	1.6	1.7
S6	90	90	90	90	90	0.228	0.260	2.1	1.6	1.5	3.4
S7	60	80	90	95	110	0.273	0.273	2.1	1.5	1.5	4.1
S8	85	85	80	80	85	0.213	0.240	2.7	1.8	1.5	2.6
S9	55	55	70	65	80	0.459	0.503	3.5	2.3	1.6	1.8
S10	75	95	100	100	120	0.229	0.229	2.0	1.5	1.5	4.1
S11	50	50	65	65	70	0.554	0.572	4.5	2.7	2.6	2.4
S12	40	60	80	80	90	0.422	0.482	2.8	1.9	1.5	4.1
S14	35	45	45	50	70	0.659	0.746	4.5	2.8	1.7	2.1
S15	80	95	95	90	90	0.295	0.331	2.2	1.6	1.5	3.2
S16	95	95	90	90	120	0.180	0.208	1.8	1.5	1.5	3.4
S18	50	45	65	65	70	0.579	0.620	4.4	2.7	1.7	2.0
S19	45	50	45	40	60	0.728	0.777	4.5	2.7	3.7	2.6
S20	45	35	40	45	65	0.725	0.787	4.4	2.7	3.0	2.4

Note: SII-Conv = SII estimations for the conventional hearing aid; SII-Comp. = SII estimations for the frequency-compression hearing aid.

The compression settings are reported as: default cutoff frequency (CF), compression ratio (CR), Fitting assistant cutoff frequency, and Fitting assistant compression ratio.

Table 3.7. Change in SII value and sound field thresholds following fitting of the frequency-compression hearing aid with reference to the conventional hearing aid for subjects in experiment two.

Subject ID	Improvement in SII	Amount	Improvement in SF Thresholds	Change in SF Threshold per Frequency (in dB)				
				2000	3000	4000	6000	8000
S2	X	.001	DNT					
S3	✓	.029	✓	0	5	20	10	0
S4	✓	.005	✓	0	5	5	5	25
S5	✓	.041	DNT					
S6	✓	.032	✓	-10	0	5	15	20
S7	X	0	✓	5	10	15	0	0
S8	✓	.027	✓	0	0	5	25	15
S9	✓	.044	✓	0	5	10	20	20
S10	X	0	✓	-5	10	25	30	5
S11	✓	.018	✓	-5	0	0	5	10
S12	✓	.060	✓	-10	5	10	5	0
S14	✓	.087	✓	5	10	10	15	15
S15	✓	.036	✓	0	5	10	0	0
S16	✓	.028	✓	0	20	35	30	10
S18	✓	.041	✓	-5	-5	10	20	15
S19	✓	.049	✓	0	-5	0	0	20
S20	✓	.062	✓	0	5	15	15	15

Note: ✓ = An improvement in SII or sound field threshold for the frequency-compression hearing aid compared to the conventional hearing aid; X = no change in SII or sound field threshold for the frequency-compression hearing aid; SF = sound field; DNT = data was not obtained for this condition or individual.

Table 3.8. Listening conditions evaluated for each test in experiment two.

	Listening Condition				
Tests	CI+HA-conventional	CI+HA-compressed	CI only	HA only-conventional	HA only-compressed
Consonant				✓	✓
Vowels				✓	✓

CHAPTER 4

RESULTS

4.1 Experiment 1: Contribution of Frequency Compression to CI+HA Performance

The primary goal of experiment one was to determine the contribution of a frequency-compression hearing aid to contralateral CI performance. Ten adult subjects were fit with a Phonak Naida hearing aid contralateral to the CI; the subjects were instructed to alternate daily for two months between conventional and frequency-compression hearing aid programs. Performance was evaluated using the CI+HA on tests of sound localization and spondee-in-noise with spatially separated target and noise sources. Two questionnaires were also administered at the two month interval to document self-report outcomes following the home trial. Subject compliance in alternating between the two hearing aid programs was tracked using a data logging feature on the hearing aid and will be presented first.

4.1.1 Subject Compliance during the Home Trial

Subjects were compliant in alternating between conventional and frequency-compression hearing aid programs as data logging showed an average of 47.6% use of the conventional program and 52.4% use of the frequency compression program during the two month interval. Data logging from subject S6 could not be retrieved from the hearing aid and was not included in the calculation of average hearing aid use. However, the subject reportedly alternated between the two hearing programs during the at-home trial as instructed.

4.1.2 Localization Performance

Shown in Figures 4.1-4.3 and Figure D1 (individual scores) are results for the horizontal localization test. Figures 4.1 displays mean localization scores (shown in degrees RMS error) from baseline to two months for the CI+HA-conventional and CI+HA-compressed conditions. Mean localization scores for the CI+HA-conventional and CI+HA-compressed conditions at baseline were 37.1° (SE = 2.7) and 35.3° RMS error (SE = 2.6), and after two months were 31.6° (SE = 2.7) and 31.7° RMS error (SE = 2.7), respectively. The results of a paired *t*-test revealed no significant difference between the CI+HA-conventional and CI+HA-compressed conditions after two months, $t(9) = -.05$, $p = .964$ (two-tailed).

Localization scores were also analyzed to compare the rate of change in performance between the two conditions by computing the difference from baseline to two months. The mean change in performance for the CI+HA-conventional condition increased by 5.47° RMS error and for the CI+HA-compressed condition, by 3.59° RMS error. Results of a paired *t*-test indicated no significant difference in the rate of change in performance between the CI+HA-conventional and the CI+HA-compressed conditions, $t(9) = .91$, $p = .386$ (two-tailed).

Figure 4.2 shows results from the localization test comparing the change in performance from baseline to two months for each subject. The x-axis displays the change in performance for the CI+HA-conventional condition and the y-axis, the CI+HA-compressed condition. Scores marked in the upper right quadrant represent an improvement for both conditions from baseline to two months, whereas scores in the lower left quadrant represent a decrement in both conditions overtime. Localization scores for 7 out of 10 subjects are shown in the upper right quadrant of Figure 4.2., indicating a trend of positive change in performance for these subjects using the CI+HA-conventional and CI+HA-compressed conditions. The open circle shows the average change in performance between the two conditions. As noted above, this difference was not statistically significant, as summarized by the results from a paired two-tailed *t*-test, $t(9) = .91$, $p = .386$.

Shown in Figure 4.3 are the individual difference scores for each subject on the localization test. In Figure 4.3, scores along the y-axis were calculated by the finding difference in RMS error for the CI+HA-conventional and CI+HA-compressed conditions at the two month interval. Difference scores in degrees RMS error are displayed in an ascending order. Results revealed that, after two months use, 3 out of 10 subjects (S5, S6, S10) had a lower localization score with the CI+HA using conventional amplification by approximately 6° RMS error. In comparison, 2 of 10 subjects (S4, S16) had a lower localization score with the CI+HA using frequency compression by at least 6° RMS error. Data were further analyzed for each individual using a paired *t*-test by comparing the mean error squared between the two conditions (see Table 4.1). Only subject S5 showed a significant improvement with CI+HA-conventional condition, $t(96) = 1.93$, $p = .018$.

In sum, mean localization performance after two months listening experience was no different using CI+HA with or without frequency compression. Furthermore, the rate of performance change from baseline to two months was also not significantly different for the CI+HA-conventional and CI+HA-compressed conditions. However, a trend emerged that 7 of 10 subjects improved in localization ability for both conditions from baseline to two months. Individual results found that 1 of 10 subjects showed significantly better localization performance with conventional amplification compared to frequency compression.

4.1.3 Spondee-in-noise Performance

Results from the spondee-in-noise test are shown in Figures 4.4-4.7. Figure 4.4 displays CI+HA scores using the conventional and frequency-compression hearing aids at baseline and two months. Scores were reported by the signal-to-noise ratio (SNR, in dB) needed to obtain a threshold of 50% correct for the spondee words. A more negative score shown by a negative SNR indicates better performance on this task. After two months, spondee-in-noise performance for the CI+HA-conventional condition ($M = -6.68$, SE = 1.40) was significantly better, shown by a more negative SNR, than the CI+HA-compressed condition ($M = -4.33$, SE = 1.71), $t(9) = -3.48$, $p = .007$ (two-tailed).

Figure 4.5 displays the change in performance for the CI+HA-conventional and CI+HA-compressed conditions from baseline to two months for all subjects. This was calculated by computing the difference in each subject's score from baseline to two months for the two listening conditions. Scores marked in the upper right quadrant represent an improvement for both conditions overtime, whereas scores in the lower left quadrant represent a decrement overtime. Individual results revealed better scores on the spondee-in-noise task from baseline to two months for three individuals using the CI+HA-conventional condition and two individuals with the CI+HA-compressed condition. In addition, one of ten subjects obtained a lower SNR score for both conditions overtime and no subjects showed a decrement in performance from baseline to two months. Results were further analyzed by comparing the mean rate of change in SNR performance between the two conditions from baseline to two months. The mean differences for the CI+HA-conventional and the CI+HA-compressed conditions from two months to baseline, indicated as 'average' in Figure 4.5, were 1.71 and 0.14, respectively. The change in performance from baseline to two months was not significantly different between the two conditions, $t(9) = 1.60$, $p = .145$ (two-tailed).

Individual difference scores after two months using the CI+HA-conventional and CI+HA-compressed on the spondee-in-noise test are shown in Figure 4.6. Recall that more negative scores on the spondee-in-noise test indicate better performance on this task. As a result, more negative scores represent better performance using conventional amplification and positive scores represent better performance using frequency compression. The critical difference score at a 95% confidence level for the spondee-in-noise test was derived from pilot testing of four adult bilateral cochlear implant users (refer to section 3.1.4 for more details). Results showed that 5 of 10 subjects (S2, S5, S7, S15, S16) performed significantly better on the spondee-in-noise test, as shown by a more negative SNR score, after two months using the CI+HA-conventional condition compared to the CI+HA-compressed condition. Individual scores for the remaining five subjects were not significantly different between the two conditions.

Figure 4.7 shows results on the spondee-in-noise task for all five listening conditions, including the CI+HA with and without frequency compression, the CI only, and the hearing aid with and without frequency compression. Performance with the CI only at baseline was -3.39 SNR (SE = 1.59) and at two months, -4.00 SNR (SE = 1.32). Comparing the two hearing aid conditions after two months of use, scores averaged 0.25 SNR (SE = 2.03) for the HA-conventional condition and 2.82 SNR (SE = 2.02) for the HA-compressed condition. Statistical analysis using paired two-tailed *t*-tests showed several conditions to be significantly different after two months of listening experience. The comparisons showing significant differences were as follows: the CI+HA-conventional and CI only conditions, $t(9) = -3.49$, $p = .007$; the CI+HA-conventional and HA-conventional conditions, $t(9) = -5.04$, $p = .001$; the CI+HA-compressed and HA-compressed conditions, $t(9) = -4.39$, $p = .002$; and the CI only and HA-compressed conditions, $t(9) = -3.36$, $p = .008$. The comparison in spondee-in-noise performance at two months between the CI only and the HA-conventional conditions was not significant, $t(9) = -2.17$, $p = .058$ as well as the CI+HA-compressed and the CI only conditions, $t(9) = -.34$, $p = .742$. However, it is important to note that these results should be considered with caution due to multiple comparisons using the paired *t*-tests.

Individual outcomes from the spondee-in-noise test are shown in Appendix D, Figures D2-4. Figure D2 shows results for each subject comparing the CI+HA-conventional and CI+HA-compressed conditions at baseline and two months. Figure D3 displays individual results for the conventional hearing aid and CI (CI+HA-conventional, CI only, HA-conventional) after two months, whereas Figure D4 shows results for the frequency compression hearing aid and CI (CI+HA-compressed, CI only, HA-compressed).

Overall, performance on the spondee-in-noise task using the CI+HA was significantly better after two months for conventional amplification compared to frequency compression. Additional analyses indicated that the type of device use, CI+HA, CI only, and hearing aid only, had a significant effect on spondee-in-noise performance. Performance on the spondee-in-noise test was significantly better with the CI+HA than the hearing aid only with and without frequency compression.

4.1.4 Questionnaire Results

Results from the Spatial Hearing Questionnaire and Sound Quality Questionnaire are shown in Figures 4.8 and 4.9. For the Spatial Hearing Questionnaire, mean scores for the CI+HA-conventional and CI+HA-compressed conditions after two months of hearing aid use are displayed along the x-axis for the eight subscales and the total score. Mean results represent scores averaged from nine subjects. Data was omitted from subject S2 because responses were only provided for one of the two listening conditions. Higher ratings on the Spatial Hearing Questionnaire indicated better spatial hearing ability, where the maximum possible rating was 100. For the CI+HA-conventional condition, scores were rated highest for the speech in quiet subscale (M = 80.7, SE = 8.4) and lowest for the localization subscale (M = 53.3, SE = 4.8). For the CI+HA-compressed condition, self-reported ability was also rated highest for the speech in quiet subscale (M = 74.1, SE = 6.5) and lowest for the music subscale (M = 51.1, SE = 5.8).

The mean total Spatial Hearing Questionnaire, calculated by an average of all 24 items, was 59.8 (SE = 5.5) and 56.2 (SE = 5.2) for the CI+HA-conventional and CI+HA-compressed conditions, respectively. Differences between the two listening conditions after two months for the total score and all eight subscales were not statistically significant (two-tailed): Total score, $t(8) = -1.09$, $p = .306$; Male voices, $t(8) = -.61$, $p = .56$; Female voices, $t(8) = -.476$, $p = .647$; Children’s voices, $t(8) = -1.11$, $p = .30$; Music, $t(8) = -1.60$, $p = .149$; Localization, $t(8) = -.59$, $p = .57$; Speech in quiet, $t(8) = -1.04$, $p =$

.329; Speech in noise from the front, *t*(8) = -.83, *p* = .432; Speech in noise with spatially separate target and noise sources, *t*(8) = -1.05, *p* = .326.

The Spatial Hearing Questionnaire total scores for each subject comparing the CI+HA-conventional and CI+HA-compressed conditions are displayed in the Appendix D, Figure D5. Scores varied considerably across subjects, from 84.6 (subject S10) to 28.3 (subject S6) for the CI+HA-conventional condition.

Outcomes from the Sound Quality Questionnaire, comparing the CI+HA-conventional and CI+HA-compressed conditions after two months of use, are shown in Figure 4.9. Sound quality ratings were averaged across eight subjects because subjects S2 and S3 did not complete the questionnaire as this was implemented later in the test protocol. Higher subjective ratings indicated better performance for that listening condition, where 100 was the maximum possible rating. Subjective ratings were significantly higher for the item concerning speech understanding for the CI+HA-conventional condition (*M* = 76.25, SE = 9.76) compared to the CI+HA-compressed condition (*M* = 63.75, SE = 10.76), *t*(7) = 2.55, *p* = .038. For all other items on the sound quality ratings questionnaire, there were no significant differences between the two listening conditions: Quality of sound, *t*(7) = 1.90, *p* = .099; Recognizing everyday sound, *t*(7) = 1.37, *p* = .213; Noise in program, *t*(7) = -.68, *p* = .519; Appreciation of music, *t*(7) = 1.67, *p* = .138; Quality of own voice, *t*(7) = 1.51, *p* = .174.

Results of the two questionnaires, the Spatial Hearing Questionnaire and the Sound Quality Ratings, revealed essentially no significant effect between the CI+HA with or without frequency compression after two months use. However, the mean subjective rating for speech understanding from the Sound Quality Ratings questionnaire were significantly higher for conventional amplification compared to frequency compression.

4.2 Experiment 2: Impact of Frequency Compression on Speech Perception

The general goal of experiment two was to determine the impact of frequency-compression on speech perception in quiet. Here, seventeen adults served as participants including the ten CI+HA users from experiment one and an additional seven HA+HA users. As in experiment one, subjects were fit with a Phonak Naida hearing aid and alternated daily for two months between the conventional and frequency-compression hearing aid programs. Subject compliance in alternating between the two hearing aid programs was tracked using a data logging feature on the hearing aid. Consonant and vowel recognition was tested monaurally at baseline and after two months of use. All consonant and vowel stimuli were presented in quiet from a single loudspeaker at 0° azimuth.

4.2.1 Subject Compliance during the Home Trial

As with experiment one, subjects in experiment two were also compliant in alternating daily between conventional and frequency-compression hearing aid programs. Data logging indicated that the conventional hearing aid program was used an average of 52.1% of the time and the frequency-compression hearing aid program was used 47.9% during the two month interval. Data from subjects S6 and S12 could not be retrieved from the hearing aid and was not included in the calculation. However, both of these subjects reported using the two hearing aid programs equally during the at-home trial.

4.2.2 Consonant Recognition Performance

Figure 4.10 shows mean consonant recognition scores, shown in percent correct, for the HA-conventional and HA-compressed conditions at baseline and two months. Mean consonant recognition scores were calculated by averaging the results from two lists consisting of a male and a female talker. After two months of listening experience, mean consonant scores were 51.0% (SE = 5.4) and 50.3% (SE = 5.2) for the HA-conventional and HA-compressed conditions. Results of a paired *t*-test indicated that there was no significant difference between the HA-conventional and HA-compressed conditions at two months, $t(16) = 1.72$, $p = .676$.

Consonant recognition scores were further analyzed to compare the rate of change in performance between the HA-conventional and HA-compressed conditions from baseline to two months. Results for all seventeen subjects as well as the mean change in performance are shown in Figure 4.11. The x-axis displays the change in performance for the HA-conventional condition and the y-axis shows the performance change for the HA-compressed condition. For the HA-compressed condition, an improvement in performance of $\geq$5% from baseline to two months was found for seven subjects whereas for the HA-conventional condition, an improvement of $\geq$5% overtime was seen for four subjects. In comparison, one subject showed a decrement in performance of $\geq$5% overtime for the HA-compressed condition, and five subjects for the HA-conventional condition.

As shown in Figure 4.11, mean scores from baseline to two months decreased by 0.22% (SE = 1.42) for the HA-conventional condition and increased by 2.45% (SE = 1.20) for the HA-compressed condition. Although scores improved for the HA-compressed condition overtime and decreased only slightly for the HA-conventional condition, results of a paired *t*-test indicated there was no significant effect in rate of performance change between the HA-conventional and the HA-compressed conditions, $t(16) = -2.02$, $p = .060$ (two-tailed).

Individual scores on the consonant recognition test are displayed in Figure 4.12, comparing individual performance at two months between the conventional and frequency-compression hearing aids. The y-axis displays the mean difference in performance between the conventional and frequency-compression hearing aid at two months. Critical difference scores at a 95% confidence level (Carney and Schlauch, 2007) were used to determine if scores were significantly different between the two conditions. As shown in Figure 4.12, results revealed that two individuals (S6, S14) demonstrated a significant improvement in consonant recognition using the conventional hearing aid compared to the frequency-compression hearing aid.

Additional individual results from the consonant recognition test are shown in Appendix D, Figure D6. Results are displayed for all seventeen subjects showing absolute scores at baseline and two months for the HA-conventional and HA-compressed conditions. After two months use, individual consonant recognition scores for the HA-conventional condition ranged from 4.5% for subject S16 to 85.3% for subject S9. For the HA-compressed condition, individual scores ranged from 7.7% for subject S16 to 87.2% for subject S9.

Despite some individual differences, overall results on the consonant recognition test revealed no significant difference in performance at two months with and without frequency compression. Also, mean results showed no significant difference in the rate of change overtime between the HA-conventional and HA-compressed conditions.

4.2.3 Vowel Recognition Performance

Figure 4.13 displays mean vowel recognition scores, shown in percent correct, for the HA-conventional and HA-compressed conditions. Scores for subject S2 at baseline were not obtained due to time constraints while testing and, therefore, not included in the average calculation at baseline. After two months of listening experience, vowel recognition scores averaged 71.4% (SE = 6.7) for the HA-conventional and 65.8% (SE = 7.1) for the HA-compressed conditions. This difference was significant, $t(16) = -3.16$, $p = .006$ (two-tailed), indicating that performance was significantly better with the HA-conventional condition compared to the HA-compressed condition after two months of use.

The rate of performance change for vowel recognition from baseline to two months between the two conditions is shown in Figure 4.14. Individual results are shown for 16 subjects (excluding subject S2) as well as the mean change in performance, marked by the open circle and labeled ‘Average’. The x-axis displays the change in performance for the HA-conventional condition and the y-axis shows the change in performance for the HA-compressed condition. For 9 of 16 subjects, results revealed a trend of positive change in performance using the HA-conventional and HA-compressed conditions, as evidenced by scores in the upper right quadrant of Figure 4.14. There were two subjects that showed a decrement in performance for both conditions from baseline to two months of approximately 10%.

The mean change in performance from baseline to two months was found to increase by 6.7% for the HA-conventional condition and 4.9% for the HA-compressed condition. The overall rate of change in vowel recognition performance between the two conditions overtime was not found to be significantly different, $t(15) = .52$, $p = .611$ (two-tailed).

Figure 4.15 shows the individual difference scores using the conventional and frequency-compression hearing aid after two months for the vowel recognition test. The critical difference scores were calculated based on a 95% confidence interval and statistical significance was defined as $p < .05$ (Carney and Schlauch, 2007). The critical differences for percent correct were derived from a list length of n = 50 items, representing the 54 items used in this vowel perception test. In Figure 4.15, individual scores are arranged in an ascending order. For subjects S3, S6, S16, S18, and S20, there was no difference in vowel perception (0% difference in scores) between the HA-conventional and HA-compressed conditions. Only two subjects (S10, S15) showed a significant improvement with the conventional hearing aid compared to the frequency-compression hearing aid. There were no subjects that showed a significant improvement using the frequency-compression hearing aid compared to conventional amplification after two months of listening experience.

Figure D7 (see Appendix D) shows the results for each subject on the vowel recognition test at baseline and two months. Data for subject S2 are displayed for the two month interval only. As seen in the figure, there were eight subjects (S3, S9, S11, S14, S15, S18, S19, S20) that scored above 80% on the vowel recognition test for at least one condition.

In sum, vowel recognition scores were significantly higher (by approximately 6%) using conventional amplification compared to frequency compression after two months of listening experience. There was a trend suggesting improved performance overtime on the vowel perception task for both listening conditions, but the overall rate of change between the two conditions was not significantly different.

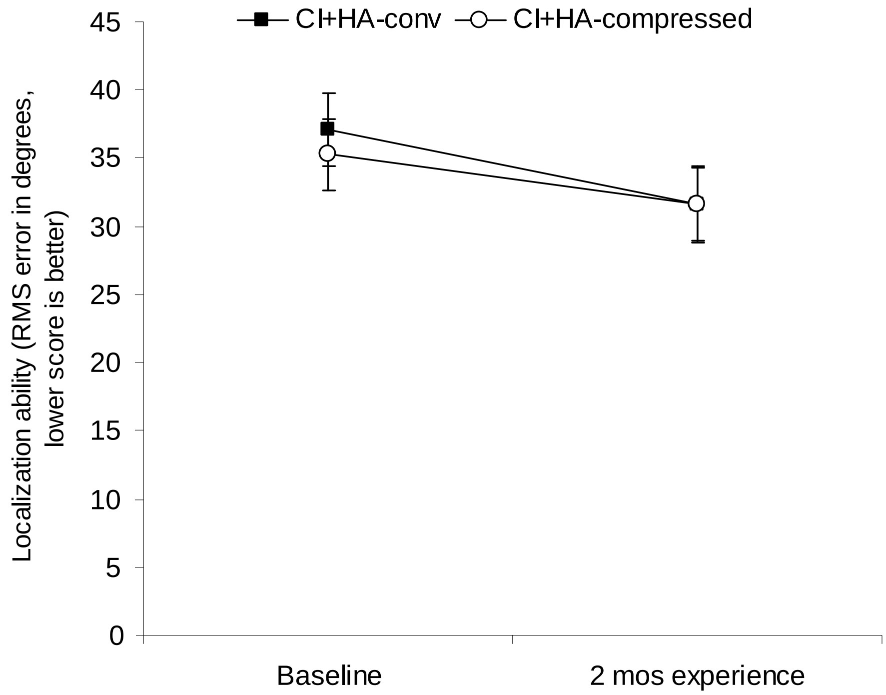


Figure 4.1. Mean results on the localization test for all participants. The y-axis shows the localization score in degrees RMS error. Lower scores on the graph indicate better localization ability. The CI+HA-conventional condition is shown by the filled squares and the CI+HA-compressed in the open circles.

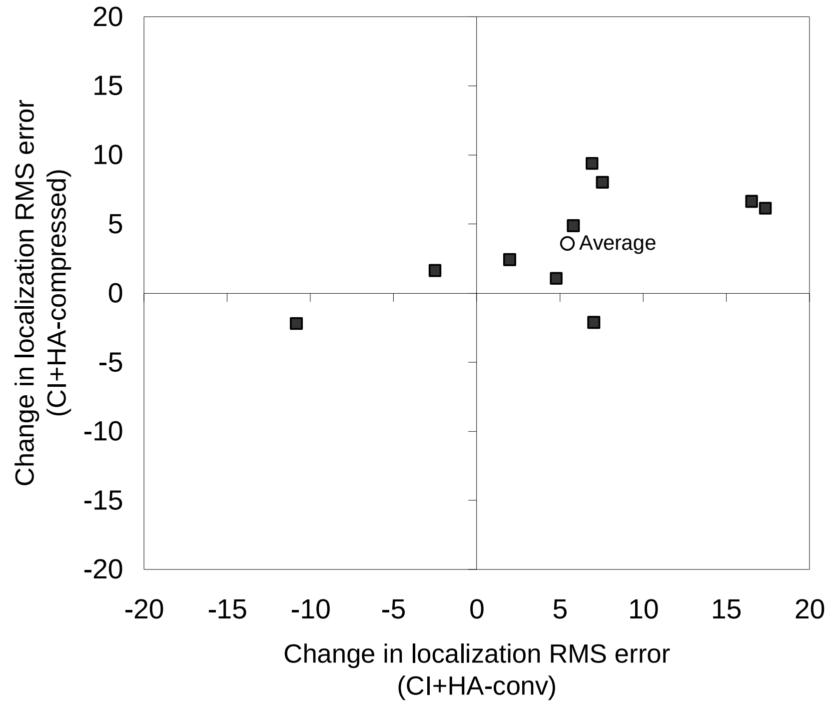


Figure 4.2 Change in localization performance using the CI+HA-conventional and CI+HA-compressed conditions from baseline to two months for all subjects. Scores with the CI+HA-conventional hearing aid condition are shown along the x-axis and the CI+HA-compressed hearing aid along y-axis. Scores are reported as the difference between localization ability in degrees RMS error from baseline to two months. Each filled square indicates the change in performance for a given subject and the mean is shown in the open circle, labeled ‘Average’.

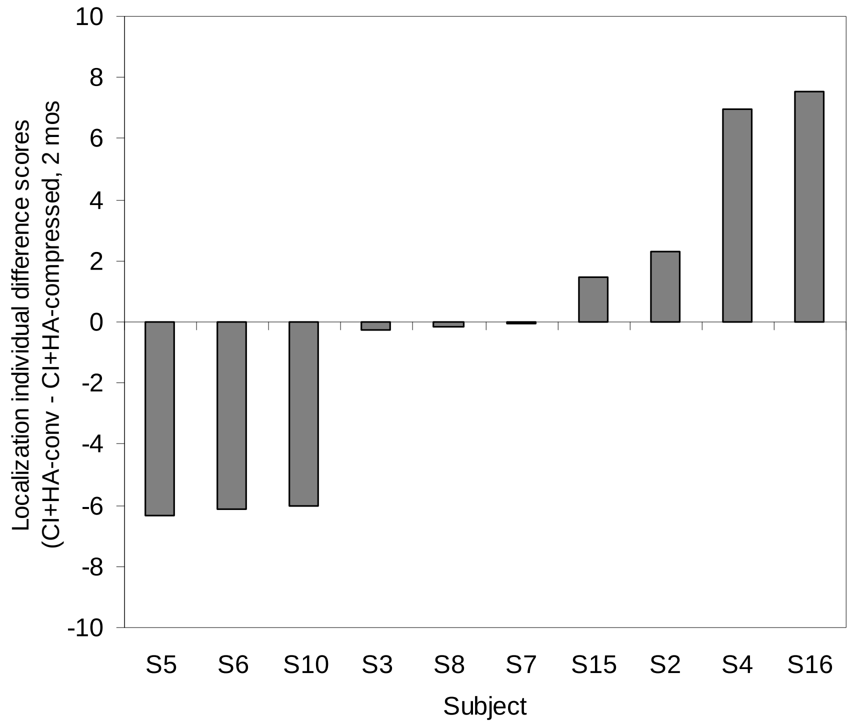


Figure 4.3. Individual difference scores between the CI+HA-conventional and CI+HA-compressed conditions at two months on the localization test. Difference scores in º RMS error are shown in ascending order.

Table 4.1. Mean squared error for each individual on the localization test comparing the CI+HA-conventional to CI+HA-compressed conditions.

	CI+HA-conventional		CI+HA-compressed			
Subject	*M*	SE	*M*	SE	*t*(95)	*p*
S2	1320.2	205.5	1490.6	219.9	-.61	.544
S3	348.8	51.8	358.8	69.0	-.143	.886
S4	1090.5	187.8	834.6	131.0	1.070	.287
S5	445.7	56.9	749.6	110.6	-2.408	**.018**
S6	1692.5	229.4	1227.9	194.5	1.495	.138
S7	1891.3	249.0	1896.8	277.4	-.015	.988
S8	969.0	142.4	978.9	156.5	-.052	.958
S10	874.7	125.9	1266.2	188.4	-1.858	.066
S15	735.1	99.7	659.8	98.8	.585	.560
S16	1850.3	245.9	1262.1	179.2	1.933	.056

Note: Lower *M* values indicate fewer errors on the localization task, which was calculated by the mean squared error from the stimulus-response values averaged across all eight loudspeakers in º.

Significant *p* values are shown in bold.

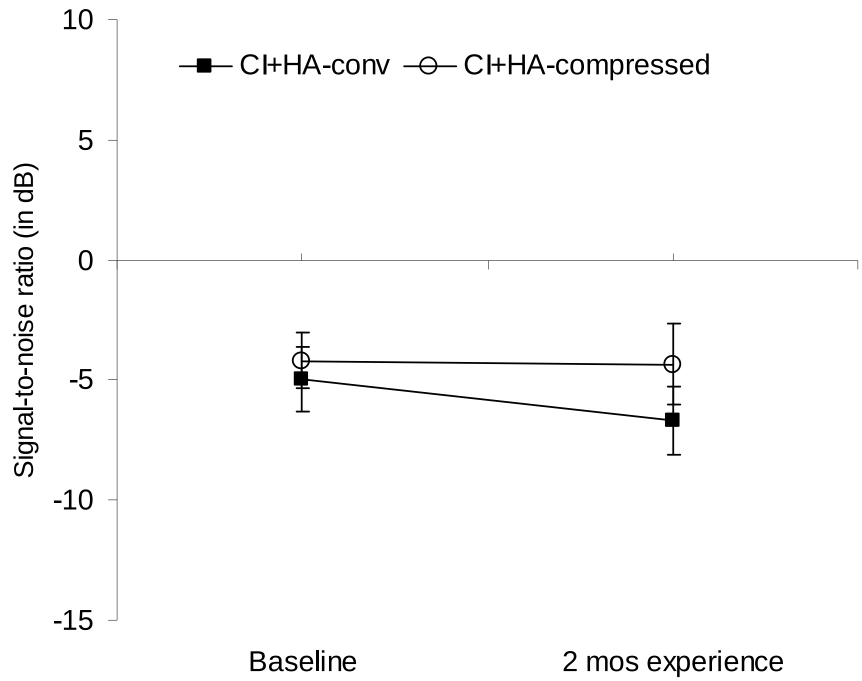


Figure 4.4. Performance on the spondee-in-noise test for all subjects. Scores are shown along the y-axis as the signal-to-noise ratio (SNR, in dB) required to obtain a 50% threshold for the spondee words. More negative scores indicate better performance on the task. Error bars represent standard errors as calculated based on the mean of two of three runs. The CI+HA-conventional is shown by the filled squares and the CI+HA-compressed in the open circles.

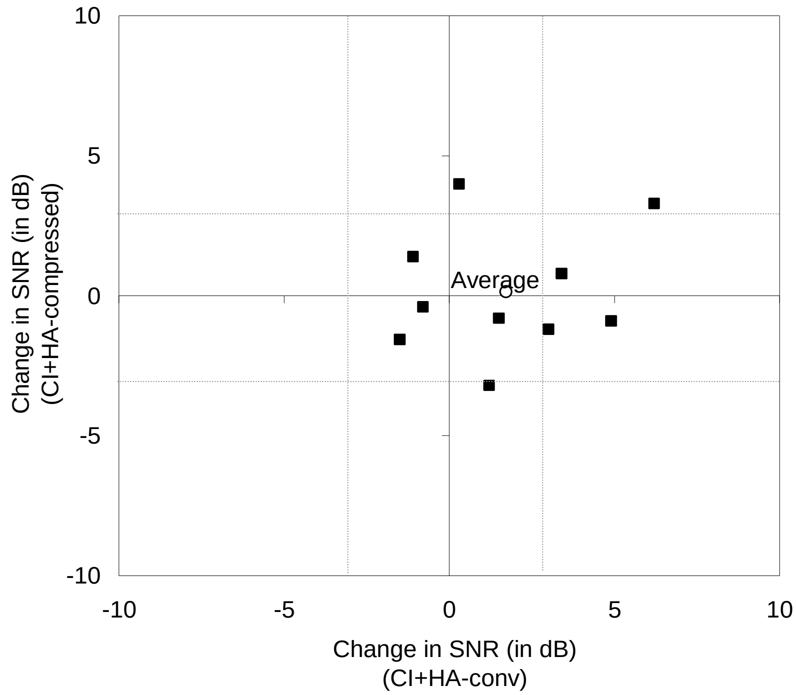


Figure 4.5. Change in spondee-in-noise performance using the CI+HA-conventional and CI+HA-compressed conditions from baseline to two months for all subjects. Scores with the CI+HA-conventional hearing aid condition are shown along the x-axis and the CI+HA-compressed hearing aid along y-axis. Scores are reported as the difference between signal-to-noise ratio (SNR) from baseline – 2 months. Each filled square indicates the change in performance for a given subject and the mean is shown in the open circle, labeled ‘Average’. The dashed lines indicate the 95% confidence intervals for each listening condition.

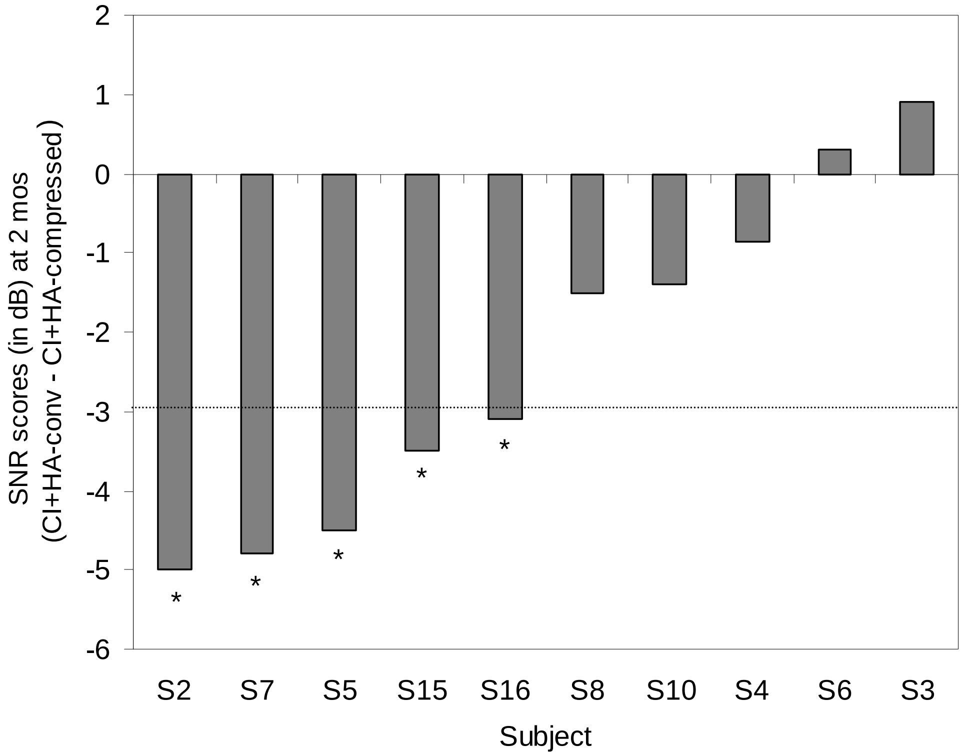


Figure 4.6. Individual difference scores for the CI+HA-conventional and CI+HA-compressed conditions at two months on the spondee-in-noise test. Individual scores are displayed the difference in signal-to-noise ratio (SNR, in dB), in ascending order. Higher scores indicate better performance for the CI+HA-compressed condition and negative scores indicate better performance for the CI+HA-conventional condition. The dashed line marks the 95% confidence interval. Significant differences based on a 95% confidence interval are marked by the asterisks. Note: *$p<0.05$.

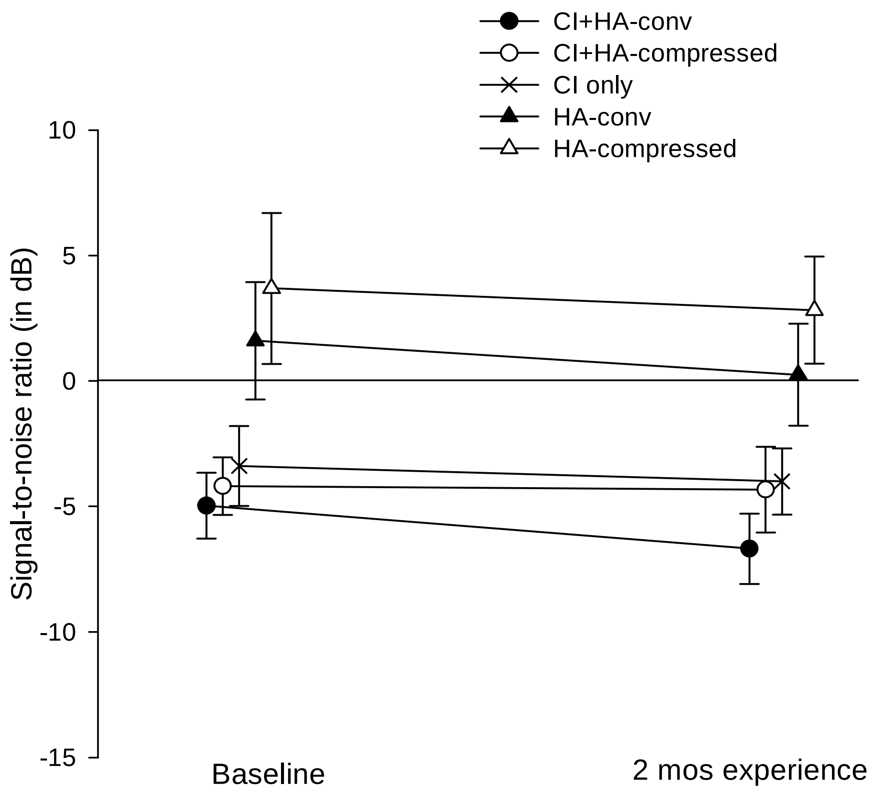


Figure 4.7. Mean performance on the spondee-in-noise test for all listening conditions: CI+HA-conventional (filled circle); CI+HA-compressed (open circle); CI only (x); HA-conventional (filled triangle); HA-compressed (open triangle). Scores for each condition are shown along the y-axis as the signal-to-noise ratio (dB) required to obtain a 50% threshold for the spondee words. More negative scores indicate better performance on the task. Error bars represent standard errors as calculated based on the mean of two of three runs.

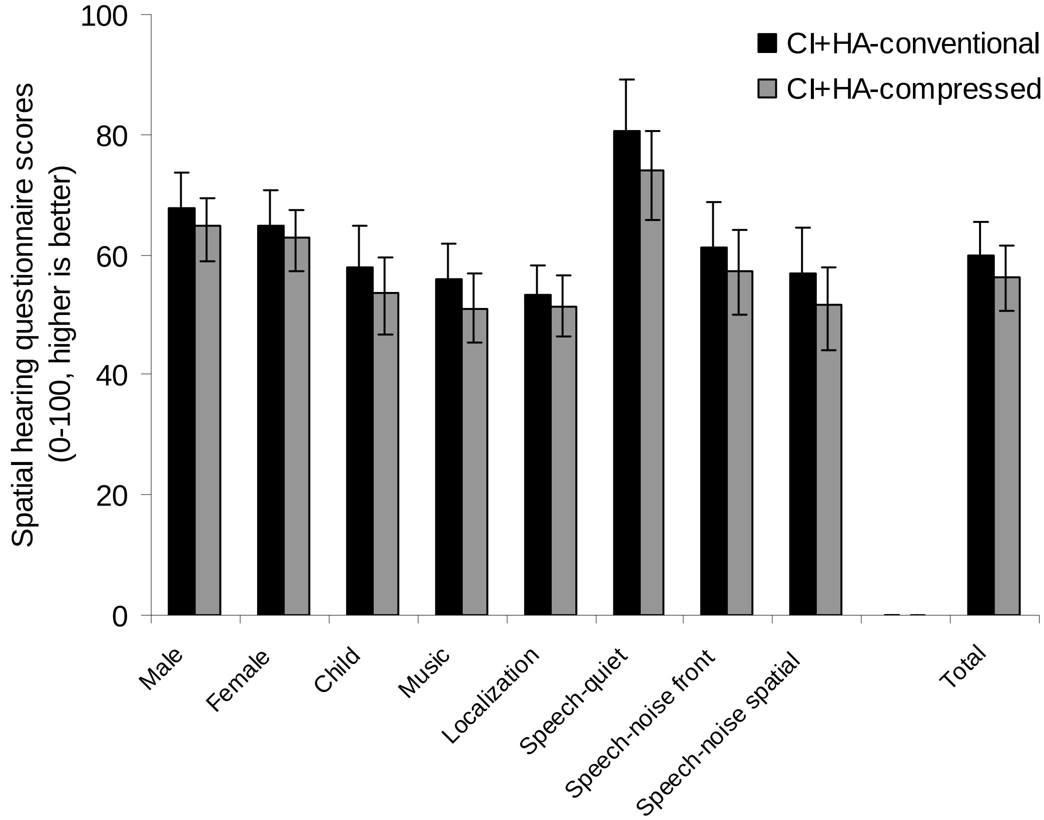


Figure 4.8. Average results on the Spatial Hearing Questionnaire for the CI+HA-conventional and the CI+HA-compressed conditions. The black columns show data for the CI+HA-conventional condition and the gray columns for the CI+HA-compressed condition. Scores are reported for the eight subscales and total score along the x-axis. Higher scores indicate better subjective spatial hearing ability. Error bars represent standard errors.

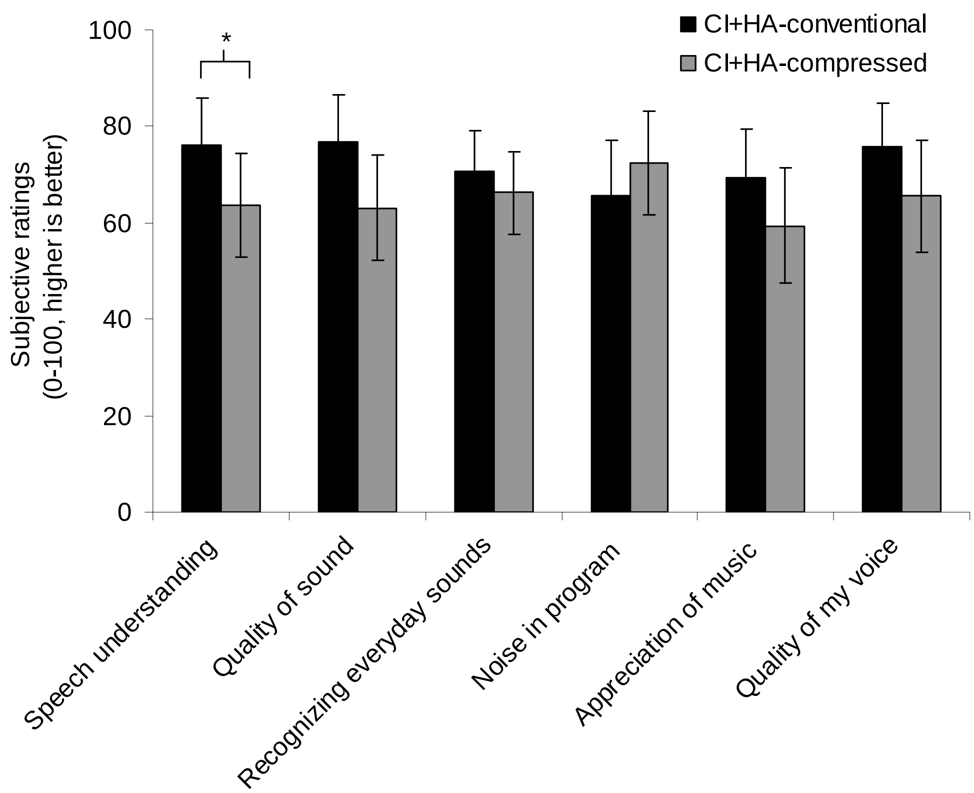


Figure 4.9. Mean sound quality ratings for the CI+HA-conventional and CI+HA-compressed conditions. The black columns show data for the CI+HA-conventional condition and the gray columns for the CI+HA-compressed condition. Higher scores represent higher subjective ratings. Error bars represent standard errors. Significant differences are marked by the brackets. Note: *p<0.05.

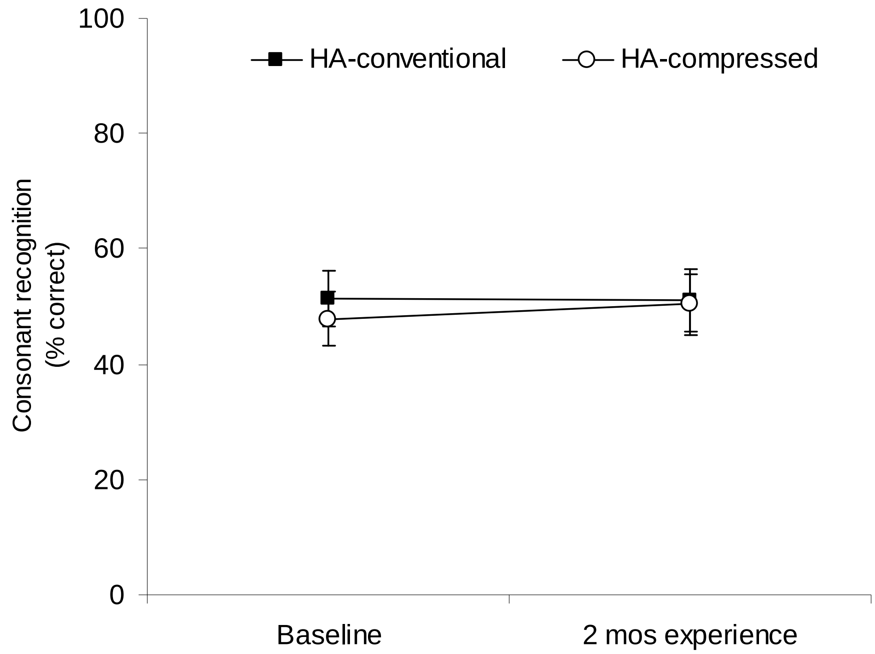


Figure 4.10. Mean consonant recognition scores at baseline and two months. Scores are shown in percent correct along the y-axis. Error bar represent standard errors as calculated based on the average of two lists of a male and female talker. The HA-conventional is shown by the filled squares and the HA-compressed in the open circles.

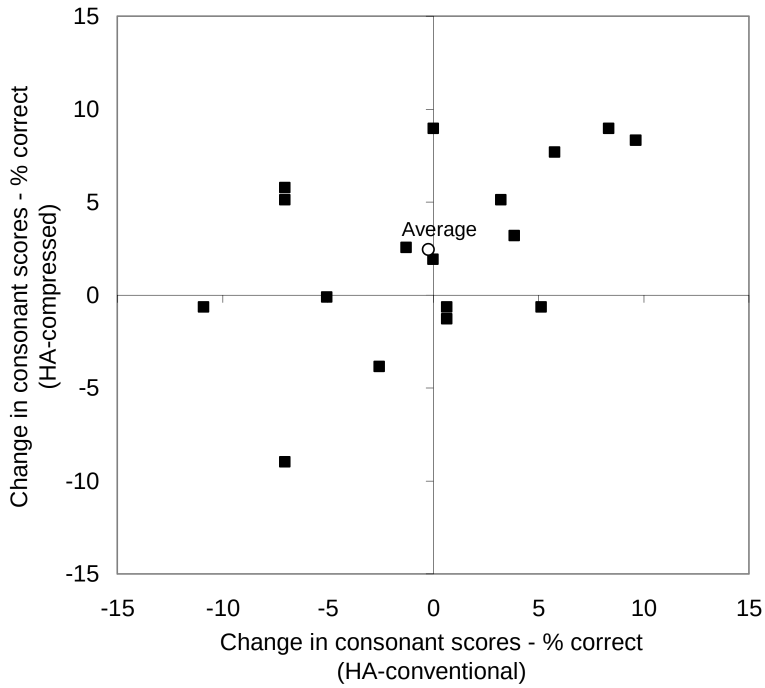


Figure 4.11. Change in consonant recognition scores using the HA-conventional and HA-compressed conditions from baseline to two months for all subjects. Scores for the HA-conventional hearing aid condition are shown along the x-axis and the HA-compressed hearing aid along the y-axis. Scores were calculated by taking the difference at two months minus baseline for each condition. Each filled square indicates the change in performance for a given subject and the mean is shown in the open circle, labeled ‘Average’.

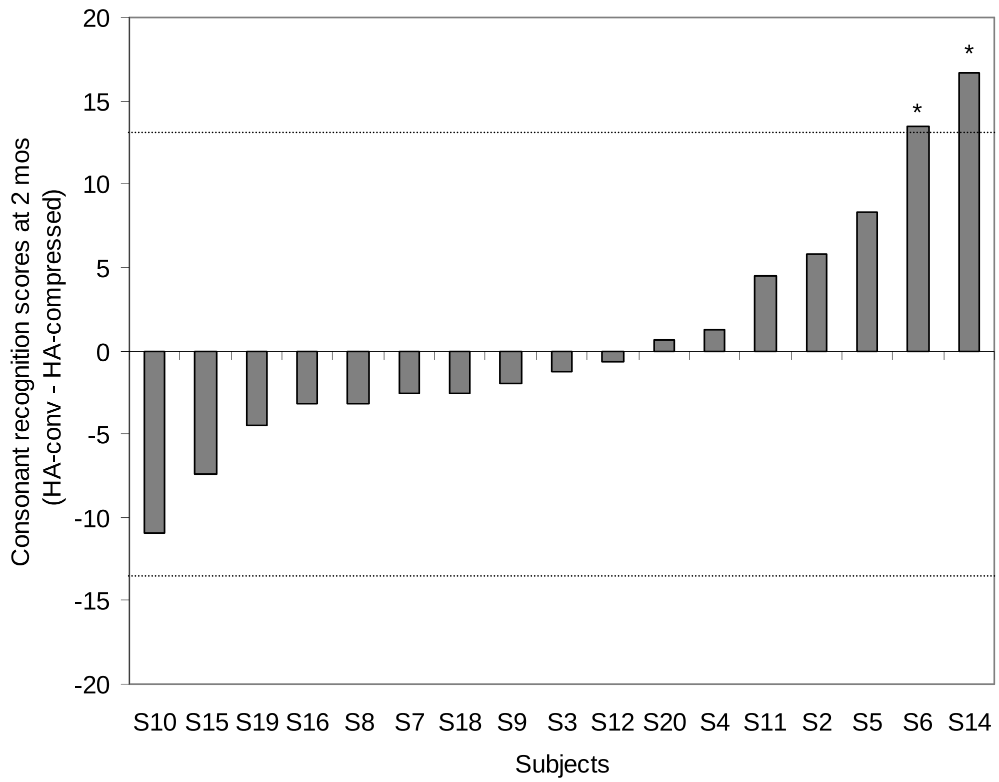


Figure 4.12. Individual difference scores between the HA-conventional and HA-compressed conditions at two months on the consonant recognition test. Individual scores are shown in ascending order. The dashed lines indicate the 95% confidence intervals (Carney and Schlauch, 2007). Significant differences based on a 95% confidence interval are marked by the asterisks. Note: **p*<0.05.

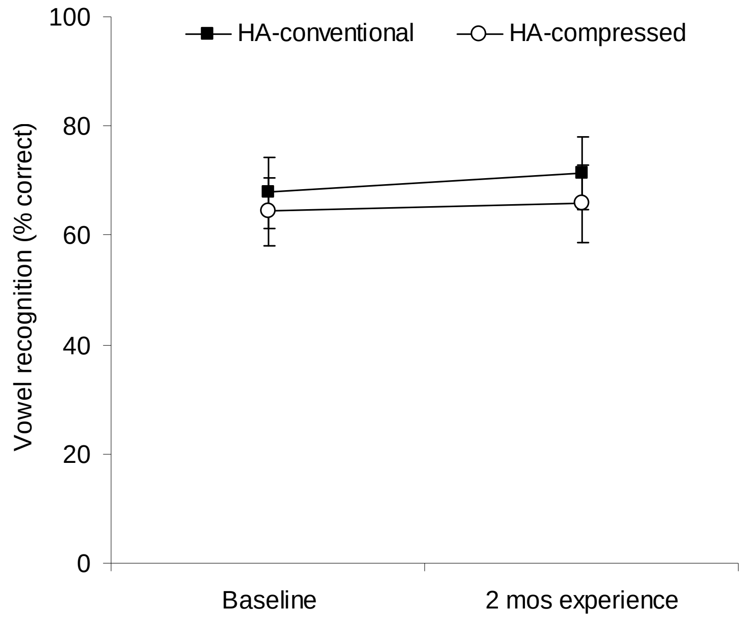


Figure 4.13. Mean vowel recognition scores at baseline and two months. Scores are shown in percent correct (along y-axis). Error bars represent standard errors. The HA-conventional scores are shown by the filled squares and the HA-compressed scores are shown in the open circles.

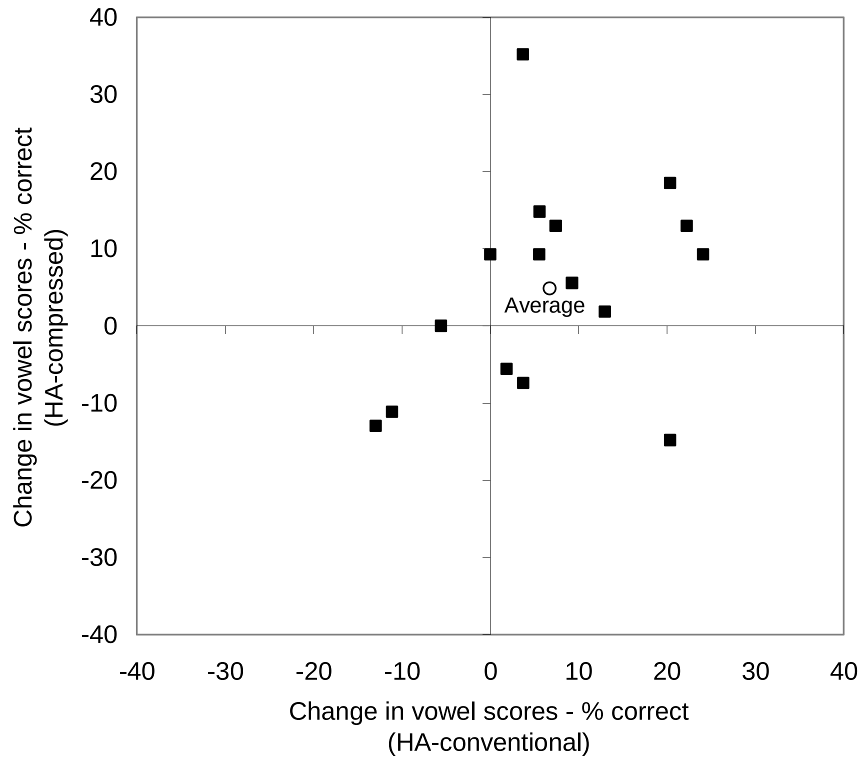


Figure 4.14. Change in vowel recognition scores using the HA-conventional and HA-compressed conditions from baseline to two months. HA-conventional hearing aid scores are shown along the x-axis and HA-compressed hearing aid scores along the y-axis. Scores were calculated by taking the difference at two months minus baseline for each condition. Each filled square indicates the change in performance for a given subject and the mean is shown in the open circle, labeled ‘Average’.

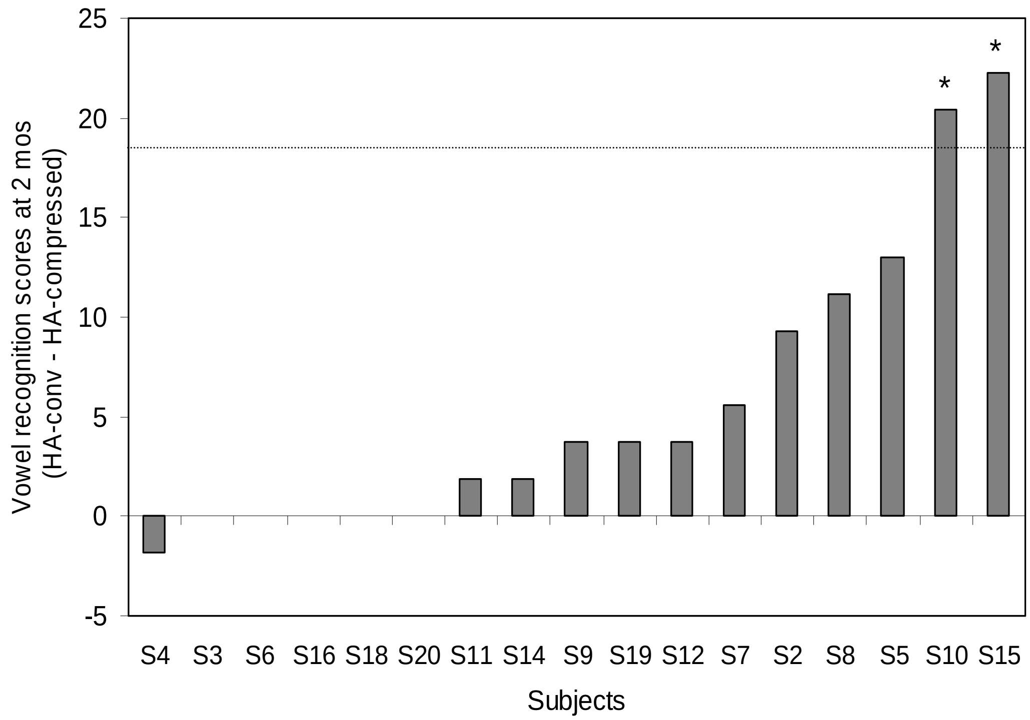


Figure 4.15. Individual difference scores between the HA-conventional and HA-compressed conditions at two months for the vowel recognition test. Individual scores are shown in ascending order. The dashed line indicates the 95% confidence interval (Carney and Schlauch, 2007). Significant differences based on a 95% confidence interval are marked by the asterisks. Note: $*p<0.05$.

CHAPTER 5

DISCUSSION

The primary goal of the current study was to investigate the contribution of a frequency-compression hearing aid for contralateral CI performance. This goal, as examined in experiment one, was separated into two research questions. The first question was to investigate sound localization using the CI+HA, comparing performance with a frequency-compression hearing aid to a conventional hearing aid. The second research question was to determine the influence of frequency compression on the binaural advantage by testing speech perception with both devices (CI+HA) and each device alone (CI only and hearing aid only) compared to conventional amplification. As investigated in experiment two, the second goal of this study was to determine the impact of a frequency-compression hearing aid on consonant and vowel perception in quiet. In both experiments, audibility was measured and quantified by comparing estimations of the Speech Intelligibility Index (SII) (ANSI S3.5-1997) for the frequency-compression hearing aid to that of the conventional hearing aid.

5.1 Experiment 1: Contribution of Frequency Compression to CI+HA Performance

5.1.1 Results from the Present Study

The first hypothesis of this study was that, following a period of adjustment, sound localization would be improved using a frequency-compression hearing aid compared to a conventional hearing aid. Results testing this hypothesis are shown in Figures 4.1-4.3 (individual results in Appendix D). Paired *t*-tests were used to compare performance between the frequency-compression and conventional hearing aid after two months, as well as the rate of change between the two conditions from baseline to two months. In both comparisons, there was no significant difference in localization performance between the frequency-compression and conventional hearing aids. The hypothesis that frequency compression would provide better localization abilities because more high-frequency cues are available was not supported by the results of this study. This suggests that, for sound localization, frequency compression is not a better bimodal option than conventional amplification for the subjects in this study.

The individual data from the localization test on the rate of change in performance from baseline to two months, as seen in Figure 4.2, supported the notion that there was a trend of improvement in localization scores overtime. This trend was observed for 7 out of 10 subjects for both the CI+HA-conventional and CI+HA-compressed conditions. This suggests that, although mean localization performance changed at the same rate using conventional amplification and frequency compression from baseline to two months, localization abilities improved overtime for some, but not all subjects. Individual difference scores, shown in Figure 4.3, comparing the two conditions at two months resulted in a significant improvement for one subject using conventional amplification and no significant difference for the remaining nine subjects. Therefore, with the exception of one subject, no significant difference was found on the localization test using frequency compression and conventional amplification after two months listening experience.

In the introduction, we discussed that the salient frequencies for sound localization are 2000 to 6000 Hz based on ILD cues as evidenced by HTRFs. Furthermore, it is important that these frequencies be coded adequately by the hearing aid and cochlear implant devices for CI+HA users to receive the necessary cues for sound localization. For the algorithm studied here, frequency compression occurs within the same high-frequency region where the important sound localization cues are found. That is, frequency compression begins at frequencies of 1500 to 6000Hz using a compression factor of 4.1:1 to 1.5:1. For 9 of 10 subjects in this experiment, the settings for the cutoff frequency were at 1500 Hz or 1600 Hz and compression ratios ranged from 4.1:1 to 1.5:1. The output for these settings would be between 2000 and 3500 Hz, thereby compressing high-frequency input signals in a range of 2000 to 3500 Hz for these subjects. Although the use of frequency compression may have provided additional audibility for high-frequencies, it is not unreasonable to presume that the large amount of frequency compression applied to the signal negatively impacted the perception of the high-frequency cues in the 2000 to 6000 Hz region. As a result, this could explain why sound localization was not improved using frequency compression as shown by the group data in this study.

It is possible that the frequency-compression algorithm used in this study might have also distorted important timing cues necessary for sound localization. As discussed in the introduction, the signal processing of the frequency-compression hearing aid in this study produced a 6 ms delay in the low-bass band while signals were processed and compressed in the high-pass band. In addition to the change along the spectral domain using frequency compression, the digital processing delay introduced in the time domain, referred to as the ‘group delay’ (Dillon et al., 2003), may also have a detrimental affect on localization abilities when a cochlear implant is used contralaterally.

Data from the localization test were analyzed by comparing the overall RMS error in degrees (°). This was calculated by a) finding the RMS error for each speaker, which takes into account the perceived and actual speaker locations and the number of presentations per speaker, b) computing the total RMS, that is the squared RMS for each speaker averaged across the number of speakers, and c) finding the total RMS error by multiplying the total RMS by the distance from two loudspeakers (15.5°). Thus, the overall RMS error is a single number that represents the variability across speakers. In this study, there were an unequal number of stimuli presentations per speaker, i.e., 16 total stimuli presented 12 times from each loudspeaker. It is important to note that this is a potential confounding factor, which may influence the results when a single number is reported as the final outcome.

The second hypothesis of this study was that a frequency-compression hearing aid would provide improved spondee-in-noise performance compared to a conventional hearing aid when a CI is used contralaterally. As reported in section 4.1.3, there was a significant effect in spondee-in-noise performance after two months of listening experience. However, the results did not support the hypothesis that frequency compression provided better performance than conventional amplification. Instead, performance after two months was significantly better for the CI+HA-conventional condition (-6.68 SNR) compared to the CI+HA-compressed condition (-4.33 SNR). Moreover, individual difference scores, shown in Figure 4.7, at two months showed better performance for the CI+HA-conventional condition over the CI+HA-compressed condition for five subjects, S2, S5, S7, S15, and S16. Lastly, performance using the CI+HA with conventional amplification and frequency compression improved overtime by 1.71 and 0.14, respectively. However, the rate of performance change between the two conditions was not significantly different. Taken together, the group and individual data indicate that frequency compression does not provide better spondee-in-noise performance compared to conventional amplification.

The two-month results indicating improved performance for conventional amplification over frequency compression on this task are not entirely surprising. Following the at-home trial with the two hearing aid programs, several subjects reported that the frequency-compression hearing aid sounded distorted, harsh, or produced an echo quality. As seen in Figure 4.9, subjective responses to the item ‘noise in program’ from the Sound Quality Questionnaire suggested that the frequency-compression hearing aid had more noise in the program than the conventional hearing aid. Achieving a better score on the spondee-in-noise task depends on the degree to which the spondee word can be parsed out or segregated from the background noise. Indeed, if the frequency-compression hearing aid provided a noisier or more distorted signal than the conventional hearing aid, the ability to segregate the target and jammer would likely be adversely affected, explaining why performance was poorer using the frequency-compression hearing aid.

Results from this study suggested that CI+HA use provides improved spondee-in-noise performance compared to CI only or hearing aid only use. Figure 4.6 shows the results for the bimodal conditions and each unilateral mode (CI and hearing aid) for frequency compression and conventional amplification. Statistical analysis using paired *t*-tests revealed a significant difference in performance between the CI+HA and the hearing aid only with or without frequency compression. Additionally, performance using the CI+HA-conventional and the CI only was significantly different.

These results suggesting better speech perception using the CI+HA over the CI and the hearing aid only can be attributed to the advantages of hearing with two ears based on the effects of binaural summation and binaural squelch. However, it is important to emphasize that these findings were only observed for conventional amplification. Put another way, performance between the CI+HA-compressed condition and the CI only was not significantly different, indicating that a binaural advantage was not observed for all bimodal and unimodal comparisons using frequency compression.

Self-report outcomes from the Spatial Hearing Questionnaire and Sound Quality Questionnaire confirmed that frequency compression was not a better bimodal option than conventional amplification. Results from the two questionnaires are shown in Figures 4.8 and 4.9. Statistical analysis of the questionnaires revealed no significant difference on the eight subscales or the total score from the Spatial Hearing Questionnaire and for five of the six items from the Sound Quality Questionnaire. There was one exception in that speech understanding was found to be significantly higher using conventional amplification compared to frequency compression on the Sound Quality Questionnaire. In summary, the data from this experiment suggest that frequency compression is not a better bimodal option than conventional amplification and, in some cases, may have a negative impact on speech perception as documented through objective test measures and subjective report.

In this first experiment, there were three individuals (S2, S7, and S10) that did not show an improvement in SII using the frequency-compression hearing aid compared to the conventional aid (see Table 3.3). Of the seven individuals with an improved SII using frequency compression, the margin of improvement was quite small, ranging from .005 to .041. On the other hand, for all eight subjects that were tested, sound field thresholds improved by at least 10 dB for one or more frequencies using frequency compression. Taken together, these results indicate that although frequency compression improved high-frequency sound detection, the margin of SII improvement was small for many subjects in this study. Therefore, the effectiveness of frequency compression may not have been remarkable for these subjects in experiment one, which is likely limited to the severity of their hearing loss.

Another major limiting factor of this experiment was the lack of participants. All adults with a CI+HA were recruited from the University of Iowa where nearly 65 adults are implanted per year and an active cochlear implant research program is in place. Despite best efforts to recruit more subjects, the total number of participants remained at 10. In fact, of the approximately 30 potential candidates from the University of Iowa, eleven individuals were dismissed due to unaided hearing thresholds in the profound hearing loss region at or below 2000 Hz. With auditory thresholds that severe in the mid-frequency range, fitting of the frequency-compression algorithm used in this study would likely not have been appropriate. As shown in Table 3.2, there were three participants in the study (S2, S7, S10) that had unaided thresholds $\geq$95 dB HL at 2000 Hz.

5.1.2 Comparison to Previously Published Literature

The results from this study suggesting that frequency compression does not provide better bimodal benefit than conventional amplification are consistent with previous findings. A recent study investigated the benefits of frequency compression to contralateral CI use for eight subjects using the identical frequency-compression algorithm studied here (McDermott & Henshall, 2010). Results revealed no difference in performance using the CI+HA with or without frequency compression on tests of consonant recognition in quiet and sentence recognition in noise.

In that study, two different compression settings were evaluated, either maximum compression (or cutoff frequency = 1500 Hz; compression ratio = 4:1.1) or the manufacturer's recommended setting (e.g., cutoff frequency = 2990 Hz; compression ratio = 2:1.1). Following two weeks of use, the subjects rated the two frequency compression settings for ease of listening and sound quality and the preferred setting was selected. Interestingly, five of the eight subjects selected the maximum compression setting as their preferred frequency compression setting. All five of these subjects had unaided audiometric thresholds ≥95 dB at 1500 Hz and four with thresholds ≥100 dB at 2000 Hz (McDermott & Henshall, 2010). Unaided thresholds for these subjects were slightly poorer than the subjects in the present study. Therefore, the difference in hearing thresholds could contribute to the preference for the more severe compression setting observed by McDermott & Henshall (2010) that was not observed in the current study.

The impact of frequency compression on localization performance has not been well documented in the literature. Only one study has investigated horizontal localization using frequency compression compared to conventional amplification and results revealed no significant difference for the participants with bilateral hearing aids (O'Brien et al., 2010). In that study, horizontal localization was tested using a 360° arc and broadband, pink-noise stimuli were presented. Further, data from this study were analyzed for front/back confusions by comparing the difference in RMS error for the presentation versus response azimuth along the front/back dimension. Given that evidence from normal hearing subjects suggests that front/back confusions are best resolved when stimuli include high frequencies (Gelfand, 2004), it is expected that frequency compression would provide significantly improved localization ability than conventional amplification. Although frequency compression is believed to extend the input bandwidth to provide more high-frequency sounds to the listener, the results from O'Brien et al. (2010) and the current study suggest that compression of high-frequency cues does not provide improved localization ability compared to conventional amplification.

The effect of the binaural advantage using a CI+HA compared to CI only or hearing aid only for speech perception has been well documented in the literature (e.g., Armstrong et al., 1997; Ching et al., 2004, 2005; Dunn et al., 2005; Luntz et al., 2005; Mok et al., 2006; Potts et al., 2010; Tyler et al., 2002). Although the fitting methods for the hearing aid and cochlear implant devices vary across these studies, all of these studies have used conventional amplification in fitting the hearing aid. In contrast, research on frequency lowering has not necessarily found a binaural advantage when comparing bimodal versus unimodal performance. Chute et al. (1995) reported that use of a different algorithm of frequency-lowering for two months did not significantly improve performance over the CI alone for four subjects. This result was documented on tests of phoneme and word discrimination and sentence recognition in quiet. Additionally, McDermott & Henshall (2010) reported a small but significant improvement using the CI+HA over the CI and hearing aid alone on sentences in noise, but not for consonant perception. Along with the results of this study, it appears that frequency-lowering hearing aids do not consistently contribute to the binaural advantage as has been observed with conventional amplification.

Thus far, research studies investigating frequency lowering and the impact on CI+HA use have been limited by the small number of participants, ranging from 5 (Chute et al., 1995) to 10 as in the current study. This small number of subjects makes the results of these studies difficult to generalize to a larger population and should be interpreted with caution. Along with the results from the current study, research would suggest that individuals with severe-to-profound hearing loss using a CI+HA would likely not benefit further by use of frequency compression. Individual differences do exist, but overall, the results to date do not support frequency compression as a better bimodal option than conventional amplification.

5.2 Experiment 2: Impact of Frequency-compression on Speech Perception

5.2.1 Results from the Present Study

The first hypothesis examined in this experiment was that consonant perception would be improved using frequency compression compared to conventional amplification. Results testing this hypothesis were presented in Figures 4.10-4.12 (refer to Figure D6 for individual data). Statistical analysis of consonant recognition scores using conventional amplification and frequency compression revealed no significant difference after two months of listening experience, and likewise in the rate of change between the two conditions overtime. As a result, the hypothesis that consonant perception would be improved using a frequency-compression hearing aid compared to a conventional hearing aid due to better access to high-frequency cues was not supported. In experiment two, the improvement in SII values using frequency compression compared to conventional amplification ranged from .005 to .087 and the improvement in sound field thresholds ranged from 10 to 35 dB (refer to Table 3.7). Although an increase in the SII value and/or improvement in aided sound field threshold was found for 16 of 17 subjects using frequency compression, the improvement was not remarkable for all subjects. Therefore, the overall impact of frequency compression on consonant perception may have been negligible.

The individual data from the consonant recognition test showed no significant difference for 15 of 17 subjects using the HA-conventional and the HA-compressed conditions after two months of listening experience. In contrast, individual difference scores based on a 95% confidence interval were significantly better using the conventional hearing aid for two subjects (S6, S14). For both of these subjects, audibility did not appear to account for this difference, as the SII value increased using the frequency-compression hearing aid compared to the conventional hearing aid. However, both subjects reported having difficulty adjusting to the frequency-compressed signal and also noted a preference for conventional amplification at the end of the home trial. Furthermore, data logging of subject S14's hearing aid use suggested that the conventional hearing aid was used more often and would likely explain his preference for conventional amplification (i.e., 75% use with the conventional condition vs. 25% for the frequency-compressed condition).

The second hypothesis stated that vowel perception would not be affected by frequency compression because the frequency compression algorithm in this study did not compress frequencies below 1500 Hz where the energy from the first formant of vowels is mostly concentrated. Section 4.2.3 summarizes the results pertaining to this hypothesis. The hypothesis was not supported by the results of this study, but instead significantly higher vowel perception scores were found for the conventional hearing aid after two months of use compared to the frequency-lowered hearing aid, as shown in Figure 4.13. Using the frequency-compression and conventional hearing aids, performance improved from baseline to two months by 4.9% and 6.7% respectively, but this also was not significantly different between the two conditions. In sum, the mean results indicate that a frequency-compression hearing aid does not provide improved speech perception abilities compared to a conventional hearing aid.

A comparison of critical difference scores on the vowel perception test revealed that 15 of 17 subjects showed no significant difference between frequency-compressed and conventional conditions after two months. There were two subjects (S10, S15) that showed a significant improvement using the conventional hearing aid over the frequency-compression hearing aid. Additionally, five showed no difference in scores (e.g., 0%)

between the HA-conventional and HA-compressed conditions. Of these five subjects, three subjects (S3, S18, and S20) scored at or above 90% on the vowel perception test. Moreover, individual results from Figure D7 (refer to Appendix D) revealed that eight total subjects (S3, S11, S14, S15, S18, S19, S20) performed ≥80% with either the conventional or frequency-compression hearing aid. Thus, nearly 50% of the subjects in the study exhibited a ceiling effect on this vowel perception task. This potential bias in the results may have an affect on the change in scores overtime as well as the differences in performance between the conventional and frequency-compression hearing aids observed at the end of the home trial. Thus, scores exhibiting a ceiling effect would show little improvement overtime or difference in scores among the two conditions even if there was a true change in performance.

Results from the vowel recognition test also indicated that vowel perception may be negatively influenced by frequency compression. As suggested by Alexander (2009), there is a tradeoff of extending the input bandwidth of the hearing aid to code more high-frequency sounds using frequency compression. In so doing, formant frequencies may be too severely compressed and can no longer be distinguished. Using a lower cutoff frequency (e.g., 1500 Hz), the vowel space along the dimension of the second formant frequency is compressed or reduced to a greater extent than when a higher cutoff frequency (e.g., 6000 Hz) is used. Furthermore, a high compression ratio (e.g., 4:1.1) can greatly reduce the overall level between the second and third formant frequencies, creating spectral smearing of the input signal compared to a lower compression ratio (e.g., 1:5:1). These effects of vowel reduction and spectral smearing arise when frequency compression is used and could explain the result that frequency compression can have a negative impact on vowel perception.

Overall, the results of this experiment did not support use of frequency compression as a way to improve speech perception abilities. Furthermore, individual data from the consonant and vowel recognition tests revealed that 15 of 17 individuals showed no significant difference using conventional amplification compared to frequency compression with listening experience. It is possible that the vowel reduction and spectral distortion introduced by compression of frequencies may be detected by some individuals, suggesting that conventional amplification provides better speech perception for some individuals compared to frequency compression.

5.2.2 Comparison to Previously Published Literature

Research studying the impact of frequency compression on consonant recognition has not produced consistent findings. Results from two studies (Glista et al. 2009; Simpson et al., 2005) reported significantly improved consonant perception abilities with frequency compression whereas results from this study and others have shown no significant difference compared to conventional amplification (McDermott & Henshall, 2010; Simpson et al., 2006). Upon further inspection, results from Simpson et al. (2005) demonstrated that individual subjects performed differently with frequency compression and conventional amplification based on consonant phoneme scores from the CNC word recognition test (Peterson & Lehiste, 1962). Moreover, 7 of 17 subjects performed better for consonant phonemes with frequency compression, 9 showed no difference in scores between the two conditions, and 1 subject showed a significant decrease in scores with frequency compression. These data agree with that from the present study that most subjects demonstrated no difference in consonant recognition between frequency compression and conventional amplification. However, as reported by Simpson et al. (2005), there may be some individuals who perform better on consonant recognition tasks using a frequency-compression hearing aid following a period of adjustment or listening experience.

In the study by Glista et al. (2009), the participants included 13 adults as well as 11 children. In addition to differences in age among the study participants, there were differences between the children and adult subjects in audiometric thresholds and the amount of prescribed hearing aid gain (e.g., DSL v. 5.0 prescribes more gain for children) (Glista et al., 2009). The factors of age, audiometric threshold, and hearing aid gain may very well have had an influence on the overall outcomes that are not consistent with results from the other studies on frequency compression.

For vowel perception, studies in the published literature have showed no differences in performance for frequency compression and conventional amplification. For instance, Glista et al. (2009) found no significant difference in vowel perception using frequency compression compared to conventional amplification based on mean scores from 24 subjects. The vowel perception task using the /hVd/ format was similar to that in the present study. Simpson et al. (2005) found no difference between frequency compression and conventional amplification in vowel phoneme scores for most of the subjects (12 of 17) in the study. Therefore, results based on group findings from the published literature largely suggest no improvement in consonant and vowel perception abilities using frequency compression compared to conventional amplification. Furthermore, results from the current study would also suggest that compressing frequencies can have a negative impact on vowel perception. Thus, potential negative effects of spectral smearing and formant frequency alteration should be more carefully studied with regard to current frequency lowering systems.

5.3. Implications for Clinical Management

The results from the present study have several implications for clinical management of patients with CIs and hearing aids. First of all, frequency compression as a bimodal fitting option is not recommended and should be used with caution. Results from this study, along with those from McDermott & Henshall (2010), demonstrate that frequency compression is not a better option than conventional amplification. In the current study, not one individual subject performed better using frequency compression compared to conventional amplification on any of the test measures. Indeed, individual data from a single subject (S10) suggested consistently better scores using conventional amplification over frequency compression on two tests. Of course, the results from the present study generalize to adult populations only. The use of frequency-compression hearing aids for children have shown promising results (e.g., Glista et al., 2009), suggesting that frequency compression may be more beneficial for pediatric populations than was demonstrated for adult subjects in this study.

Secondly, for patients using a single CI with residual hearing in the opposite ear, use of a contralateral hearing aid should continue to be recommended. Results from the present study, as well as evidence from the published literature (e.g., Armstrong et al., 1997; Ching et al., 2004, 2005, 2007; Dunn et al., 2005; Flynn & Schmidtke, 2004; Kileny et. al, 2004; Luntz et al., 2005; Mok et al., 2006; Potts et al., 2010; Tyler et al., 2002), indicate that bilateral device use via a CI and hearing aid provide important binaural hearing advantages that can not be gained by use of a CI alone. Although all participants in this study with a CI+HA were experienced hearing aid users, the benefits afforded by CI+HA use have been documented for experienced *and* non-experienced hearing aid users (Ching et al., 2004).

Additionally, many of the participants using a CI+HA had profound high-frequency hearing loss in the aided ear and may be better candidates for bilateral cochlear implantation rather than frequency compression. For example, one subject (S7) showed no improvement with listening experience using the study hearing aid, as scores stabilized to $\leq$30% on consonant and vowel perception tests and remained poor at 43° RMS error on the localization test. For this individual, sequential bilateral cochlear implantation may be a better option to improve speech perception abilities and localization performance than continued use of a contralateral hearing aid. Consistent with previous reports (Ching et al., 2007), recommendations for CI+CI use over bimodal use should continue to be made on an individual basis until controlled, clinical trials have been performed. Furthermore, it is important to consider the amount of residual hearing for each individual as well as the contribution of each ear to the binaural advantage when making fitting recommendations for use of CIs and/or hearing aids (Perreau et al., 2007).

5.4 Future Studies

This study presents several limitations that should be investigated in future studies. In the present study, there were three individuals with audiometric high-frequency thresholds in the profound region that may not have benefited fully from frequency compression due to the severity of their hearing loss. However, individual results from some studies (e.g., Glista et al., 2009) have suggested that frequency compression may be more beneficial for individuals with greater amounts of high-frequency hearing loss. Future studies should more carefully examine the influence of the severity and configuration of hearing loss (e.g., flat versus steeply sloping) on the amount of benefit that can be derived from frequency compression. Based on the results of Simpson et al. (2006) and McDermott & Dean (2000), it is expected that frequency compression will prove less beneficial for listeners with a steeply sloping hearing loss compared to those with a flat hearing loss. Additionally, because the effects of dead regions have previously been reported as an influencing factor for persons with steeply sloping losses, dead regions should be considered in future studies when fitting frequency-compression hearing aids. As demonstrated by Robinson, Baer, & Moore (2007), the frequency compression cutoff can be individually selected based on possible dead regions. This provides a potential advantage to improve audibility for listeners with dead regions because frequency compression moves the input signal from the dead region into a region of better hearing sensitivity.

Future studies should investigate whether fitting a frequency-compression hearing aid requires more or less gain compared to fitting a conventional hearing aid. In the present study, frequency compression was activated after hearing aid gain was set using conventional amplification. Loudness balancing between the bilateral devices occurred after the fitting of the hearing aid. However, additional gain may often be required compared to a conventional hearing aid fit and the degree to which additional gain is needed should be further examined. In addition, the impact of the frequency compression settings (e.g., cutoff frequency and compression ratio) on speech perception and sound quality should be closely examined. Current studies vary considerably in how frequency compression is set, yet compressing high-frequencies too strongly can have detrimental affects on speech perception and should be further studied.

Other frequency lowering options should be investigated and/or developed with the goal to improve audibility for the high frequencies and provide better speech perception abilities. For instance, is frequency transposition a better option to improve high frequency audibility compared to frequency compression? In what ways can the current frequency-lowering technology be improved to provide better results for speech perception and localization? Finally, efforts to improve CI+HA fittings should also continue, as Ching et al. (2004) demonstrated that loudness balancing between the CI and hearing aid is effective when performed for each individual. Can CI+HA fittings be improved by matching loudness at multiple azimuths to provide better localization? What is the effect of minimizing processing time delays between the CI and hearing aid on speech perception and localization, as often results when using amplitude compression and directional microphones? Because research shows that CI+HA use provides better speech perception and localization abilities than a CI alone, future studies should continue to study ways to improve CI+HA fittings.

CHAPTER 6

CONCLUSIONS

This study investigated the contribution of a frequency-compression hearing aid to bimodal performance on tests of sound localization and spondee-in-noise using spatially-separate target and noise sources. In addition, the impact of frequency compression on consonant and vowel perception abilities was studied. The results from the first experiment did not support the hypothesis that frequency compression was a better bimodal option than conventional amplification. Instead, results revealed that sound localization was not different using frequency compression and conventional amplification even after a period of adjustment for these listeners with moderate-to-severe hearing loss. On the spondee-in-noise test, group and individual results suggested that performance was significantly better using conventional amplification. Self-report outcomes from two questionnaires revealed no significant improvement using frequency compression compared to conventional amplification after listening experience. The hypothesis in the second experiment of this study was also not supported by the results. The group data demonstrated that, for the subjects in this study, frequency compression does not provide improved speech perception abilities compared to a conventional hearing aid. In fact, vowel perception abilities were significantly better using conventional amplification compared to frequency compression after listening experience. The individual data showed no significant improvements using frequency compression for either consonant or vowel recognition abilities. Furthermore, there were 2 of 17 individuals who performed significantly better using conventional amplification on the consonant and vowel tests compared to frequency compression. Group and individual results from this study suggest that the frequency compression scheme studied here does not provide better speech perception abilities compared to conventional amplification.

APPENDIX A

VERIFICATION OF FREQUENCY LOWERING TECHNOLOGY

Twelve Phonak Naida IX SP and two Phonak Naida V UP hearing aids were provided by Phonak AG in order to complete this study. Prior to the onset of the study, the frequency lowering algorithm and the overall function of each hearing aid were verified by analyzing the spectral output and performing electroacoustic tests according to ANSI standards (1996). Electroacoustic measurements were completed individually for each hearing aid with the volume control set at full-on and at reference test gain position. All hearing aids were determined to be functioning properly with adequate gain, output sound pressure level (OSPL), and appropriate distortion and equivalent input noise. For the spectral analysis, stimuli were created digitally on a computer using Adobe software and the frequency output was analyzed using Spectra Pro. Hearing aid measurements were made using a Bruel & Kjaer anechoic test box. The stimuli consisted of a modified pure tone sweep where a 30 s tone was presented every 500 Hz beginning at 1000 Hz and ending with 4500 (see part b below) or 6500 Hz (as in part a). Between each 500 Hz interval, a 10 s pause or off-phase was included. The intensity of each frequency tone was 64 dB SPL with variation <1 dB across frequencies. Two spectral analyses were completed: a) to determine the spectral output for all thirteen cutoff frequency and compression ratio settings plus the uncompressed condition, and b) to determine if the hearing aids produced the same spectral output for a given frequency cutoff and compression ratio. In part b, the frequency cutoff was set to 1500 Hz and the compression ratio was set to 4.1:1, the condition producing the most severe frequency compression. For both analyses, the hearing aid programming was completed using Phonak I-PFG v. 2.4 and NOAH software.

Figure A1 displays the spectral input (shown on x-axis) vs. the output (shown on y-axis) for the thirteen cutoff frequency (CF) and compression ratio (CR) settings plus the uncompressed condition. The uncompressed condition is shown by a diagonal line, indicating no compression of the input. Conversely, the most severe frequency compression condition (CF = 1.5; CR = 4.1:1) is shown on the lower portion of the graph. Considering this condition by itself, for inputs of 1500 to 6500 Hz, the output using 4.1:1 compression and a cutoff frequency of 1500 Hz is less than 2000 Hz.

Figure A2 displays the input (along x-axis) vs. the output (along y-axis) for 11 frequency-lowering hearing aids programmed to provide the most severe frequency compression (CF = 1.5; CR = 4.1:1). The horizontal dotted black line indicates the pre-determined frequency cutoff of 1500 Hz. The overall output ranges from 990.53 Hz to 1787.26 Hz for inputs of 1000 Hz to 4500 Hz. For an input of 2000 Hz, there were significant differences in output. Four of 11 hearing aids produced an output of 1528.86 Hz where as 7 of 11 hearing aids (circled in red) produced an input of 1356.59 Hz. Because a frequency cutoff of 1500 Hz was selected, it would be predicted that no hearing aid should have an output below 1500 Hz. However, the results of this analysis show that over half (7/11) of the hearing aids indeed have a lower output than the pre-determined frequency cutoff.

Following this result, the hearing aid manufacturer, Phonak AG, was contacted to provide a possible explanation. According to one Phonak researcher, the observed errors can be attributed to two factors (M. Boretski, personal communication, December 21, 2009). First, above the cutoff frequency, frequencies are compressively shifted with a frequency resolution of 160 Hz. This is both resolution of the Fast Fourier Transform (FFT) and is the basis of frequency compression. As a result, an error of up to +/- 80 Hz is possible for particular frequencies. Second, the internal oscillator affects the frequency compressor and produces some error that may change the FFT bin assignment. The internal oscillator is the element in the system which triggers the sampling rate in analog-digital conversion and serves as the internal ‘clock’ of the system. Therefore, the hearing aids with the processing errors were noted, but these devices were used in the study only when needed. It was suggested that the processing errors would likely not influence the processing of broadband sounds, such as everyday sounds and speech (M. Boretski, personal communication, December 21, 2009).

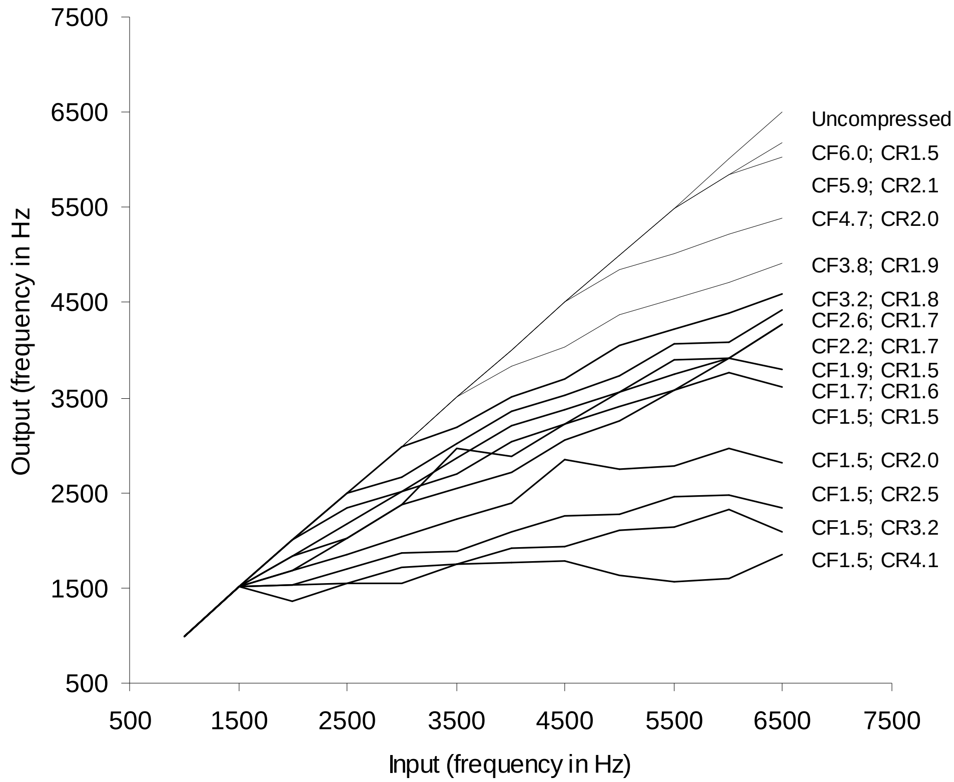


Figure A1. Input/output curves for thirteen different cutoff frequency and compression ratio settings plus the uncompressed condition. Data are shown in frequency (Hz).

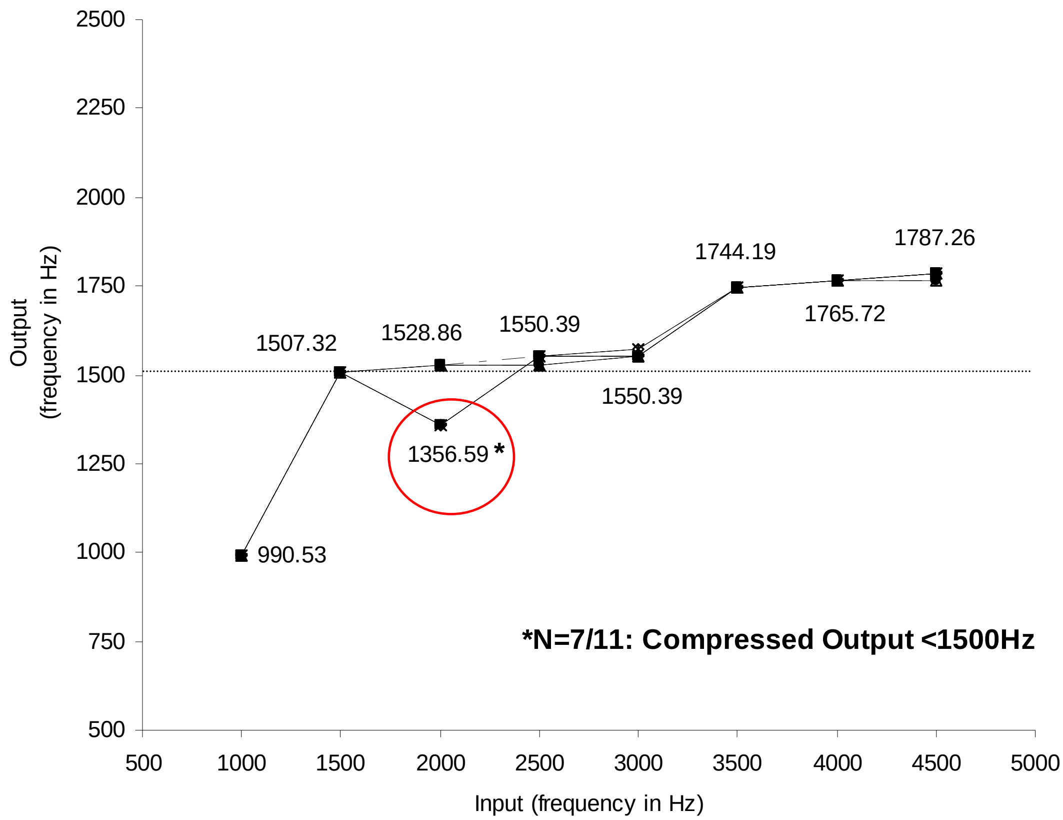


Figure A2. Output of eleven frequency-compression hearing aids as a function of the input frequency (in Hz).

APPENDIX B

MODIFIED AUDIBILITY INDEX FOR FREQUENCY-COMPRESSION HEARING AIDS

To verify the audibility for a frequency-compression hearing aid, an algorithm was recently developed that derives SII values for frequency compression (Bentler, Cole, & Wu, 2011). This derivation was implemented in software that was modified for this purpose by the developer of the Verifit probe microphone system (Cole, Personal Communication). In this algorithm, the weighted audibility for frequency compression was calculated by using band-passed speech stimuli from the Verifit. The SPL thresholds of the amplified standard speech signal were adjusted for four one-third octave speech bands (e.g., 3150, 4000, 5000 and 6300 Hz) to produce the same sensation level (SL) as the speech stimuli after frequency compression. Then, audibility for those frequencies of 3150, 4000, 5000, and 6300 Hz was multiplied by the band importance functions (BIFs; ANSI S.35-1997) for the standard speech stimuli and the level distortion factor (LDF) for the level of the band-passed speech stimuli after frequency compression.

Figures B1 and B2 illustrate the calculation of the modified audibility index for the frequency-compression hearing aids. Figure B1 displays the data entry screen from the software where (from left to right) hearing thresholds, prescriptive targets, and the real ear aided response (REAR) were entered into the algorithm for calculation. The REARs for the amplified standard speech signal and frequency-compressed signal were both used in the calculation of audibility. The SPL thresholds (TSPL) and aided LTASS data for each one-third octave band were then transposed to Figure B2 (shown in columns G and H) for calculation of the modified SII for frequency compression. As shown in Figure B2, the SII calculation occurred in six steps. First, audibility was calculated for each one-third octave band by the equation,

$$A = (LTASS_a - TSPL + SFA)/30$$

where A = audibility, $LTASS_a$ = aided LTASS, TSPL = SPL threshold, Stimulus Factors Adjustment (SFA) = 15 + Minimum Audible Pressure (MAP) – Bandwidth adjustment – Average real ear unaided gain (REUG) – Ref. Internal spectrum noise level (x). Once audibility was quantified, the second step was to convert this value to a number between 0 and 1. In step three, the level distortion factor (L) was computed by

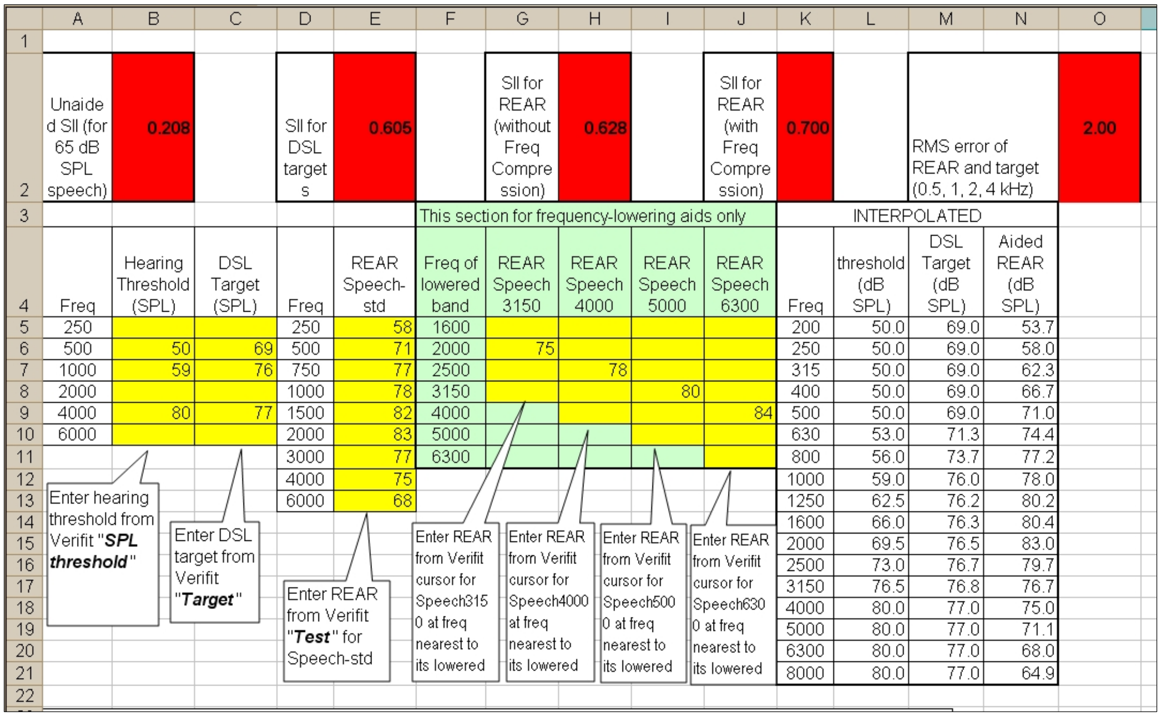
$$L = LFA - 0.00625 * LTASS_a$$

where $LTASS_a$ = aided LTASS, Levels Factor Adjustment (LFA) = 1 + (Bandwidth Adjustment = Average REUG + SII standard normal speech + 10)/60. As with audibility, the fourth step was to convert the level distortion factor to a number between 0 and 1. Next, the SII band contribution was calculated in step five by

$$\text{SII for each band} = A * BIF * L$$

where A = audibility, BIF = band-importance function, L = level distortion factor. Lastly in step six, the band contributions were summed to derive the final SII value for the frequency-compressed audibility.

For the subjects in this study, SII for the compressed condition was calculated for each individual as described here. Two researchers, including the author, calculated these measurements from the real ear measurements obtained during the hearing aid fitting. Interrater reliability was very good and differences in SII for the compressed condition were less than 0.105 for all subjects combined in experiments one and two.



	A	B	C	D	E	F	G	H	I	J	K	L	M	N	O
1															
2	Unaided SII (for 65 dB SPL speech)	0.208		SII for DSL targets	0.605		SII for REAR (without Freq Compression)	0.628		SII for REAR (with Freq Compression)	0.700		RMS error of REAR and target (0.5, 1, 2, 4 kHz)		2.00
3						This section for frequency-lowering aids only						INTERPOLATED			
4	Freq	Hearing Threshold (SPL)	DSL Target (SPL)	Freq	REAR Speech-std	Freq of lowered band	REAR Speech 3150	REAR Speech 4000	REAR Speech 5000	REAR Speech 6300	Freq	threshold (dB SPL)	DSL Target (dB SPL)	Aided REAR (dB SPL)	
5	250			250	58	1600					200	50.0	69.0	53.7	
6	500	50	69	500	71	2000	75				250	50.0	69.0	58.0	
7	1000	59	76	750	77	2500		78			315	50.0	69.0	62.3	
8	2000			1000	78	3150			80		400	50.0	69.0	66.7	
9	4000	80	77	1500	82	4000				84	500	50.0	69.0	71.0	
10	6000			2000	83	5000					630	53.0	71.3	74.4	
11				3000	77	6300					800	56.0	73.7	77.2	
12				4000	75						1000	59.0	76.0	78.0	
13				6000	68						1250	62.5	76.2	80.2	
14											1600	66.0	76.3	80.4	
15											2000	69.5	76.5	83.0	
16											2500	73.0	76.7	79.7	
17											3150	76.5	76.8	76.7	
18											4000	80.0	77.0	75.0	
19											5000	80.0	77.0	71.1	
20											6300	80.0	77.0	68.0	
21											8000	80.0	77.0	64.9	
22															

Enter hearing threshold from Verifit "***SPL threshold***"

Enter DSL target from Verifit "***Target***"

Enter REAR from Verifit "***Test***" for Speech-std

Enter REAR from Verifit cursor for Speech3150 at freq nearest to its lowered

Enter REAR from Verifit cursor for Speech4000 at freq nearest to its lowered

Enter REAR from Verifit cursor for Speech5000 at freq nearest to its lowered

Enter REAR from Verifit cursor for Speech6300 at freq nearest to its lowered

Figure B1. The data entry screen for the software showing the SII estimations with and without frequency compression. These SII estimations are shown at the top of the screen in columns H, K. Hearing thresholds, prescriptive targets, and the real ear aided responses (REAR) with and without frequency compression are shown in columns B, C, E-J.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N	O	P	Q	R	S	T
4	Center Freq	Band importance function (BIF)	Stimulus factors adjustment (SFA)	Level Factor Adjustment (LFA)	Minimum Audible Pressure (MAP)		Enter pure-tone TM threshold (dB SPL)	Enter aided LTASS (dB SPL)	STEP 1	STEP 2	STEP 3	STEP 4	STEP 5	STEP 6						
5	200	0.0096	21.55	1.385	22		50.0	53.7	0.841	0.841	1.0498	1	0.0081	0.700	SII					
6	250	0.0151	18.25	1.396	18		50.0	58.0	0.875	0.875	1.0338	1	0.0132							
7	315	0.0292	17.05	1.400	16		50.0	62.3	0.979	0.979	1.0106	1	0.0286							
8	400	0.0444	15.05	1.411	13		50.0	66.7	1.057	1.000	0.9942	0.9942	0.0441							
9	500	0.0583	14.25	1.417	12		50.0	71.0	1.175	1.000	0.9733	0.9733	0.0567							
10	630	0.0659	12.75	1.413	11		53.0	74.4	1.139	1.000	0.9481	0.9481	0.0625							
11	800	0.0717	10.75	1.400	9.6		56.0	77.2	1.066	1.000	0.9177	0.9177	0.0658							
12	1000	0.0825	10.25	1.383	9		59.0	78.0	0.975	0.975	0.8954	0.8954	0.0720			SL for lowered frequency:				
13	1250	0.0851	9.85	1.379	9		62.5	80.2	0.918	0.918	0.8778	0.8778	0.0686			3150	4000	5000	6300	Effective
14	1600	0.0890	10.15	1.387	11.5		66.0	80.4	0.819	0.819	0.8842	0.8842	0.0644	0	1600					threshold
15	2000	0.0906	9.05	1.412	15		69.5	83.0	0.752	0.752	0.9436	0.9436	0.0643	75	2000	5.5				due to freq
16	2500	0.0876	8.25	1.423	16.5		73.0	79.7	0.498	0.498	0.9352	0.9352	0.0408	78	2500		5.0			lowering
17	3150	0.0851	10.55	1.408	15		71.2	76.7	0.535	0.535	0.9075	0.9075	0.0413	80	3150			3.5		71.2
18	4000	0.0778	9.95	1.396	13		70.0	75.0	0.498	0.498	0.8705	0.8705	0.0337	84	4000				4.0	70.0
19	5000	0.0532	10.25	1.354	13		67.6	71.1	0.458	0.458	0.9095	0.9095	0.0222	0	5000					67.6
20	6300	0.0367	8.25	1.317	15.5		64.0	68.0	0.408	0.408	0.8915	0.8915	0.0134	0	6300					64.0
21	8000	0.0187	5.65	1.285	18		80.0	64.9	-0.317	0.000	0.8795	0.8795	0							
22																				

Figure B2. The algorithm for calculating SII for frequency-compression hearing aids illustrating six steps in computing the final SII value. SII was calculated for each one-third octave band as shown in the left most column. Columns I and J represent calculation of audibility for steps one and two based on SPL threshold and aided LTASS (shown in columns G, H); columns K and L shows the level distortion factor calculation for the compressed frequencies in steps three and four; column M represents the calculation of the SII for each band in step five; and column N represents calculation of the final SII value summed across all speech bands.

APPENDIX C

AIDED SOUND FIELD THRESHOLDS

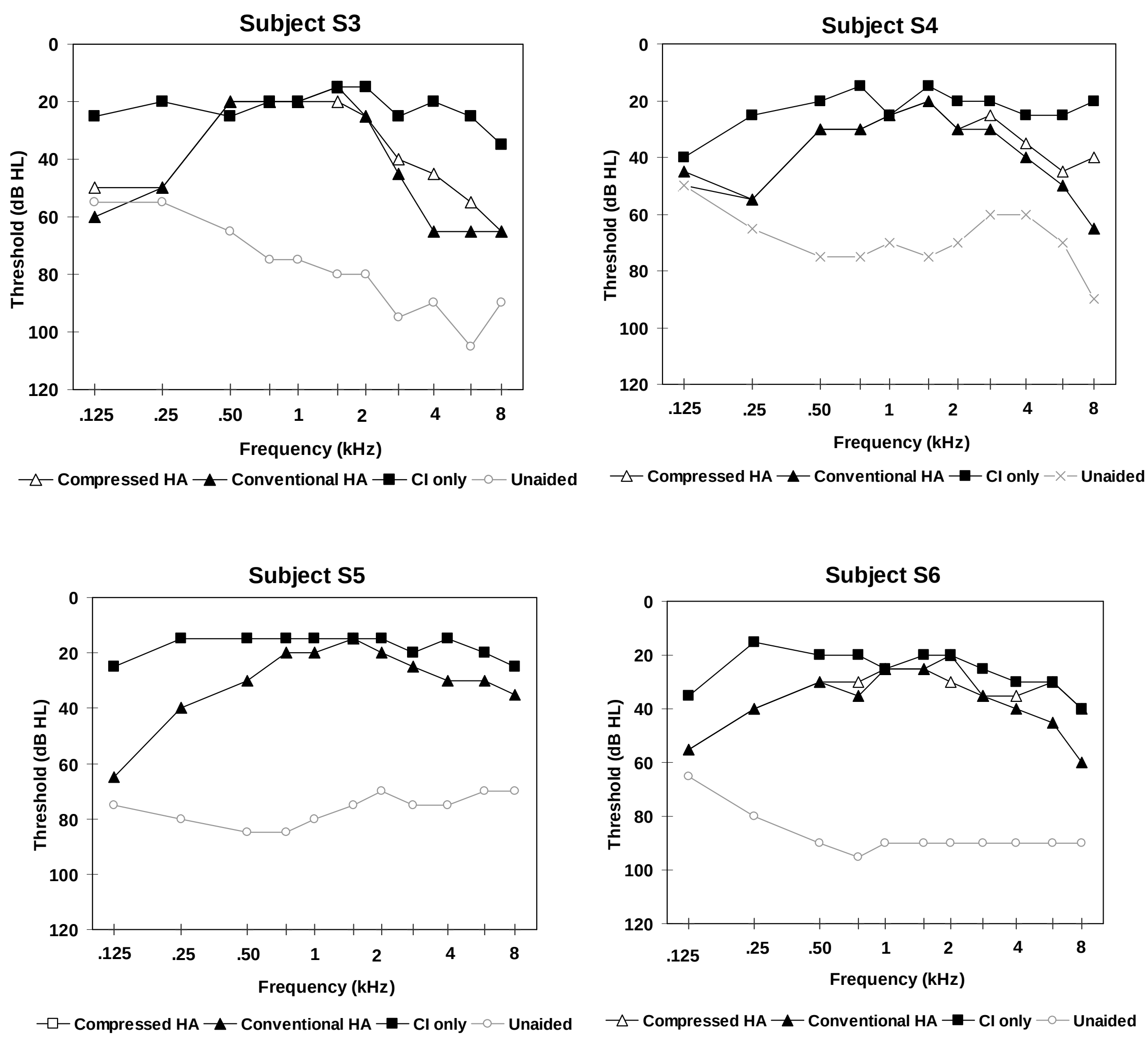
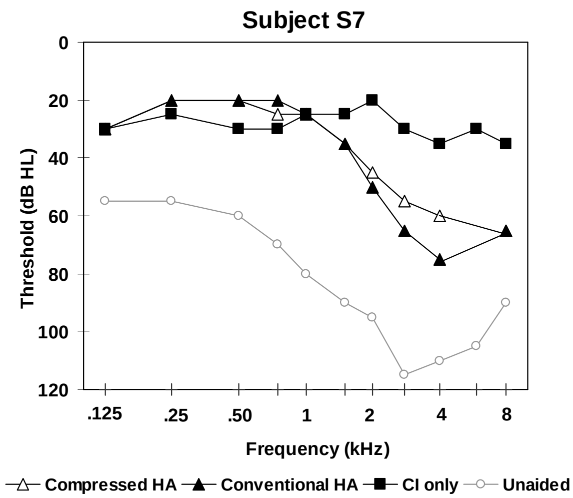
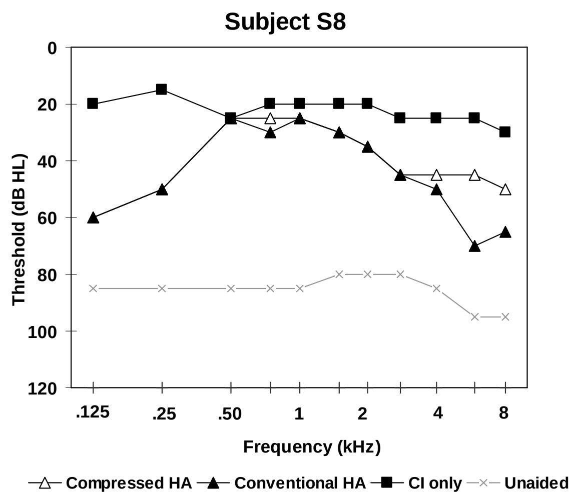
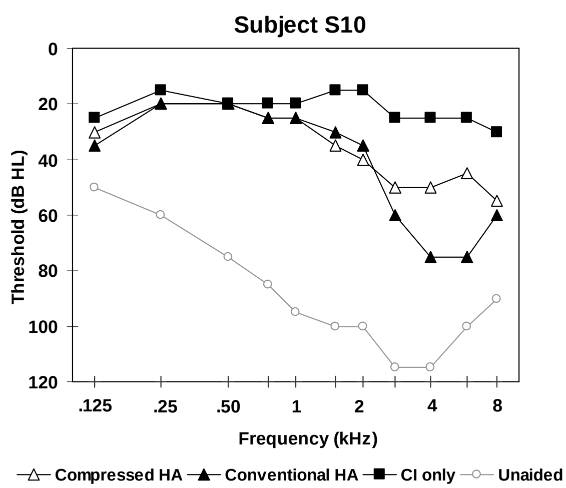


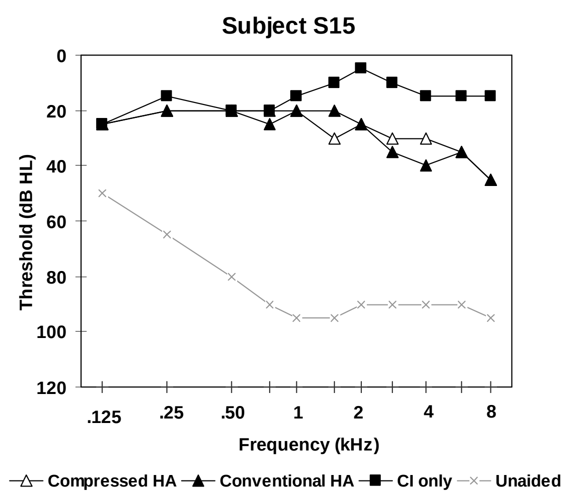
Figure C1. Aided sound field thresholds for subjects in experiment one using the frequency-compression hearing aid, the conventional hearing aid, and the CI only. Data were obtained using one-third octave narrow band noise. Data for subject S5 using the frequency-compression hearing aid could not be measured and were not reported.

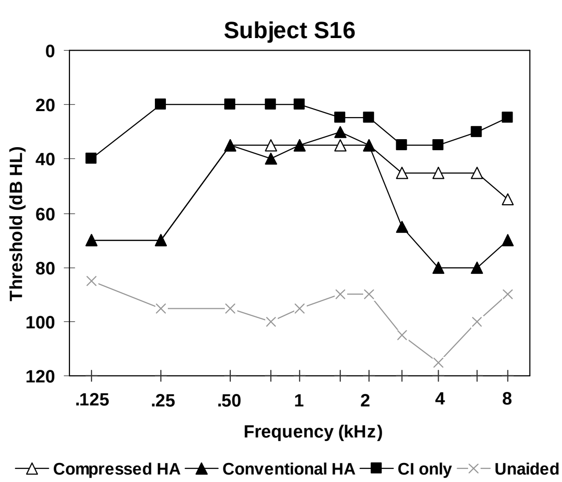
Figure C1—continued











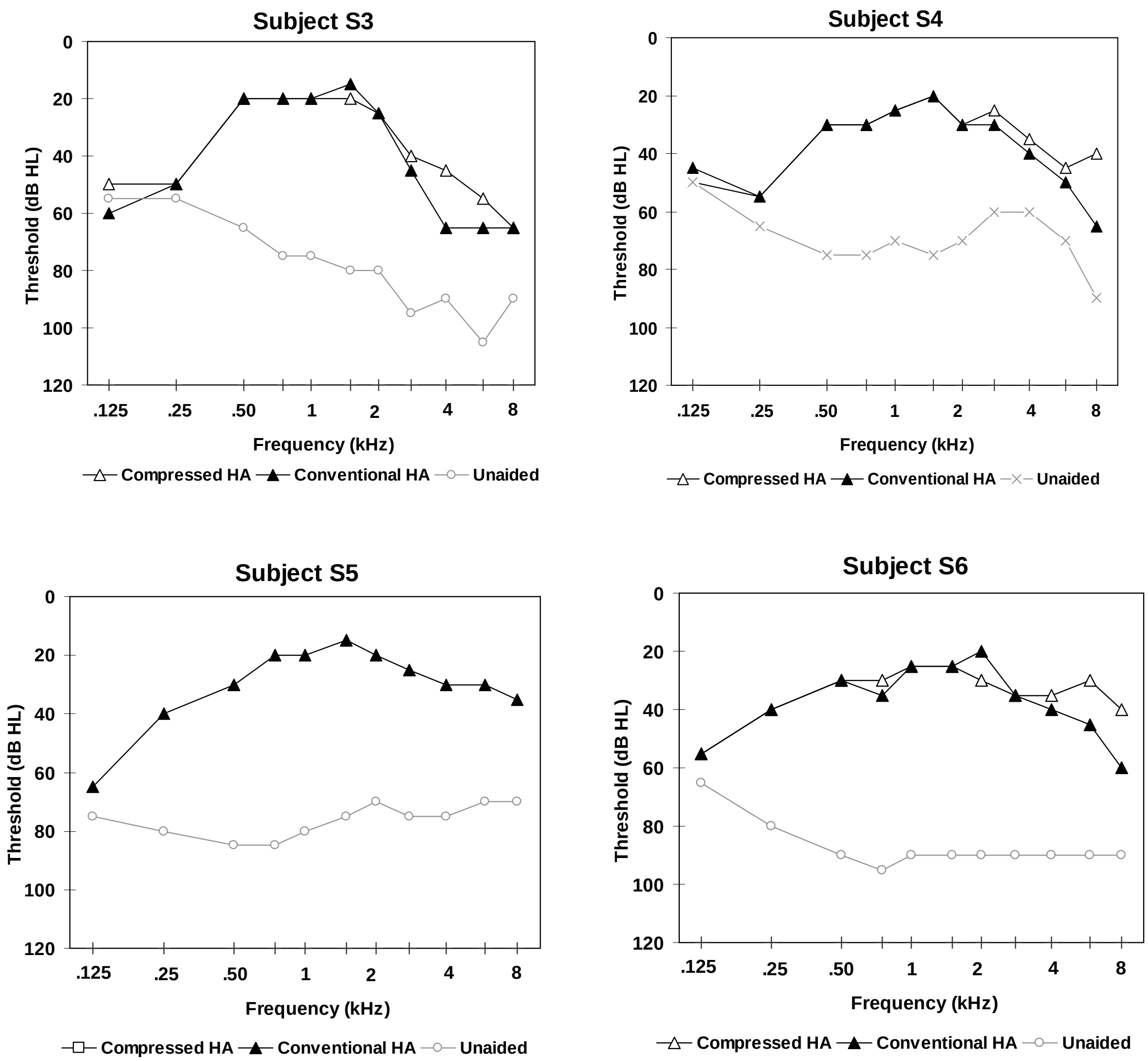


Figure C2. Aided sound field thresholds for subjects in experiment two using the frequency-compression hearing aid and the conventional hearing aid. Data were obtained using one-third octave narrow band noise.

Figure C2—continued

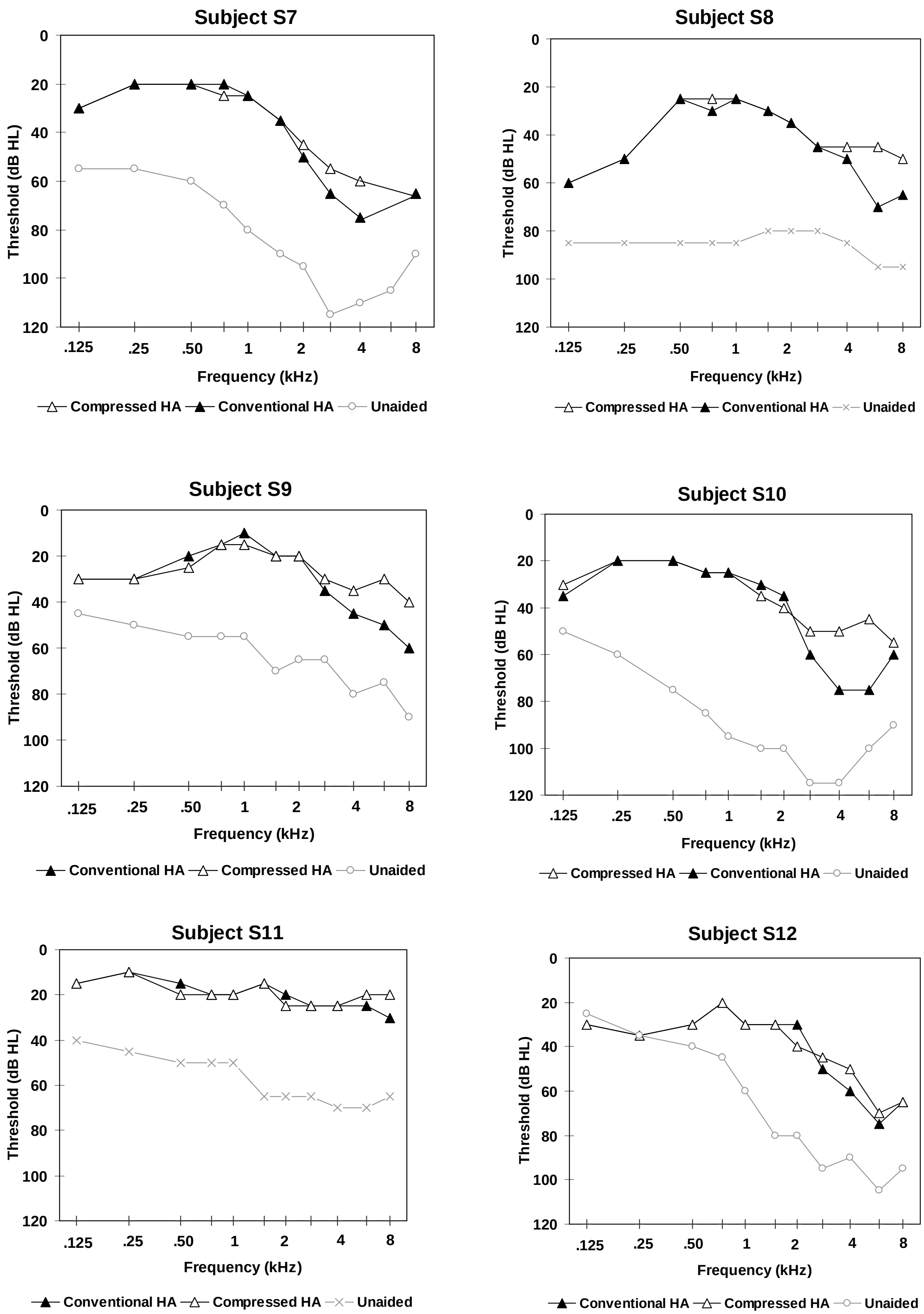
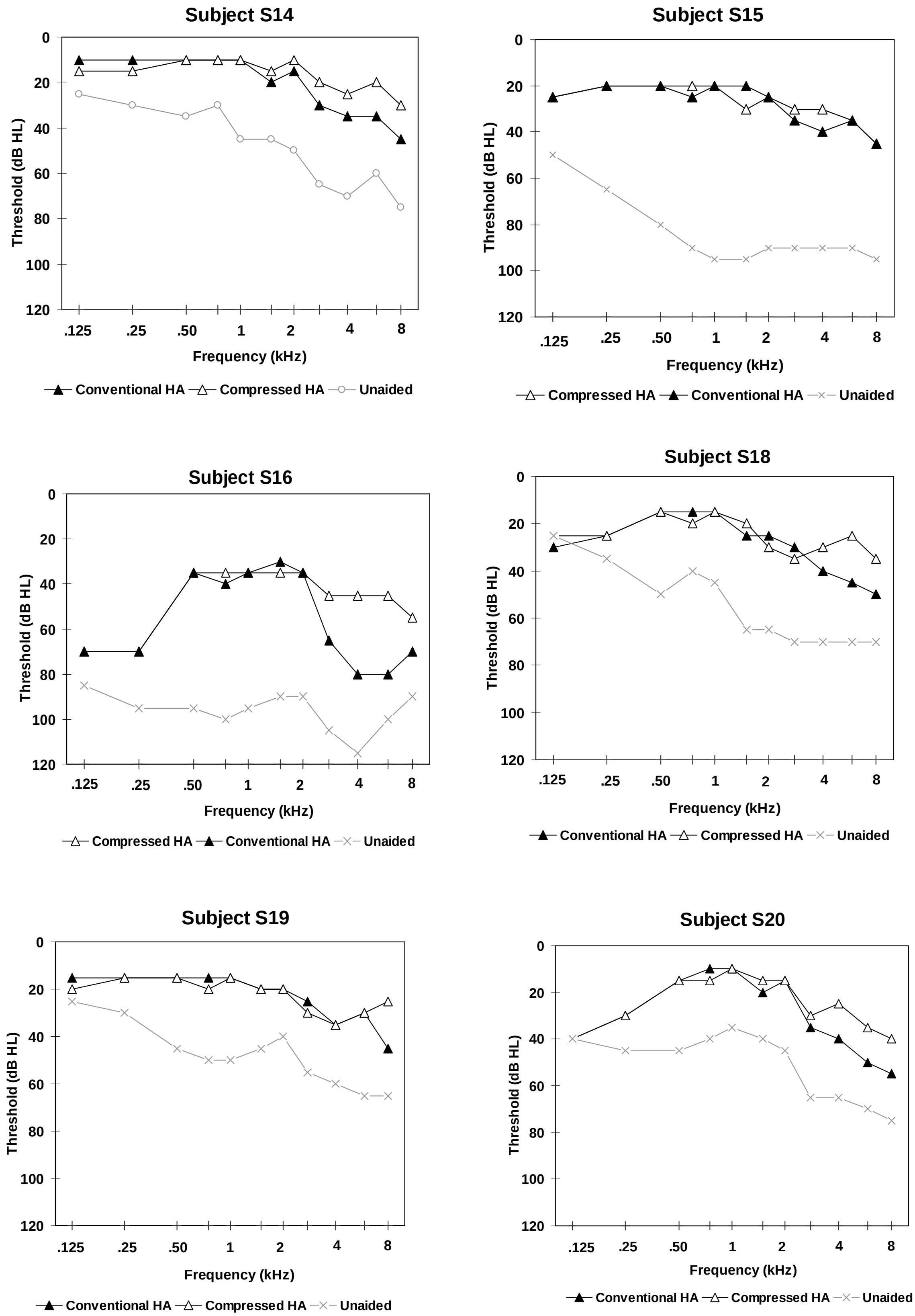


Figure C2—continued



APPENDIX D

INDIVIDUAL RESULTS

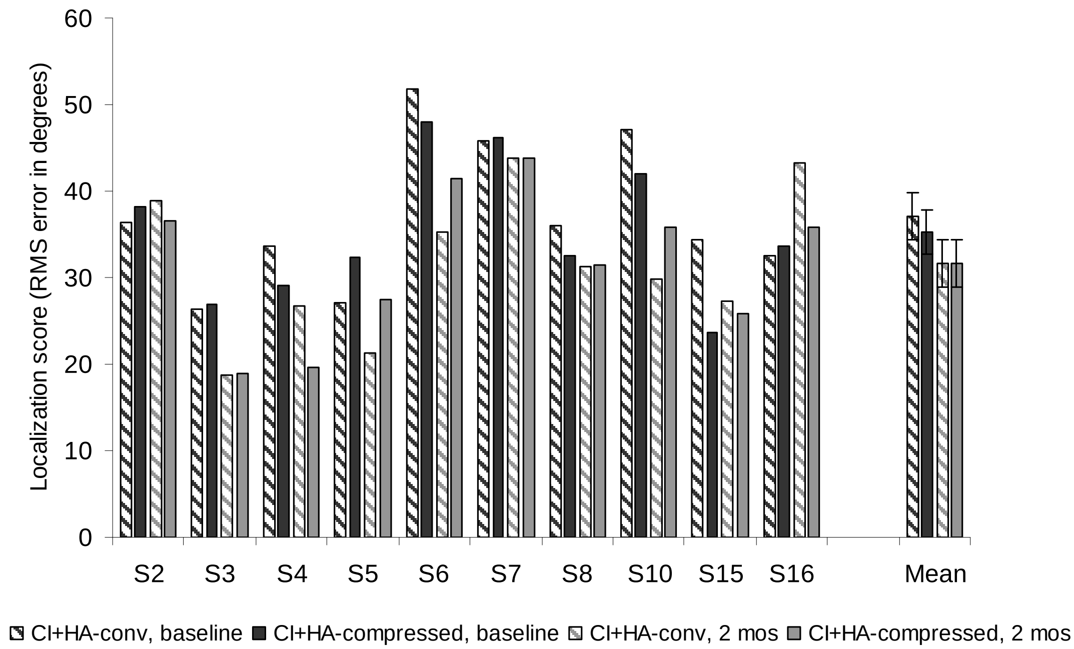


Figure D1. Individual and mean results from subjects in experiment one on the localization test at baseline and two months. The y-axis shows the localization score for each subject in RMS error. Lower scores on the graph indicate better localization ability. The striped columns show scores for the CI+HA-conventional and the solid columns show scores for the CI+HA-lowered. The black columns indicate baseline performance and the gray columns indicate performance after two months experience.

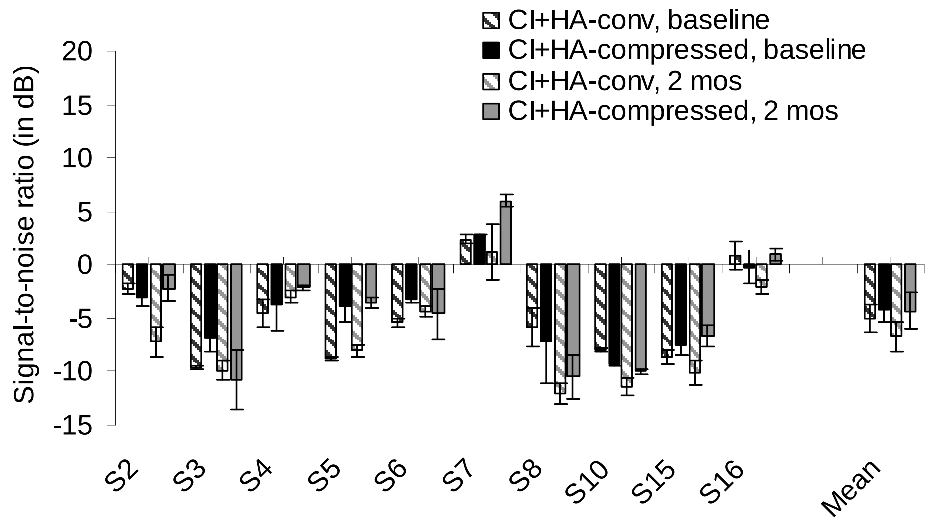


Figure D2. Individual and mean scores on the spondee-in-noise test for subjects in experiment one using the CI+HA-conventional and CI+HA-compressed conditions at baseline and two months. The striped columns show scores for the CI+HA-conventional and the solid columns show scores for the CI+HA-lowered. The black columns indicate baseline performance and the gray columns indicate performance after two months experience. Scores are shown along the y-axis in signal-to-noise ratio (dB) where more negative scores indicate better performance on the task. Error bars represent standard errors as calculated based on the mean of two of three runs.

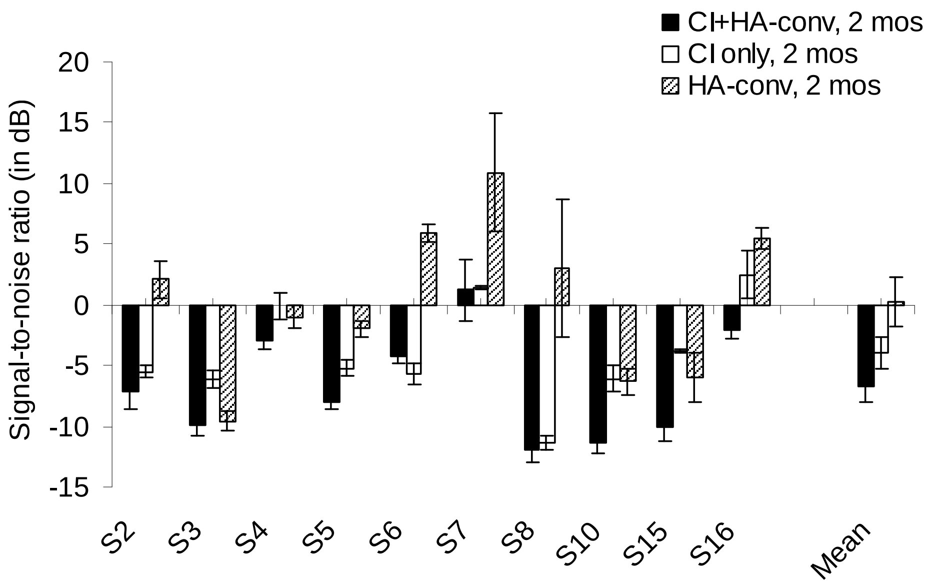


Figure D3. Individual and mean results on the spondee-in-noise test at two months for the conventional hearing aid and the CI. The CI+HA-conventional scores are shown by the black columns, the CI only by the white columns, and the HA-conventional by the striped columns. Scores are shown along the y-axis in signal-to-noise ratio (in dB); more negative scores indicate better performance. Error bars represent standard errors as calculated based on the mean of two of three runs.

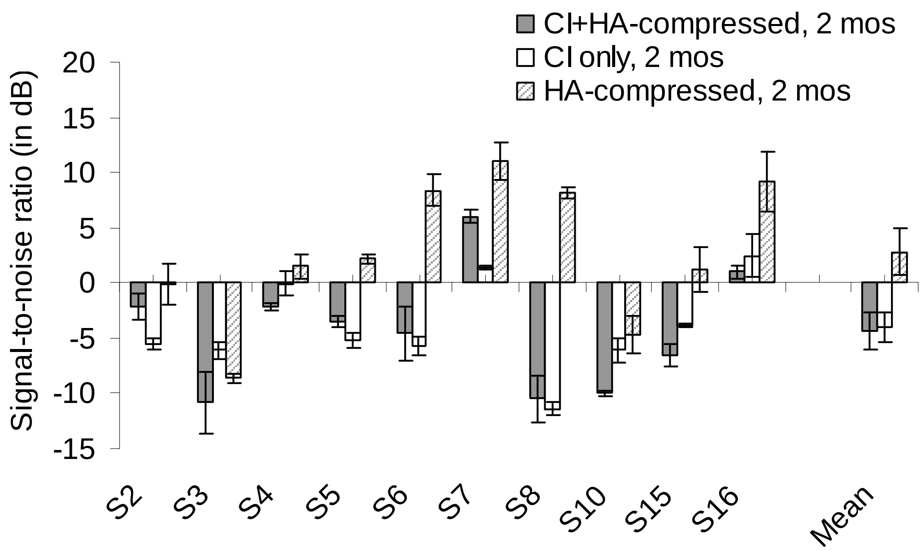


Figure D4. Individual and mean results on the spondee-in-noise test at two months for the frequency-compression hearing aid and the CI. The CI+HA-compressed scores are shown by the gray columns, the CI only by the white columns, and the HA-compressed by the striped columns. Scores are shown along the y-axis in signal-to-noise ratio (in dB); more negative scores indicate better performance. Error bars represent standard errors as calculated based on the mean of two of three runs.

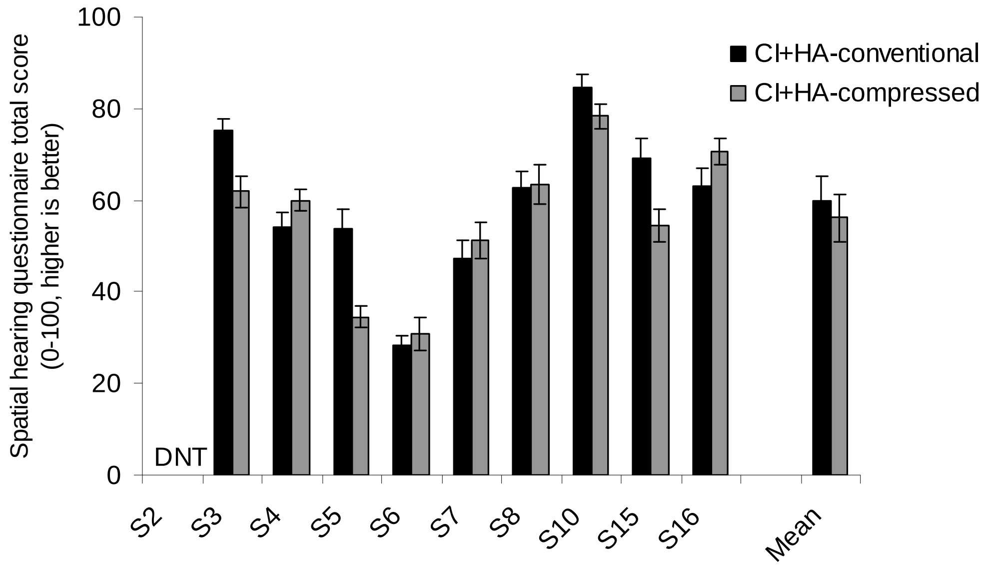


Figure D5. Spatial Hearing Questionnaire individual and mean total scores for the CI+HA-conventional and the CI+HA-compressed conditions. The CI+HA-conventional scores are shown by the black columns and the CI+HA-compressed scores are shown by the gray columns. Higher scores indicate better subjective spatial hearing ability. Error bars represent standard errors calculated from the mean of 24 questionnaire items.

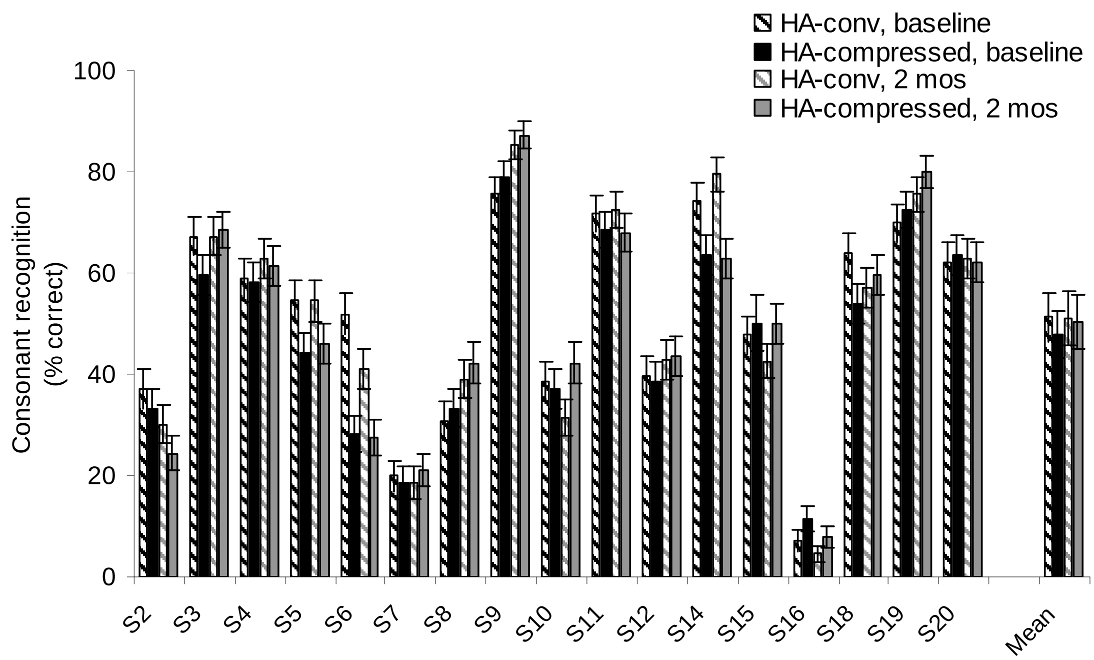


Figure D6. Individual and mean scores on the consonant recognition test at baseline and two months for the conventional and frequency-compression hearing aids. The HA-conventional scores are shown by the striped columns and the HA-compressed scores are shown in the solid columns. The black columns indicate baseline performance and the gray columns indicate performance after two months experience. Error bar represent standard errors as calculated based on the average of two lists of a male and female talker.

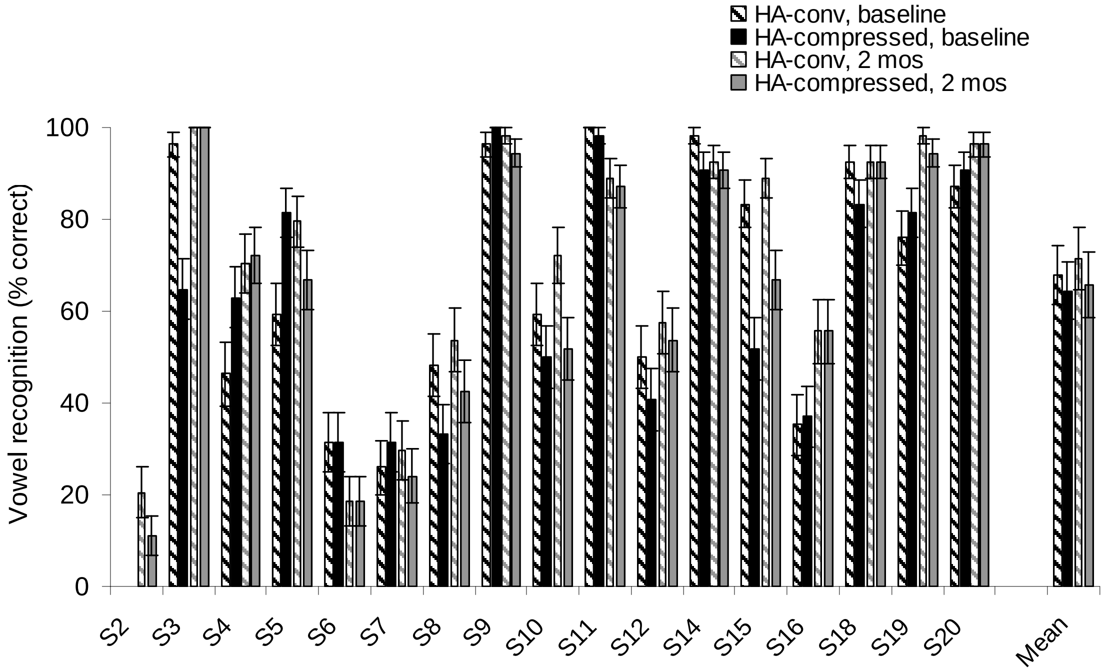


Figure D7. Individual and mean scores on the vowel recognition test at baseline and two months for the conventional and frequency-compression hearing aids. The HA-conventional scores are shown by the striped columns and the HA-compressed scores are shown in the solid columns. The black columns indicate baseline performance and the gray columns indicate performance after two months experience. Error bars represent standard errors.

REFERENCES

Alexander, J.M. (2009, November). *Current frequency lowering options in hearing aids.* Paper presented at the American Speech-Language and Hearing Association (ASHA) Convention, New Orleans, LA.

ANSI (1997). *American national standard methods for calculation of the speech intelligibility index* (ANSI S3.5-1997). New York: American National Standards Institute.

Armstrong, M., Pegg, P., James, C., & Blamey, P. (1997). Speech perception in noise with implant and hearing aid. *American Journal of Otology, 18(6)*(Supplement), S140-S141.

Bagatto, M., Moodie, S., Scollie, S., Seewald, R., & Moodie, S. K., et al. (2005). Clinical protocols for hearing impairment fitting in the desired sensation level method. *Trends in Amplification, 9*(4), 199-226.

Beasley, D. S., Mosher, N. L., & Orchik, D. J. (1976). Use of frequency shifted/time compressed speech with hearing-impaired children. *Audiology, 15*, 395-406.

Bennett, D. N., & Byers, V. W. (1967). Increased intelligibility in the hypacusic by slow-play frequency transposition. *J. Aud. Res., 7*, 107-118.

Bentler, R., Cole, B., Wu, Y-H. (2011, March). *Deriving an audibility index for frequency-lowered hearing aids.* Poster session presented at the meeting of the American Auditory Society, Scottsdale, AZ.

Biondi, E., & Biondi, L. (1968). The sampling of sounds as a new means of making speech intelligible to the profoundly deaf. *Alta. Frequenza, 37*, 180-191.

Blamey, P. J., & Saunders, E. (2008). A review of bimodal binaural hearing systems and fitting. *Acoustics Australia, 36*(3), 87-91.

Boothroyd, A., Hanin, L., & Hnath, T. (1985). A sentence test of speech perception: Reliability, set equivalence, and short term learning [Internal Report RCI 10]. *Speech and Hearing Science Report*, New York: City University of New York.

Braida, L. D., Durlach, N. I., Lippmann, R. P., Hicks, B. L., Rabinowitz, W. M., & Reed, C. M. (1979). Hearing aids--a review of past research on linear amplification, amplitude compression, and frequency lowering. *ASHA Monographs*, 19, 1-14.

Byrne, D., Dillon, H., Ching, T., Katsch, R., & Keidser, G. (2001). NAL-NL1 procedure for fitting nonlinear hearing aids: Characteristics and comparisons with other procedures. *Journal of the American Academy of Audiology, 12*(1), 37-51.

Carney, E., & Schlauch, R.S. (2007). Critical difference table for word recognition testing derived using computer simulation. *Journal of Speech, Language, and Hearing Research*, 50(5), 1203-1209.

Ching, T. Y. C., Hill, M., Brew, J., Incerti, P., Priolo, S., Rushbrook, E., et al. (2005). The effect of auditory experience on speech perception, localization, and functional performance of children who use a cochlear implant and a hearing aid in opposite ears. *International Journal of Audiology, 44*(12), 677-690.

Ching, T. Y., Incerti, P., & Hill, M. (2004). Binaural benefits for adults who use hearing aids and cochlear implants in opposite ears. *Ear & Hearing, 25*(1), 9-21.

Ching, T. Y., van Wanrooy, E., & Dillon, H. (2007). Binaural-bimodal fitting or bilateral implantation for managing severe to profound deafness: A review. *Trends in Amplification, 11*(3), 161-92.

Ching, T. Y. C., Psarros, C., Hill, M., Dillon, H., & Incerti, P. (2001). Should children who use cochlear implants wear hearing aids in the opposite ear? *Ear & Hearing, 22*(5), 365-380.

Chute, P. M., Gravel, J. S., & Popp, A. L. (1995). Speech perception abilities of adults using a multichannel cochlear implant and frequency transposition hearing aid. *The Annals of Otology, Rhinology Laryngology. 166* (Suppl), 260-263.

Cochlear Corporation. (1995). *Procedures Manual: Nucleus 22 Channel Cochlear Implant System*. Englewood, CO: Cochlear Corporation.

Cox, R. M. (1997). Administration and application of the APHAB. *Hearing Journal, 50*, 32-48.

Cox, R. M., & Alexander, G. C. (1995). The abbreviated profile of hearing aid benefit. *Ear & Hearing, 16*(2), 176-186.

Dillon, H., Keider, G., O'Brien, A., & Silberstein, H. (2003). Sound quality comparisons of advanced hearing aids. *Hearing Journal, 56*, 30-40.

Dorman, M. F., & Dahlstrom, L. (2004). Speech understanding by cochlear-implant patients with different left- and right-ear electrode arrays. *Ear & Hearing, 25*(2), 191-194.

Dunn, C. C., Tyler, R. S., & Witt, S. A. (2005). Benefit of wearing a hearing aid on the unimplanted ear in adult users of a cochlear implant. *Journal of Speech, Language, and Hearing Research, 48*(3), 668-680.

Feddersen, W. E., Sandel, T. T., Teas, D. C., & Jeffress, L. A. (1957). Localization of high-frequency tones. *Journal of the Acoustical Society of America, 29*, 988-991.

Firszt, J. B., Reeder, R. M., & Skinner, M. W. (2008). Restoring hearing symmetry with two cochlear implants or one cochlear implant and a contralateral hearing aid. *Journal of Rehabilitation Research and Development, 45*(5), 749-767.

Flynn, M. C., & Schmidtke, T. (2004). Benefits of bimodal stimulation for adults with a cochlear implant. *International Congress Series, 1273*, 227-230.

Foust, K. O., & Gengel, R. W. (1973). Speech discrimination by sensorineural hearing impaired persons using a transposer hearing aid. *Scand. Audiol., 2*, 161-170.

Francart, T., Brokx, J., & Wouters, J. (2008). Sensitivity to interaural level difference and loudness growth with bilateral bimodal stimulation. *Audiology & Neuro-Otology, 13*(5), 309-319.

Francart, T., Brokx, J., & Wouters, J. (2008). Sensitivity to interaural time differences with combined cochlear implant and acoustic stimulation. *Journal of the Association for Research in Otolaryngology, 10*(1), 131-141.

Gatehouse, S., & Noble, W. (2004). The speech, spatial and qualities of hearing scale (SSQ). *International Journal of Audiology, 43*(2), 85-99.

Gelfand, S. A. (2004). Binaural hearing. *Hearing: An introduction to psychological and physiological acoustics* (4th edition ed., pp. 369-400). New York: Marcel Dekker.

Gfeller, K. E., Olszewski, C., Turner, C., Gantz, B., & Oleson, J. (2006). Music perception with cochlear implants and residual hearing. *Audiology & Neurotology, 11*(Suppl 1), 12-15.

Gfeller, K., Turner, C., Oleson, J., Zhang, X., Gantz, B., Froman, R., et al. (2007). Accuracy of cochlear implant recipients on pitch perception, melody recognition, and speech reception in noise. *Ear & Hearing, 28*(3), 412-423.

Gifford, R. H., Dorman, M. F., Spahr, A. J., & McKarns, S. A. (2007). Effect of digital frequency compression (DFC) on speech recognition in candidates for combined electric and acoustic stimulation (EAS). *Journal of Speech, Language, and Hearing Research, 50*, 1194-1202.

Glasberg, B. R., & Moore, B. C. (1990). Derivation of auditory filter shapes from notched-noise data. *Hearing Research, 47*(1-2), 103-138.

Glista, D., Scollie, S., Bagatto, M., Seewald, R., Parsa, V., & Johnson, A. (2009). Evaluation of nonlinear frequency compression: Clinical outcomes. *International Journal of Audiology,* 1-13.

Grantham, D. W., Ashmead, D. H., Ricketts, T. A., Labadie, R. F., & Haynes, D. S. (2007). Horizontal-plane localization of noise and speech signals by postlingually deafened adults fitted with bilateral cochlear implants. *Ear & Hearing, 28*(4), 524-541.

Guttman, N., Levitt, H., & Bellefleur, P. A. (1970). Articulation training of the deaf using low-frequency surrogate fricatives. *J. Speech Hearing Res., 13*, 19-29.

Guttman, N., & Nelson, J. R. (1968). An instrument that creates some artificial speech spectra for the severely hard of hearing. *Am. Ann. Deaf, 113*, 295-302.

Haspiel, G. S. (1969). *Speech compression and frequency translation for hearing-impaired auditors* (Tech. Prog. Rep., Proj. HEW SRS, RD 2575-S)

Hawley, M. L., Litovsky, R. Y., & Colburn, H. S. (1999). Speech intelligibility and localization in a multi-source environment. *Journal of the Acoustical Society of America, 105*(6), 3436-3448.

Johansson, B. (1961). A new coding amplifier system for the severely hard of hearing. Stutgart. 655-657.

Johansson, B. (1966). The use of the transposer for the management of the deaf child. *International Audiology, 5*, 362-373.

Johansson, B., & Sjogren, H. (1965). The effect on normal hearing subjects and profoundly deaf children by coding voiceless consonants using frequency transposition. Liege. 1-4.

Kileny, P. R., Snorrason, R. M., Zwolan, T. A., Macpherson, E. A., & Middlebrooks, J. C. (2004). Contralateral hearing aid benefit in patients with cochlear implants. *International Congress Series, 1273*, 231-234.

Kluk, K., & Moore, B. C. J. (2005). Factors affecting psychophysical tuning curves for hearing-impaired subjects with high-frequency dead regions. *Hear Res, 200*, 115-131.

Kong, Y., Stickney, G. S., & Zeng, F. (2005). Speech and melody recognition in binaurally combined acoustic and electric hearing. *Journal of the Acoustic Society of America, 117*(3), 1351-1361.

Kuk, F., Keenan, D., Korhonen, P., & Lau, C. C. (2009). Efficacy of linear frequency transposition on consonant identification in quiet and in noise. *Journal of the American Academy of Audiology, 20*(8), 465-479.

Kuk, F., Peeters, H., Keenan, D., Jessen, A., & Andersen, H. (2006). Linear frequency transposition: Extending the audibility of high-frequency information. *Hearing Review, 13*, 44-46.

Kuk, F., Peeters, H., Keenan, D., & Lau, C. (2007). Use of frequency transposition in a thin-tube open-ear fitting. *Hearing Journal, 60*(4), 59-63.

Laback, B., Pok, S., Baumgartner, W., Deutsch, W. A., & Schmid, K. (2004). Sensitivity to interaural level and envelope time differences of two bilateral cochlear implant listeners using clinical sound processors. *Ear & Hearing, 25*(5), 488-500.

Laszig, R., Aschendorff, A., Stecker, M., Müller-Deile, J., Maune, S., Dillier, N., et al. (2004). Benefits of bilateral electrical stimulation with the nucleus cochlear implant in adults: 6-month postoperative results. *Otology & Neurotology, 25*(6), 958-968.

Lawson, D. T., Wilson, B. S., Zerbi, M., van den Honert, C., Finley, C. C., Farmer, J. C., Jr, et al. (1998). Bilateral cochlear implants controlled by a single speech processor. *The American Journal of Otology, 19*(6), 758-761.

Ling, D. (1968). Three experiments on frequency transposition. *Am. Ann. Deaf, 113*, 283-294.

Ling, D., & Druz, W. S. (1967). Transposition of high-frequency sounds by partial vocoding of the speech spectrum: Its use by deaf children. *J. Aud. Res., 7*, 133-144.

Litovsky, R., Parkinson, A., Arcaroli, J., & Sammeth, C. (2006). Simultaneous bilateral cochlear implantation in adults: A multicenter clinical study. *Ear & Hearing, 27*(6), 714-731.

Long, C. J., Eddington, D. K., Colburn, H. S., & Rabinowitz, W. M. (2003). Binaural sensitivity as a function of interaural electrode position with a bilateral cochlear implant user. *Journal of the Acoustic Society of America, 114*(3), 1565-1574.

Luntz, M., Shpak, T., & Weiss, H. (2005). Binaural-bimodal hearing: Concomitant use of a unilateral cochlear implant and a contralateral hearing aid. *Acta Oto-Laryngologica, 125*(8), 863-869.

MacArdle, B. M., West, C., Bradley, J., Worth, S., Mackenzie, J., & Bellman, S. C. (2001). A study of the application of a frequency transposition hearing system in children. *British Journal of Audiology, 35*(1), 17-29.

Maillet, C. J., Tyler, R. S., & Jordan, H. N. (1995). Change in the quality of life of adult cochlear implant patients. *The Annals of Otology, Rhinology & Laryngology.Supplement, 165*, 31-48.

McDermott, H. J., & Dean, M. R. (2000). Speech perception with steeply sloping hearing loss: Effects of frequency transposition. *British Journal of Audiology, 34*(6), 353-361.

McDermott, H. J., Dorkos, V. P., Dean, M. R., & Ching, T. Y. (1999). Improvements in speech perception with use of the AVR TranSonic frequency-transposing hearing aid. *Journal of Speech, Language, and Hearing Research, 42*(6), 1323-1335.

McDermott, H., & Henshall, K. (2010). The use of frequency compression by cochlear implant recipients with postoperative acoustic hearing. *Journal of the American Academy of Audiology, 21*(6), 380-389.

McDermott, H., & Knight, M. (2001). Preliminary results with the AVR ImpaCt frequency-transposing hearing aid. *Journal of the American Academy of Audiology, 12*(3), 121-127.

Middlebrooks, J. C., & Green, D. M. (1991). Sound localization by human listeners. *Annual Review of Psychology, 42*, 135-159.

Miller-Hansen, D. R., Nelson, P. B., Widen, J. E., & Simon, S. D. (2003). Evaluating the benefit of speech recoding hearing aids in children. *American Journal of Audiology, 12*(2), 106-113.

Mok, M., Grayden, D., Dowell, R. C., & Lawrence, D. (2006). Speech perception for adults who use hearing aids in conjunction with cochlear implants in opposite ears. *Journal of Speech, Language, and Hearing Research, 49*(2), 338-351.

Moore, B. C. (2003). Coding of sounds in the auditory system and its relevance to signal processing and coding in cochlear implants. *Otology & Neurotology, 24*(2), 243-254.

Moore, B. C. J., Glasberg, B. R., & Stone, M. A. (2004). New version of the TEN test with calibrations in dB HL. *Ear & Hearing, 25*, 478-487.

Morera, C., Manrique, M., Ramos, A., Garcia-Ibanez, L., Cavalle, L., Huarte, A., et al. (2005). Advantages of binaural hearing provided through bimodal stimulation via a cochlear implant and a conventional hearing aid: A 6-month comparative study. *Acta Oto-Laryngologica, 125*, 596-606.

Musa-Shufani, S., Walger, M., von Wedel, H., & Meister, H. (2006). Influence of dynamic compression on directional hearing in the horizontal plane. *Ear and Hearing, 27*(3), 279-285.

Nyffeler, M. (2008). Study finds that non-linear frequency compression boosts speech intelligibility. *The Hearing Journal, 61*(12), 22-26.

O'Brien, A., Yeend, I., Hartley, L., Keidser, G., & Nyffeler, M. (2010). Evaluation of frequency compression and high-frequency directionality. *The Hearing Journal, 63*(8), 32-37.

Oeken, F. W. (1963). Can hard of hearing patients suffering from high-tone perceptive deafness be improved by frequency transposition? *J. Internat. Audiol., 2,* 263-266.

Parent, T. C., Chmiel, R., & Jerger, J. (1998). Comparison of performance with frequency transposition hearing aids and conventional hearing aids. *Journal of the American Academy of Audiology, 9*(1), 67-77.

Parkinson, A. J., Arcaroli, J., Staller, S. J., Arndt, P. L., Cosgriff, A., & Ebinger, K. (2002). The nucleus 24 contour cochlear implant system: Adult clinical trial results. *Ear & Hearing, 23*(1 Suppl), 41S-48S.

Perreau, A.E., Tyler, R.S., Witt, S., & Dunn, C. (2007). Selection strategies for binaural and monaural cochlear implantation. *American Journal of Audiology, 16,* 85-93.

Peterson, G.E. & Lehiste, I. (1962). Revised CNC lists for auditory tests. *Journal of Speech and Hearing Disorders, 27,* 62-70.

Potts, L. G., Skinner, M. W., Litovsky, R. A., Strube, M. J., & Kuk, F. (2009). Recognition and localization of speech by adult cochlear implant recipients wearing a digital hearing aid in the nonimplanted ear (bimodal hearing). *Journal of the American Academy of Audiology, 20*(6), 353-373.

Redden, R. B. (1973). The effects of frequency compression and transposition on the intelligibility of monosyllabic words in normal hearing listeners. Unpublished E.D. thesis, Boston University.

Robinson, J. D., Baer, T., & Moore, B. C. J. (2007). Using transposition to improve consonant discrimination and detection for listeners with severe high-frequency hearing loss. *International Journal of Audiology, 46,* 293-308.

Robinson, J. D., Stainsby, T. H., Baer, T., & Moore, B. C. J. (2009). Evaluation of a frequency transposition algorithm using wearable hearing aids. *International Journal of Audiology, 48,* 384-393.

Rosen, S. (1992). Temporal information in speech and its relevance for cochlear implants. *Philosophical Transactions of the Royal Society of London, 336,* 367.

Rosenhouse, J. (1990). A new transposition device for the deaf. *Hearing Journal, 43,* 20-25.

Scollie, S., Seewald, R., Cornelisse, L., Moodie, S., & Bagatto, M. P., et al. (2005). The desired sensation level multistage input/output algorithm. *Trends in Amplification, 9*(4), 159-197.

Seeber, B. U., Baumann, U., & Fastl, H. (2004). Localization ability with bimodal hearing aids and bilateral cochlear implants. *The Journal of the Acoustical Society of America, 116*(3), 1698-1709.

Seeber, B. U., & Fastl, H. (2008). Localization cues with bilateral cochlear implants. *The Journal of the Acoustical Society of America, 123*(2), 1030-1042.

Sek, A., Alcantara, J., Moore, B. C., Kluk, K., & Wicher, A. (2005). Development of a fast method for determining psychophysical tuning curves. *International Journal of Audiology, 44*(7), 408-420.

Senn, P., Kompis, M., Vischer, M., & Haeusler, R. (2005). Minimum audible angle, just noticeable interaural differences and speech intelligibility with bilateral cochlear implants using clinical speech processors. *Audiology & Neurotology, 10*(6), 342-352.

Shallop, J. K., Arndt, P. L., & Turnacliff, K. A. (1992). Expanded indications for cochlear implantation: Perceptual results in seven adults with residual hearing. *J Speech Lang Pathol Audiol, 16*, 141-148.

Shaw, E. A. C. (1974). The external ear. In W. D. Keidel, & W. D. Neff (Eds.), *Handbook of sensory physiology* (pp. 450-490). New York: Springer-Verlag.

Simpson, A. (2009). Frequency-lowering devices for managing high-frequency hearing loss: A review. *Trends in Amplification, 13*(2), 87-106.

Simpson, A., Hersbach, A. A., & McDermott, H. J. (2006). Frequency-compression outcomes in listeners with steeply sloping audiograms. *International Journal of Audiology, 45*(11), 619-629.

Simpson, A., Hersbach, A., & McDermott, H. (2005). Improvements in speech perception with an experimental nonlinear frequency compression hearing device. *International Journal of Audiology, 44*(5), 281-292.

Spahr, A. J., & Dorman, M. F. (2004). Performance of subjects fit with the Advanced Bionics CII and Nucleus 3G cochlear implant devices. *Archives of Otolaryngology—Head and Neck Surgery, 130*, 624-628.

Summerfield, A. Q., Barton, G. R., Toner, J., McAnallen, C., Proops, D., Harries, C., et al. (2006). Self-reported benefits from successive bilateral cochlear implantation in post-lingually deafened adults: Randomised controlled trial. *International Journal of Audiology, 45 Suppl 1*, S99-S107.

Takefuta, Y., & Swigart, E. (1968). Intelligibility of speech signals spectrally compressed by a sampling-synthesizer technique. *IEEE Trans. Aud. Electroacoust., AU-16*, 271-274.

Thornton, A. R., & Abbas, P. J. (1980). Low-frequency hearing loss: Perception of filtered speech, psychophysical tuning curves, and masking. *Journal of the Acoustical Society of America, 67*, 638-643.

Turner, C. W., Gantz, B. J., Vidal, C., Behrens, A., & Henry, B. A. (2004). Speech recognition in noise for cochlear implant listeners: Benefits of residual acoustic hearing. *Journal of the Acoustic Society of America, 115*(4), 1729-1735.

Turner, C. W., & Hurtig, R. H. (1999). Proportional frequency compression of speech for listeners with sensorineural hearing loss. *The Journal of the Acoustical Society of America, 106*(2), 877-886.

Tyler, R. S., Dunn, C. C., Witt, S. A., & Noble, W. G. (2007). Speech perception and localization with adults with bilateral sequential cochlear implants. *Ear & Hearing, 28*(2 Suppl), 86S-90S.

Tyler, R. S., Dunn, C. C., Witt, S. A., Noble, W., Gantz, B. J., Rubinstein, J. T., et al. (2006). Soundfield hearing for patients with cochlear implants and hearing aids. In H. R. Cooper, & L. C. Craddock (Eds.), *Cochlear implants: A practical guide* (2nd ed., pp. 338-365; 15) Whurr Publishers, Ltd.

Tyler, R. S., Lowder, M. W., Otto, S. R., Preece, J. P., Gantz, B. J., & McCabe, B. F. (1984). Initial iowa results with the multichannel cochlear implant from melbourne. *Journal of Speech and Hearing Research, 27*(4), 596-604.

Tyler, R. S., Lowder, M. W., Parkinson, A. J., Woodworth, G. G., & Gantz, B. J. (1995). Performance of adult ineraid and nucleus cochlear implant patients after 3.5 years of use. *Audiology, 34*(3), 135-144.

Tyler, R. S., Parkinson, A. J., Wilson, B. S., Witt, S., Preece, J. P., & Noble, W. (2002). Patient utilizing a hearing aid and a cochlear implant: Speech perception and localization. *Ear & Hearing, 23*(2), 98-105.

Tyler, R., Perreau, A., & Ji, H. (2009). Validation of the spatial hearing questionnaire. *Ear & Hearing, 30*(4), 466-475.

van Hoesel, R. J. M., & Tyler, R. S. (2003). Speech perception, localization, and lateralization with bilateral cochlear implants. *Journal of the Acoustic Society of America*, *113*(3), 1617-1630.

Van Hoesel, R., Ramsden, R., & Odriscoll, M. (2002). Sound-direction identification, interaural time delay discrimination, and speech intelligibility advantages in noise for a bilateral cochlear implant user. *Ear & Hearing, 23*(2), 137-149.

Velmans, M. (1971). Aids for deaf persons. U.K. Patent 1340105.

Velmans, M. (1973). Speech imitation in simulated deafness, using visual cues and recoded auditory information. *Language and Speech, 16*, 224-236.

Velmans, M. (1974). The design of speech recoding devices for the deaf. *Brit. J. Audiol., 8*, 1-5.

Velmans, M. (1975). Effects of frequency recoding on the articulation learning of perceptively deaf children. *Language and Speech, 18*, 180-193.

Vermeire, K., Brokx, J. P., Wuyts, F. L., Cochet, E., Hofkens, A., & Van de Heyning, P. H. (2005). Quality-of-life benefit from cochlear implantation in the elderly. *Otology & Neurotology, 26*(2), 188-195.

Vickers, D., Robinson, J. D., Fullgrabe, C., Baer, T., & Moore, B. C. J. (2009). Relative importance of different spectral bands to consonant identification: Relevance for frequency transposition in hearing aids. *Internation Journal of Audiology, 48*, 334-345.

Waltzman, S. B., Cohen, N. L., & Roland, T. J., Jr. (1999). A comparison of the growth of open-set speech perception between the nucleus 22 and nucleus 24 cochlear implant systems. *The American Journal of Otology, 20*(4), 435-441.

Waltzman, S. B., Cohen, N. L., & Shapiro, W. H. (1992). Sensory aids in conjunction with cochlear implants. *The American Journal of Otology, 13*(4), 308-312.

Wedenberg, E. (1959). Auditory training of the severely hard of hearing using a coding amplifier. *Proceedings of the 3rd International Congress on Acoustics, Stuttgart,* , 658-660.

Yost, W. A., & Dye, R. H., Jr. (1988). Discrimination of interaural differences of level as a function of frequency. *The Journal of the Acoustical Society of America, 83*(5), 1846-1851.

Zemlin, W. R. (1966). The use of bandwidth and time compression for the hearing handicapped. Louisville. 95-97.

Zurek, P. M. (1993). Binaural advantages and directional effects in speech intelligibility. In G. A. Studebaker, & I. Hochberg (Eds.), *Acoustical factors affecting hearing aid performance* (2nd ed., pp. 255-276; 15). Needham Heights, MA.: Allyn and Bacon.