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Manipulation of Auditory Feedback in Individuals with Normal Hearing and Hearing Loss

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Abstract

Auditory feedback, the hearing of one's own voice, plays an important role in the detection of speech errors and the regulation of speech production. The limited auditory cues available with a hearing loss can reduce the ability of individuals with hearing loss to use their auditory feedback. Hearing aids are a common assistive device that amplifies inaudible sounds. Hearing aids can also change auditory feedback through digital signal processing, such as frequency lowering. Frequency lowering moves high frequency information of an incoming auditory stimulus into a lower frequency region where audibility may be better. This can change how speech sounds are perceived. For example, the high frequency information of /s/ is moved closer to the lower frequency area of /ʃ/. As well, real-time signal processing in a laboratory setting can also manipulate various aspects of speech cues, such as intensity and vowel formants. These changes in auditory feedback may result in changes in speech production as the speech motor control system may perceive these perturbations as speech errors. A series of experiments were carried out to examine changes in speech production as a result of perturbations in the auditory feedback in individuals with normal hearing and hearing loss. Intensity and vowel formant perturbations were conducted using real-time signal processing in the laboratory. As well, changes in speech production were measured using auditory feedback that was processed with frequency lowering technology in hearing aids. Acoustic characteristics of intensity of vowels, sibilant fricatives, and first and second formants were analyzed. The results showed that the speech motor control system is sensitive to changes in auditory feedback because perturbations in auditory feedback can result in changes in speech production. However, speech production is not completely controlled by auditory feedback and other feedback systems, such as the somatosensory system, are also involved. An impairment of the auditory system can reduce the ability of the speech motor control system to use auditory feedback in the detection of speech errors, even when aided with hearing aids. Effects of frequency lowering in hearing aids on speech production depend on the parameters used and acclimatization time.

Keywords

Auditory feedback, speech production, hearing aids, hearing loss, aging, frequency lowering, non-linear frequency compression, vowels, fricatives, intensity, formants, aging

Co-Authorship Statement

I, Le Truc Linh Vaccarello, am responsible for the research studies presented in this dissertation. I was responsible for the design of the studies, data collection, statistical analyses, results interpretation and the writing for all the chapters. Below is a list of the co-authors and their contributions for the chapters.

Chapter 1: I was the sole author of this chapter. David Purcell and Susan Scollie reviewed the chapter.

Chapter 2. Linh Vaccarello, Takashi Mitsuya, Susan Scollie, and David Purcell
Takashi Mitsuya helped with the design of the project. Susan Scollie reviewed the chapter. David Purcell was involved with the study design, provided the computer algorithms for the study, supported data analyses, interpretation of the results and reviewed the chapter.

Chapter 3. Linh Vaccarello, Susan Scollie, and David Purcell
Susan Scollie supported study design, data analyses and reviewed the chapter. David Purcell was involved with the study design, provided the computer algorithms for the study, supported data analyses, interpretation of the results and reviewed the chapter.

Chapter 4. Linh Vaccarello, Scott Adams, Susan Scollie, and David Purcell
David Purcell provided the computer algorithms for the study, supported data analyses, interpretation of the results and reviewed the chapter. Scott Adams provided support for data analyses, interpretation of results and reviewed the chapter. Susan Scollie supported data analyses and reviewed the chapter.

Chapter 5. Linh Vaccarello, Danielle Glista, David Purcell, Vijay Parsa, and Susan Scollie
Danielle Glista supported data interpretation. David Purcell provided support for the interpretation of results and reviewed the chapter. Vijay Parsa provided the MATLAB code for extraction of fricative values. Susan Scollie provided support for study design, data analyses, interpretation of results and reviewed the chapter.

Chapter 6. Linh Vaccarello, Danielle Glista, Takashi Mitsuya, Marianne Hawkins, David Purcell, Vijay Parsa, and Susan Scollie

Danielle Glista provided support for study design, data analyses and interpretation of results. Takashi Mitsuya provided support for data analyses. Marianne Hawkins provided support for data collection. David Purcell provided support for interpretation of results and reviewed the chapter. Vijay Parsa provided the MATLAB code for extraction of fricative values. Susan Scollie provided support study design, data analyses, interpretation of results, and reviewed the chapter.

Chapter 7: I was the sole author of this chapter. David Purcell and Susan Scollie reviewed the chapter.

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List of Acronyms

AC	Air conduction
BC	Bone conduction
BTE	Behind the ear
CI	Confidence interval
CNE	Could not extract
CR	Compression ratio
CT	Cut-off frequency for SoundRecover1
CT1	Cut-off frequency 1 for SoundRecover2
CT2	Cut-off frequency 2 for SoundRecover2
d-index	Dissimilar index
DAI	Direct audio input
DSL	Desired Sensation Level
dB	Decibel
dBA	Decibel (weighted A)
F0	Fundamental frequency
F1	First formant
F2	Second formant
F3	Third formant
F4	Fourth formant
HA	Hearing aid
HF-PTA	High frequency pure tone average
HL	Hearing level
Hz	Hertz
kHz	Kilohertz
MAOF	Maximum audible output frequency
n	Sample size
NLFC	Non-linear frequency compression
PTA	Pure tone average
RMS	Root-mean-square
s	Seconds
SD	Standard deviation
SPL	Sound pressure level
SR1	SoundRecover1
SR2	SoundRecover2
TEN	Threshold equalizing noise
ms	Milliseconds
µs	Microseconds
vs	Versus
VSA	Vowel space area
yrs	Years

Chapter 1

1 Introduction

Human beings are social animals that use communication to start and maintain relationships with one another. Communication can range from nonverbal cues such as body language and touch as well as verbal cues such as spoken language. A major component of spoken language is fluency, which is defined as the ability of the talker to express themselves easily and accurately (Fluent [Def. 1], 2017). Accuracy in speech production is important as it plays an important role in conveying the message (Shannon, 1948). To achieve accuracy, there needs to be minimization of errors in speech production. This implies that speech production involves targets and the speech system is trying to maintain speech sounds to fit within the targets. Such targets have usually been defined in acoustic terms, such as characterizing vowels by formants (Hillenbrand *et al.*, 1995; Ozbic & Kogovsek, 2010) and fricatives by the frequency characteristics of the spectral noise (Forrest *et al.*, 1988).

The speech control system uses two types of systems to maintain and regulate speech production: feedforward and feedback, as described in the State Feedback Control model in Houde and Nagarajan (2011) and the Directions into Velocities of Articulators model in Tourville and Guenther (2011). The feedforward system involves the motor cortex providing commands to the vocal tract and articulators. In contrast, the feedback speech control system compares incoming auditory or somatosensory speech signals to target or predicted speech sounds within cortical areas (Houde & Nagarajan, 2011; Tourville & Guenther, 2011). If there is a difference between the incoming speech signals and the target speech sounds, the speech control system determines there is an error in speech production and adjusts the motor commands sent to the articulators to correct the speech sound. With persistent errors, the internal target speech sound is updated. The feedforward and feedback systems work together to detect errors in speech and maintain accurate speech productions. An impairment in one system may lead to poor interactions between the systems or poor performance of the feedback system and a reduction in speech accuracy. For example, postlingual deafened individuals usually continue to produce intelligible speech for years following their hearing loss due to the feedforward commands they acquired while they could hear. (Cowie & Douglas-Cowie, 1983; Menard *et al.*, 2007). However, the reduction in auditory feedback does cause a degradation of their speech and the degree of degradation varies from one individual with hearing loss to another (Langereis *et al.*, 1997; Menard *et al.*, 2007).

Past studies have manipulated somatosensory and auditory feedback to measure changes in speech production. This type of perturbation experiment allows investigators to examine the talker's corrective behaviour to study how the speech target is defined by their somatosensory and auditory targets and how the system controls speech. For example, a study by Tremblay, Shiller and Ostry (2003) showed that talkers changed the position of their jaw when their jaw was pulled forward during talking. Other somatosensory perturbation studies have shown that changes to the positions of articulators when speaking will result in compensatory positional change of the articulators (Folkins & Abbs, 1975; Folkins & Zimmermann, 1982; Shaiman, 1989). Similarly, changes to auditory feedback will result in compensation by the talker. For example, Mitsuya and colleagues (2015) had participants repeatedly say a targeted vowel in /hVd/ context and increased or decreased the first formant (F1). The results showed that talkers changed their production of the targeted vowel by compensating in the opposite direction of the perturbation. Other studies have manipulated the second formant of vowels (MacDonald, Goldberg, & Munhall, 2010; MacDonald, Purcell, & Munhall, 2011; Munhall *et al.*, 2009; Villacorta, Perkell & Guenther, 2007), fundamental frequency (Burnett *et al.*, 1998; Jones & Munhall, 2000), intensity (Bauer *et al.*, 2006; Heinks-Maldonado & Houde, 2005; Larson, Sun & Hain, 2007) and spectral noises of fricatives (Casserly, 2011, Shiller *et al.*, 2009).

The accuracy of speech production is maintained by an intricate system of various feedforward and feedback systems. Perturbed auditory feedback studies elicit speech compensation that is proportional to the perturbation. Talkers in these studies compensate approximately 25%-50% of the formant manipulation (Houde & Jordan, 1998; Liu & Larson, 2007; MacDonald *et al.*, 2010; Munhall *et al.*, 2009; Purcell & Munhall, 2006; Villacorta *et al.*, 2007). This partial compensation shows that speech is controlled by the interaction of various systems, such as the feedforward, somatosensory feedback and auditory feedback systems (Nasir & Ostry, 2008; Tremblay *et al.*, 2003). Partial compensation may occur for various reasons. The physical constraints of the articulators are a possible reason that may prevent complete compensation. For example, speech compensation for /i/ in the positive direction would not be possible as the tongue would need to go higher than the palate (MacDonald *et al.*, 2010). For each vowel, the speech motor control system may balance the weighting of somatosensory feedback and auditory feedback differently as each vowel has different articulator positions. For example, regulating the production of the vowel /i/ may rely more on somatosensory feedback than auditory feedback. As well, partial compensation may occur because of the perturbation magnitudes used (MacDonald *et al.*, 2010). If the perturbation magnitude is too large, the speech motor control system may ignore the auditory feedback and characterize it as unrealistic (MacDonald *et al.*, 2010). The speech motor control system may also attribute the large perturbations were due to other sources like the environment (MacDonald *et al.*, 2010). Thus, in altered auditory feedback studies, partial compensations to the perturbations are expected.

The speech motor control system does not always require conscious mental effort to regulate speech production. Formant compensation in altered auditory feedback studies has been found to occur automatically and unconsciously. A study by Munhall *et al.* (2009) compared speech compensation patterns between three groups of talkers that had different instructions. The different instructions were: (1) control: the talkers received no information about the experiment and were naïve to the feedback, (2) ignore headphones: the talkers were told about the changes in auditory feedback that would occur in the headphones and were told to ignore the auditory feedback, and (3) avoid compensation: the talkers were told of the manipulation and were told to maintain regular speech production without compensating. All three groups compensated in the opposite direction to the formant perturbation and there were no significant differences in the compensation magnitudes between the groups. Similarly, Houde and Jordan (2002) conducted post-experiment interviews and found that their participants did not notice the feedback manipulation and did not know they were changing their speech throughout the experiment. This suggested that conscious strategies were not used to compensate for the manipulated feedback.

There are other ways to change the incoming auditory feedback signal in talkers, such as digital signal processing in hearing aids. Hearing aids manipulate auditory sounds to increase audibility in individuals with hearing loss. For example, hearing aids can reduce background sounds in the environment and enhance speech signals based on microphone configurations or adaptive noise reduction algorithms (Dillon, 2012). Another digital signal processor that can change the incoming sound is frequency lowering technology. Frequency lowering technology is used by clinicians to improve audibility for high frequency sounds by moving high frequency sounds to a lower frequency range where audibility is more likely (Kuk *et al.*, 2009; Wolfe *et al.*, 2010). The amount of lowering and changes to speech sounds is dependent on the hearing loss, the patient's preference or clinician's fitting goals (Scollie *et al.*, 2016). All these processes in hearing aids may result in changes to the hearing aid user's speech production.

There are three main types of frequency lowering technology in current hearing aids: non-linear frequency compression (NLFC), frequency transposition, and frequency translation (Scollie, 2013). Figure 1 is a conceptual illustration showing how NLFC moves energy from a high frequency region to a lower frequency region in a commercially available hearing aid.

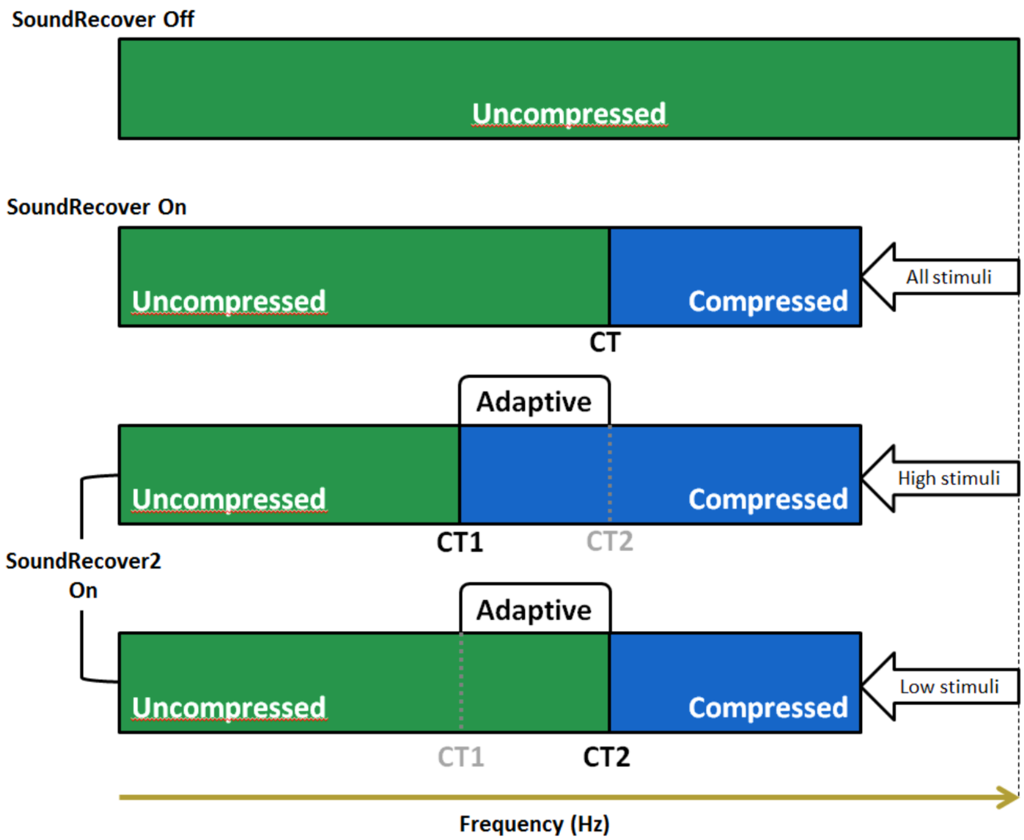

Figure 1. Conceptual illustration of non-linear frequency compression in Phonak's SoundRecover. SoundRecover Off is the top graph, SoundRecover1 is the second graph from the top, and the two non-linear frequency compression modes within SoundReover2 are the bottom two graphs. For SoundRecover2, "High Stimuli" indicates high-frequency content and "Low stimuli" indicates low frequency content. "CT" indicate the cut-off frequency of SoundRecover1. "CT1" and "CT2" indicates the different cut-off frequencies for SoundRecover2. This figure was from Figure 1 of Glista *et al.*, 2016.

In NLFC, inputs are compressed above a cut-off frequency by a specified ratio so that high frequency inputs are shifted to a lower frequency range where sufficient audibility is more likely to be attained. Inputs below the cut-off frequency are not compressed and do not overlap with the compressed frequency region, so natural formant ratios of vowels and fundamental frequencies are maintained (Wolfe *et al.*, 2010). There are adaptive and non-adaptive NLFC. Phonak, a hearing aid manufacturer, uses both types of NLFC in their frequency lowering program called SoundRecover. Their older version of SoundRecover, SoundRecover1 (SR1), uses non-adaptive NLFC, where the cut-off frequency and compression ratio remains the same for all incoming stimuli (Glista *et al.*, 2016; Rehmann, Jha, & Baumann, 2016). In contrast, Phonak's newest version of SoundRecover, SoundRecover (SR2), uses adaptive NLFC. In adaptive NLFC, the compression ratio remains the same for all incoming stimuli but the cut-off frequency changes based on the incoming signal (Glista *et al.*, 2016; Rehmann *et al.*, 2016). If the incoming signal is high frequency dominant, it will use a lower cut-off frequency (CT1). If the incoming signal is low frequency dominant, it will use a higher cut-off frequency (CT2). The non-adaptive type should reduce the distortions in the sounds caused by the NLFC, because strong processing effects should only occur for high frequency stimuli, thereby protecting vowel formants, tonal structures of speech, and other low frequency information better than adaptive NLFC (Glista *et al.*, 2016; Rehmann *et al.*, 2016).

1.1 Purpose of the current research

The main focus of this dissertation was to understand how speech production is influenced by changes in auditory feedback. This was examined by using real-time perturbations in auditory feedback via laboratory manipulations of formants and intensity, or manipulations of incoming stimuli by frequency lowering technology in hearing aids. Specifically, the following experiments examined whether the processes of error feedback were similar in younger and older adults, how the error feedback changed with a hearing loss, how hearing aids may have changed auditory stimuli and how the processed hearing aid sounds influenced error feedback.

1.2 Research questions

The following questions were used as directions for the five integrated-style chapters in the dissertation:

1) How does the interaction between air and bone conduction influence vowel compensation in formant manipulated studies? (Chapter 2)
2) How does the speech motor control system differ between younger adults and older adults? (Chapters 3 & 4)

3) How does a hearing loss and the use of hearing aids affect the speech motor control system? (Chapters 3 & 4)
4) Are there differences in vowel and fricative productions between non-adaptive and adaptive non-linear frequency compression? (Chapters 5 & 6)

1.3 Thesis Outline

The hearing of our own voice occurs through air and bone conduction where the cochlea responds to the linear sum of voice signals arriving through the two routes (Stenfelt, 2007). To study how auditory feedback is used in detecting speech errors and controlling speech, the perturbed auditory feedback in the perturbation paradigm needs to be at a higher level than the incoming unprocessed air and bone conduction sounds that arrive directly from the talker's speech productions (Purcell & Munhall, 2006). Speech production changes may depend on the ability of the talker to detect errors in their speech: if the level of the altered air conduction signal relative to the unaffected bone conduction signal is greater, then compensation can occur. Past studies have set the altered auditory stimuli to 80 dBA such that the level of the altered air conduction signal is greater than the unaffected air and bone conduction signals (Larson *et al.*, 2007; Mitsuya *et al.*, 2015, Purcell & Munhall, 2006). It is unclear how high level the perturbed auditory feedback needs to be to overcome the unprocessed air and bone conductions sounds. Chapter 2 examines differences in speech compensation behaviours at different headphone sound pressure levels with the altered auditory feedback task.

Most auditory feedback manipulation studies have examined compensatory speech responses in younger adults with normal hearing (Jones & Munhall, 2000; Mitsuya *et al.*, 2015; Villacorta *et al.*, 2007). It is unclear whether the results could be generalized to a wider range of ages. There are anatomical, cognitive, and general physiological effects of aging that may affect speech production and the ability to detect speech errors in older adults. Chapters 3 and 4 examine the responses of younger and older adults with normal hearing in vowel and intensity manipulated feedback, respectively.

A key component of auditory feedback is the ability for the individual to hear the incoming auditory cues. Individuals with hearing loss have an impaired auditory system that may have difficulties with detecting speech errors. However, hearing aids provide amplification to these individuals, such that the increased audibility of speech sounds may be enough for the speech motor control system to detect speech errors in auditory feedback. Chapters 3 and 4 include a group of older adults with hearing loss who wore binaural hearing aids to determine if their compensatory responses were different or similar to individuals with normal hearing.

The altered auditory feedback in hearing aids may induce perceived speech production errors in talkers. NLFC induces changes in hearing aid sounds as it moves high frequency information to a lower frequency region. The amount of frequency lowering that occurs depends on the settings, such that a weak setting of NLFC (i.e. a high cut-off frequency) may have less distortion in the amplified sound than a stronger setting of NLFC (i.e. a low cut-off frequency). As well, the type of NLFC may also change the sound quality, such that non-adaptive NLFC may have more distortion than adaptive NLFC. Thus, the amount of speech production change with the use of NLFC may be mediated by the strength of the frequency lowering processor and the type of NLFC used. Chapter 5 examines changes in vowel and sibliant /s/ productions across different settings for adaptive and non-adaptive NLFC in younger and older individuals with normal hearing and older adults with hearing loss.

Individuals with high frequency hearing loss may need additional digital signal processing to perceive high frequency sounds. NLFC may increase audibility in the high frequencies as it moves the high frequency sounds to a lower frequency area where audibility is most likely. The speech and auditory systems may change as they acclimatize to the new sounds introduced by NLFC. Chapter 6 examines changes in vowel and sibilant /s/ production after acclimatization to non-adaptive and adaptive NLFC in individuals with hearing loss.

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Chapter 2

2 Interaction of air and bone conduction signals in speech production during the altered auditory feedback paradigm

2.1 Introduction

When a person vocalizes, they can perceive their own voice through two sound transmission pathways. One pathway is through air conduction (AC), where the sound of their voice exits their mouth and travels to the cochlea via the ear-canal, tympanic membrane, and middle ear ossicles. The other pathway is through bone conduction (BC), where the sound from the oral cavity travels to the cochlea via the skull bone. Even though the transmission pathway to the cochlea is different between AC and BC, both sounds excite the basilar membrane the same way (Stenfelt, 2007). A mixture of AC and BC sounds or specific multi-band filtering is needed for a participant to recognize their own voice recordings as sounding most familiar (Maurer & Landis, 1990; Shuster & Durrant, 2003).

The relative contributions of AC and BC sounds to the perception of a person's own voice during vocalization are similar in magnitude but are frequency dependent. Von Bekesy (1949) attached tubes filled with cotton to participants' ears to attenuate the AC component without changing BC sounds. The decrease in loudness between open ear canals and when the tubes were applied was around 6 dB suggesting that AC and BC components were similar in magnitude. Further, Porschmann (2000) determined that BC contributions were greater between 0.7 and 1.2 kHz and AC components were greater below 0.7 and above 1.2 kHz by comparing masked thresholds for AC and BC sounds.

Vocalization patterns may also affect the relative contributions of AC and BC sounds to the perception of a person's own voice. Reinfeldt and colleagues (2010) asked participants to vocalize ten different phonemes from different phoneme categories. Their findings showed that different phonemes have different AC and BC contributions but if the phonemes have similar vocalization patterns the AC and BC contributions are similar.

For example, front vowels /e/ and /i/ were dominated by BC sounds between 700 Hz and 2.1 kHz whereas fricatives /s/ and /ɕ/ were dominated by BC sounds below 350 Hz and between 1.4 and 2.1 kHz. Similarly, Porschmann (2000) found greater BC contributions for voiced sound /z/ than unvoiced sound /s/. Von Bekesy (1949) also found that BC contributions are relatively higher for sounds produced with small opening of the mouth than by a large opening. Vocalization patterns may change the relative contributions of AC and BC sound; overall, BC contributions are greatest approximately below 2 kHz.

Auditory feedback, hearing our own voice, is important to maintain accurate speech production. Clinical studies have shown that speech articulation from individuals with hearing loss or who are post-lingually deafened is more variable than individuals with normal hearing (Waldstein, 1990; Cowie & Douglas-Cowie, 1992; Schenk *et al.*, 2003). As well, speech intelligibility is higher and speech production is more normal after a cochlear implant than before having an implant (Kishon-Rabin *et al.*, 1999; Richardson *et al.*, 1993; Svirsky *et al.*, 1992, 2000; Tobey & Hasenstab, 1991; Ubrig *et al*, 2011). Laboratory studies that manipulate a person's auditory feedback system also show the importance of hearing one's own voice for accurate speech production. These studies use a real-time auditory feedback paradigm where an acoustic parameter of interest is manipulated in real-time and presented through a headset to the participant while the participant is producing a target sound. The speech motor control system of the participant may detect a difference or an error between the manipulated auditory feedback signal and the intended target sound, as a result, the participant may change their articulation. Various acoustic parameters such as voice pitch (Burnett *et al.*, 1998; Jones & Munhall, 2000; Larson *et al.*, 2007), voice amplitude (Bauer *et al.*, 2006), vowel formants (Purcell & Munhall, 2006b; Mitsuya *et al*, 2015; Villacorta *et al.*, 2007), and fricative noise (Shiller *et al.*, 2009; Casserly, 2011) have been manipulated with this real-time auditory feedback paradigm.

In vowel formant manipulation studies, the first and/or second formant (F1 and F2, respectively) are perturbed such that their formant values are increased or decreased in real time. This results in a slightly different feedback vowel sound than the intended target sound. For example, while participants are producing the word "head", the F1 value of the vowel /ɛ/ is decreasing in Hertz (Hz) such that the feedback sounds more like "hid". As a result, participants will increase their F1 value slightly towards /æ/. Most participants will partially compensate in the opposite direction of the manipulation (Mitsuya *et al.*, 2011; Munhall *et al.*, 2009; Purcell & Munhall, 2006b). The magnitude of compensation varies based on perturbation magnitude, direction of manipulation and vowel (MacDonald *et al*, 2010; Mitsuya *et al.*, 2015).

The manipulated sounds in the real-time perturbation studies are presented to the participants at relatively high sound pressure levels (SPLs) through circumaural headphones, such as 75 dB SPL (Jones & Munhall, 2000; Eckey & MacDonald, 2015), 80 dBA (Larson *et al.*, 2007; Mitsuya *et al.*, 2015), or 87 dB SPL (Villacorta *et al.*, 2007). Presenting the manipulated sounds at relatively high SPLs lowers the possibility that the participant can hear their own unprocessed voice through AC or BC. Circumaural headphones are acoustically open and do not have an occlusion effect. The unprocessed BC sounds can be easily masked by the high presentation levels. However, the occlusion effect may occur with insert earphones such that lower unprocessed BC sounds (< 1000 Hz; Killion *et al.*, 1988) in the ear canals may be increased and the high level processed sounds may not mask the BC sounds. A study by Mitsuya *et al.* (2016) examined whether using insert earphones or circumaural headphones would elicit different compensatory productions when F1 was manipulated in real-time for /ɛ/ with a higher F1 and /ɪ/ with a lower F1. The different transducers and vowels elicited similar compensatory formant productions. This suggests that the high SPL presentation (i.e. 80 dBA) masked the unprocessed AC and BC for circumaural headphones and insert earphones. However, it is unknown if lower SPL presentations of the processed sounds would elicit similar corrective behaviours as high SPL presentations.

The processed sounds presented through headphones may also be mixed with pink or speech-shaped noise (Larson *et al.*, 2007; Mitsuya *et al.*, 2015). The purpose of the noise is to reduce any artifact or unnatural sounds that could occur through the formant manipulations, such as clicks. The noise may also mask the unprocessed BC sounds. It is uncertain if the presence or absence of the noise has an effect on speech compensation behaviours.

The effects of BC feedback on formant compensation can also be examined by varying headphone presentation levels. With lower headphone levels, there is a higher probability that the participant can hear the unprocessed BC sounds of their own voice. Less compensation may occur as the speech motor control system may be controlled more by the unprocessed BC sounds than the processed AC sounds. In the current study, we evaluated compensation of F1 for /ɛ/ by varying headphone SPLs mixed with speech-shaped noise or not mixed with speech-shaped noise to identify differences in compensation behaviours.

2.2 Method

2.2.1 Participants

Twenty-five female speakers were recruited from Western University in Canada to participate in the current study. Their ages ranged from 21 to 31 years with a mean of 24.63 years and a standard deviation of 3.06 years. All speakers considered themselves native English speakers and all speakers but five learned English in Ontario, Canada. Two speakers came from Quebec, Canada and Maryland, USA. The remaining speakers immigrated to Ontario, Canada at less than three years of age from South Africa, England, and Iran. All had normal hearing thresholds within the range of 250-8000 Hz [$\leq$20 dB hearing level (HL)] and none reported a history of neurological, language, or speech impairments. Data from one participant was discarded because she was unable to attend the second session.

Each participant was tested in two sessions and was given four different conditions of headphone levels (50, 60, 70, 80 dBA). The order of the headphone levels was counterbalanced but the same order was used for both sessions for each participant. The first session consisted of speech feedback only (noise absent) and the second session consisted of speech feedback with speech noise (noise present). When each session was completed, participants were compensated $5 for every half hour for their time.

2.2.2 Equipment

Equipment used in the current study was similar to that reported in Mitsuya *et al.* (2015). Participants were seated in a sound attenuated booth (Eckel Industries of Canada, Ontario, Canada; model C26). Participants wore a headset microphone (Shure WH20) and were prompted to speak when the target word appeared on a computer screen at a rate of approximately once every four seconds. The microphone signal was amplified with a microphone amplifier (Tucker-Davis Technologies MA3), low pass filtered with a cut-off frequency of 4500 Hz (Frequency Devices type 901), digitized at a 10 kHz sampling rate with 18-bit precision and filtered in real time to produce formant shifts (National Instruments PXI-6289M input/output board). The processed signal was then amplified to the various headphone level conditions (50, 60, 70 or 80 dBA) using the Madsen Itera and played through headphones (Sennheiser HD 265) for the first condition (noise absent). In the second condition (noise present), the processed signal was also mixed with speech shaped noise (Madsen Itera) at 50 dBA.

2.2.3 Online acoustic analysis and model order estimation

A statistical amplitude threshold technique was used to detect voicing and an infinite impulse response filter previously described by Purcell and Munhall (2006a) was used to shift formants in real-time. An iterative Burg algorithm was used to estimate formants every 900 μs (Orfanidis. 1988). Filter coefficients were calculated based on these formant estimates such that a pair of spectral zeros was placed at the existing formant frequency and a pair of spectral poles was placed for the new formant to de-emphasize and emphasize existing voice harmonics, respectively.

The number of coefficients needed for the autoregressive analysis is called the model order. This was estimated by collecting six tokens of the English vowel /ε/ in /hVd/ context. The word “head” was presented on a computer screen for 2.5 s with an inter-trial interval of 1.5 s. Speakers were instructed to speak in their normal voice without pitch gliding. The best model order was chosen based on minimum variance of F1 and F2 frequencies over the middle portion of /ε/.

2.2.4 Offline formant analysis

The method for offline formant analysis is the same method reported in Munhall *et al.* (2009). The harmonicity of the power spectrum was used to estimate the vowel boundaries. The boundaries were inspected and corrected if necessary. Vowel formants (F1, F2, and F3) were estimated from the middle 40-80% of the vowel's duration, with a 25 ms window that was shifted in 1 ms increments until the end of the middle portion of the vowel segment. A single average value for each of the formants was calculated from these sliding window estimates. Formant estimates were examined and were relabeled if incorrect (e.g. F2 being labelled as F1) or removed if the formant under examination was well beyond the distribution of other tokens.

2.2.5 Experimental phases

All participants performed two sessions for the experiment. The time between sessions ranged from 8 to 60 days, with a mean of 29 days and a standard deviation of 12.69 days. Each session consisted of the short model order estimation segment and the four headphone level tasks (50, 60, 70, and 80 dBA). The order of the level conditions was counterbalanced but the same order was used for both sessions. The first and second sessions were identical, except that the first condition consisted of speech feedback only (noise absent) and the second session consisted of speech feedback with speech noise (noise present).

For the perturbation task, speakers produced 120 utterances of the word "head" when a visual prompt was presented. These 120 trials were divided into five experimental phases. In the Acclimatization phase (utterances 1-20), participants received normal feedback. These utterances were discarded during analyses. In the Baseline phase (utterances 21-40), participants received normal feedback. In the Ramp phase, (utterances 41-70), the F1 value was increased by 50 Hz every 10 utterances (Ramp50, Ramp100 and Ramp150). In the Hold phase (Hold200; utterances 71-90), the maximum +200 Hz F1 perturbation was held constant. At utterance 91, the End phase began in which the perturbation was removed and the participants received normal feedback until the end of the condition (End0, utterances 91-120). A schematic of the experimental phases can be seen in Figure

2. Participants were asked to take off their headphones and were given a passage to read aloud with a five minute break to normalize their speech productions after each condition.

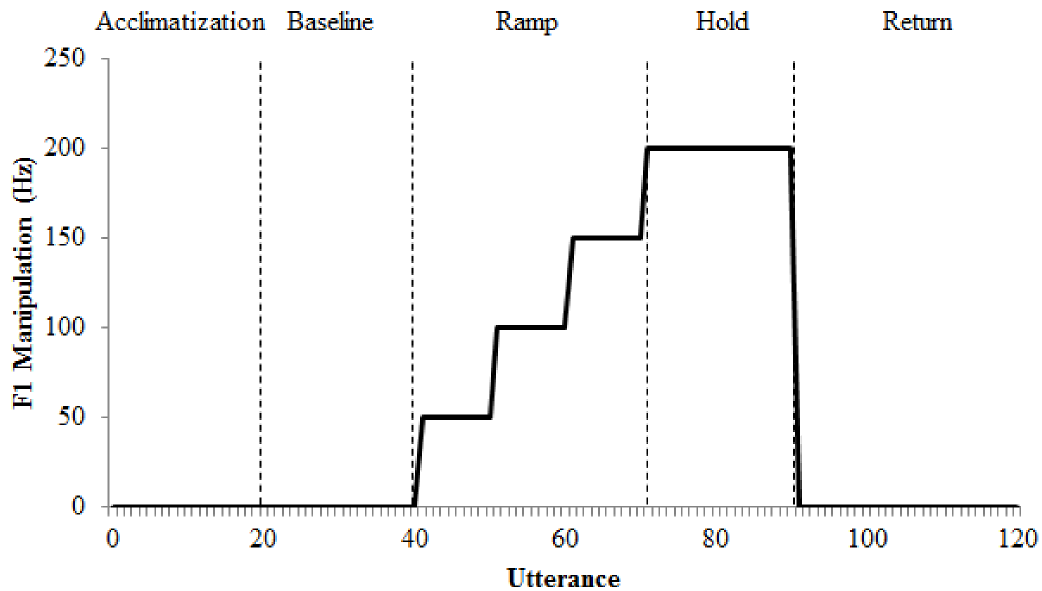

Figure 2. Schematic procedure of the feedback perturbation applied to the first formant. The vertical dotted lines represent the boundaries of the five experimental phases: Acclimatization, Baseline, Ramp, Hold, and Return (from left to right).

2.3 Results

Statistical analyses in this study were completed using SPSS (version 24; IBM, Armonk, NY). The baseline average of F1 was calculated using the utterances from the Baseline phase (i.e. utterances 21-40). The F1 values were then normalized by subtracting a speaker's baseline average from each utterance. To quantify a change in formant production, the average normalized F1 value during each phase was calculated. Figure 3 shows the average compensation for each phase and headphone level across participants.

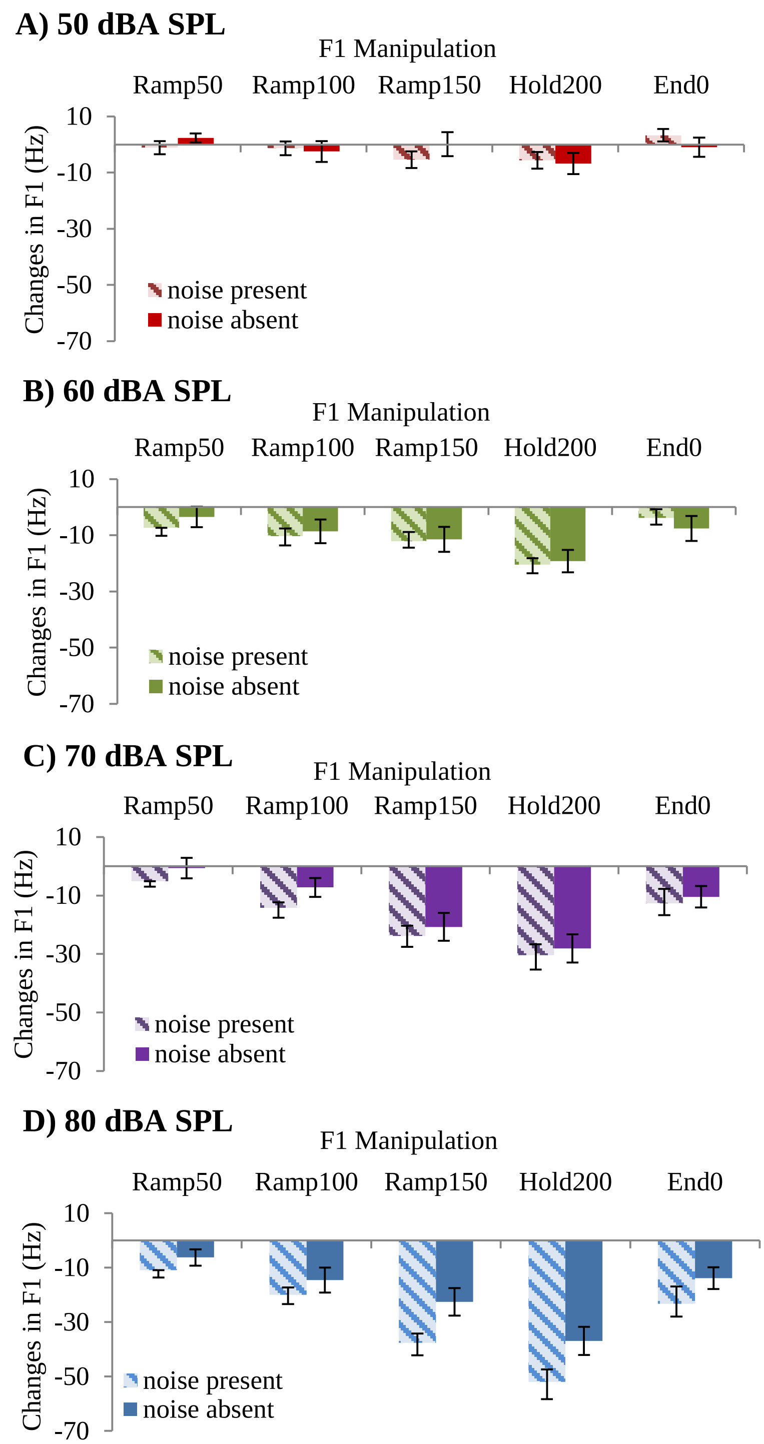

Figure 3. Average magnitude of compensation across headphone levels: A) 50 dBA; B) 60 dBA; C) 70 dBA; D) 80 dBA. Striped columns indicate conditions with speech-shaped noise present. Solid columns indicate conditions with speech-shaped noise absent. The error bars indicate ± 1 standard error.

A three-way repeated measures analysis of variance (RM-ANOVA) was completed with headphone levels (four levels: 50, 60, 70, and 80 dBA), noise (two levels: present or absent) and phases (five levels: Ramp50, Ramp100, Ramp150, Hold200, and End0) as the three within-subject factors and magnitude of F1 change as the dependent measure. For all statistical analyses, the Greenhouse-Geisser corrected degrees of freedom were used to adjust for lack of sphericity prior to interpretation of effects. Results of the RM-ANOVA were interpreted at an α of 0.05.

The main effect of noise was non-significant [$F(1, 23) = 3.15$, $p = 0.09$, $\eta_p^2 = 0.12$]. The main effects of headphone levels and phases were significant [headphone levels: $F(3, 69) = 22.11$, $p < 0.001$, $\eta_p^2 = 0.49$; phases: $F(2.41, 55.39) = 43.10$, $p < 0.001$, $\eta_p^2 = 0.65$]. Figure 4 shows the average normalized F1 change across participants for each headphone level.

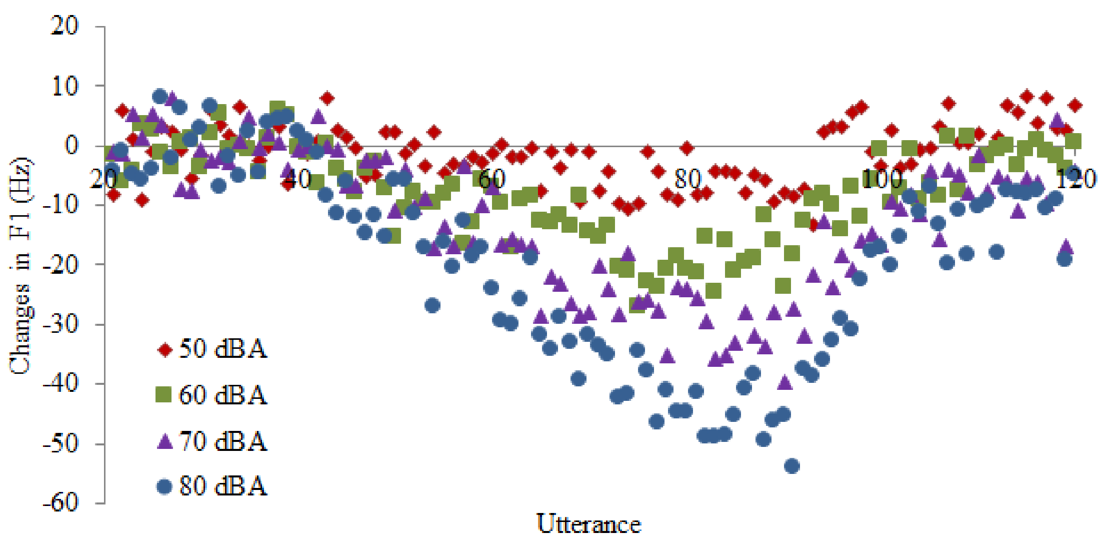

Figure 4. Average change in F1 values across participants for each utterance from Baseline (utterance 20) at each headphone level.

The interaction of the three variables was non-significant [$F(6.27, 144.30) = 1.08$, p = 0.38, $\eta_p^2 = 0.05$]. The interaction between noise and phases was non-significant [$F(2.83, 65.18) = 1.34$, $p = 0.27$, $\eta_p^2 = 0.06$]. The interaction between noise and headphone levels was non-significant [$F(2.34, 53.77) = 1.64$, $p = 0.20$, $\eta_p^2 = 0.07$]. The interaction between headphone levels and phases was significant [$F(7.08, 162.74) = 8.78$, $p < 0.001$, η_p^2 = 0.28]. Post hoc analysis, corrected for multiple comparison using Bonferroni corrections, is shown in Figure 5. Overall, headphone level 50 dBA had less F1 change than 80 dBA across phases and the interaction with phases was just as expected.

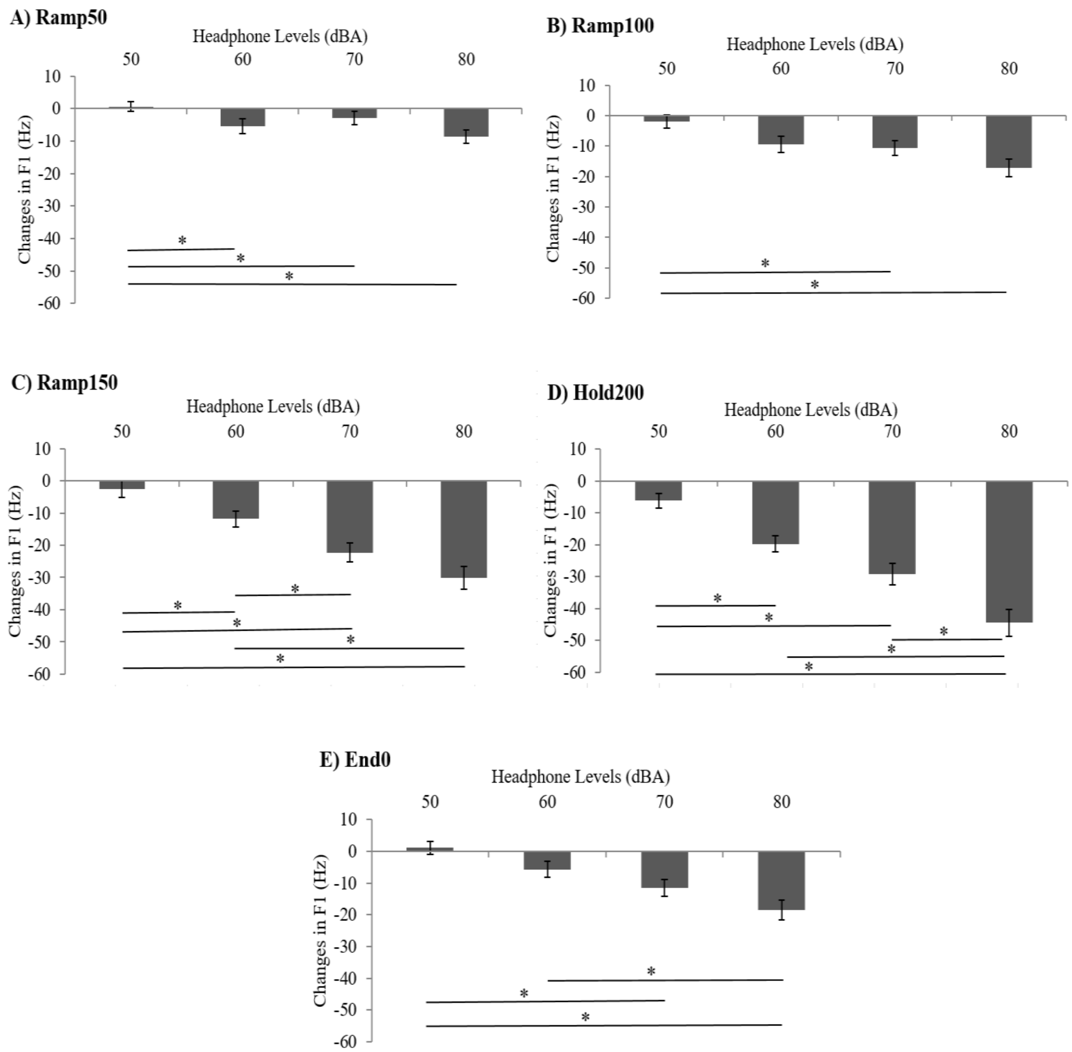

Figure 5. Average magnitude of compensation across phases: A) Ramp50; B) Ramp100; C) Ramp150; D) Hold200; E) End0. The error bars indicate ±1 standard error. * indicates a significant difference ($p \leq 0.05$).

2.4 Discussion

The purpose of the study was to examine whether talkers' compensatory formant production in response to formant manipulation depends on the sound pressure of the auditory feedback presented through headphones. Different headphone SPLs were examined in young adults with normal hearing. The current study gradually increased F1 of the vowel /ɛ/ to a perturbation of 200 Hz while talkers produced the word head. This perturbation made the headphone feedback sound more like "had", and as a result, talkers lowered their F1 production to counteract the perturbed auditory feedback by producing a sound more like /ɪ/ (as in "hid"). The data showed that compensatory formant productions significantly differed between the different headphone SPLs.

The Ramp phase of the experiment influenced F1 compensation. Each step within the Ramp phase increased the F1 perturbation by 50 Hz until the maximum perturbation of 200 Hz at the Hold phase. When the F1 perturbation was small (i.e. +50 Hz at Ramp50), higher SPL presentations (60, 70 and 80 dBA) had larger F1 compensations than the lowest SPL presentation of 50 dBA. This suggested small auditory feedback errors may be more difficult to detect, such that small perturbations need to be presented at a higher level for the speech motor control system to detect the error. Thus, when there is a small perturbation in auditory feedback, the altered auditory feedback signal needs to be high enough level to mask the BC sound reaching the cochlea so that the perturbation can be detected. In contrast, when the F1 perturbation was larger, all the headphone SPLs elicited significantly different magnitudes of compensations relative to each other. This suggested that large auditory feedback errors may be easier to detect and these large errors can be detected at lower level presentations.

The compensation differences between different headphone SPLs did not remain consistent at each step of the Ramp phase (see Figure 5). For example, at Ramp50, 50 dBA had significantly less F1 compensation than 60, 70, and 80 dBA. Whereas, at Ramp150 there were more significant differences between the headphone SPLs, such that 50 dBA had the least amount of compensation, 60 and 70 dBA were significantly different from each other. Feedback of 60 and 80 dBA also elicited significantly different compensation. This showed that at each headphone SPL, the contribution of AC relative to BC varied. With increasing SPL, the processed AC sounds were more able to mask the unprocessed BC sounds at the cochlea.

Formant perturbation studies have mainly focused on the magnitude of compensation at the Hold phase (Munhall *et al.*, 2009; Mitsuya *et al.*, 2015; Villacorta *et al.*, 2007). At the Hold phase, the maximum magnitude of perturbation is held constant. The study by MacDonald and colleagues (2009) determined that the rate of change within the Ramp phase did not affect the compensation at the Hold phase. The current study found differences in compensation between headphone SPLs within the different steps in the Ramp phase, however, the most differences and largest magnitude of compensations occurred at the Hold phase.

The magnitude of F1 compensations significantly differed across the different SPLs. The largest magnitude of F1 compensation occurred with 80 dBA. This SPL is a relatively high level presentation such that the subtle effect of BC sound on the net speech signal at the cochlea might have been masked by the high level presentation of altered AC feedback. In comparison, the smallest magnitude of F1 compensation occurred with 50 dBA. This SPL is a relatively low level presentation such that the altered AC feedback was not able to mask the BC sound in the net speech signal at the cochlea. The interaction of air and bone conducted sounds at the cochlea may be level dependent, such that if the AC sound is at a high enough level, it may mask BC sound and vice versa. Similarly, a common complaint in hearing aid users is the occlusion effect (Chung, 2004; Stender & Appleby, 2009; Winkler, Latzel, & Holube, 2016). The occlusion effect is where the BC sound from the user's speech production is trapped in the ear canal and is transmitted to the cochlea via the AC pathway including the ear drum and middle ear. A solution to reduce the occlusion effect is to increase the hearing aid gain (increase the amplified AC sound) such that it masks the BC sound (Chung, 2004).

The study by Mitsuya and Purcell (2016) studied the occlusion effect in formant compensation behaviours. They compared the formant compensation patterns between headphones and insert earphones. Their results found no differences in F1 compensation between the two transducers for F1 perturbations of /ɛ/ and /ɪ/. They indicated the high level of 80 dBA they used may have masked the unprocessed AC and BC sounds and the occlusion effect may not have occurred in their study. The current study found differences between the different headphone SPLs; it would be interesting to replicate the study by Mitsuya and Purcell (2016) with different headphone and insert earphone SPLs. Such a study might further our understanding of the ability of the speech motor control system to regulate speech production using AC and BC feedback.

The results showed the subtle influences of BC feedback on the regulation of feedback. These results have implications for patients who use BC hearing devices such as Bone Anchored Hearing Aids (BAHA®) by Cochlear (Cochlear Ltd, 2017) or the Ponto by Oticon Medical (Oticon Medical, 2017). These patients have conductive hearing losses, such that sound transmission is reduced or blocked through the external and/or middle ear (Gelfand, 2009). As a result, they have difficulties hearing air conducted sounds and have difficulties with AC devices. It is important for BC devices to provide speech sounds that are not distorted to avoid negatively affecting speech production. As well, these BC devices must provide enough amplification such that the users can detect errors in their speech production. Future studies may want to include individuals who use BC hearing devices to determine the speech motor control system's abilities to use and process air and bone conducted sounds for speech regulation.

Further, the speech motor control system in individuals with hearing loss may be different compared to individuals with normal hearing. A person with hearing loss may not rely on their auditory feedback as their hearing loss has reduced their ability to hear sounds (Laugesen *et al.*, 2008). Their speech motor control system may rely more on other feedback systems, such as somatosensory (Laugesen *et al.*, 2009). Future studies that manipulate auditory feedback may want to include individuals with hearing loss to determine how the speech motor control system changes when the auditory system is impaired.

The current study manipulated the F1 of /ɛ/ to study compensatory behaviours across different headphone SPLs. The study by Mitsuya & Purcell (2016) found no significant differences in formant compensatory responses between use of headphones or insert earphones for /ɛ/ or /ɪ/. However, they found that the talker's voice amplitude differed between the two vowels. As well, Mitsuya and colleagues (2015) found different F1 compensation behaviours across seven English vowels. They suggested that different vowels may vary in their use of different feedback modalities to detect errors. Further, Reinfeldt and colleagues (2010), Porschmann (2000), and von Bekesy (1949) found different AC and BC contributions across different phonemes. The results of the current study may be limited to the vowel /ɛ/ and future studies may want to include other vowels or phonemes.

The addition of noise in perturbation experiments may change the level at which a person speaks (Siegel & Pick, 1974) or may change how the spectral noise of fricatives is perceived (Casserly, 2011). In formant perturbation studies (Larson *et al.*, 2007; Mitsuya *et al.*, 2015; Purcell & Munhall, 2006b), speech-shaped or pink noise was added to the feedback of headphones to reduce the perception of artifact sounds and distortions caused by the formant perturbation signal processing. The results in the current study showed no significant differences between conditions with and without speech-shaped noise. These results indicate a reduction in concern for these sound distortions and do not show the need to use speech-shaped noise in real-time formant perturbation paradigms.

In conclusion, the current data show that headphone level elicited differences in formant compensatory responses to real-time formant perturbations. When providing participants with altered auditory feedback, high level presentation is strongly recommended to produce the highest magnitude of speech compensation. High level presentations of altered auditory feedback may be able to mask the BC signal at the cochlea, such that the speech motor control system will only receive information from altered auditory feedback. In order to discern how the speech motor control system is able to use and process acoustic information from AC and BC feedback, further examinations are needed, such as having individuals with hearing loss as participants.

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Chapter 3

3 Speech compensations to real time formant perturbations in individuals with normal hearing and hearing loss

3.1 Introduction

During vocalization, the shape of the vocal tract and movements of the articulators emphasize harmonics in certain frequency bands to produce vowels. These emphasized bands containing harmonics are called formants and the formant frequencies composing each vowel define its identity. Each vowel has multiple formants, however, the first (F1) and second (F2) formants are adequate to perceptually distinguish different vowels and are the most important for vowel quality (Peterson & Barney, 1952; Potter & Steinberg, 1950). Vowels are involved in both prosodic and segmental features of speech; as a result, they play an important role in speech production and perception (Ozbic & Kogovsek, 2010). It is important to maintain accurate vowel production, however, this accuracy can be reduced due to aging or a hearing loss.

Aging can have an effect on vowel production. Some aging effects include centralization of vowels (Benjamin, 1982; Liss, Weismer, & Rosenbek, 1990; Torre & Barlow, 2009; Zraick, Gregg & Whitehouse, 2006), lowering of formant frequencies (Benjamin, 1982; Endres, Bamback, & Flosser, 1971; Harrington, Palethope, & Watson, 2007; Linville & Fisher, 1985; Liss, Weismer, & Rosenbek, 1990; Torre & Barlow, 2009; Xue & Hao, 2003; Zraick, Gregg, & Whitehouse, 2006), a decreased rate of frequency change along formant transitions (Zraick, Gregg, & Whitehouse, 2006), and longer vowel durations (Benjamin, 1982; Zraick, Gregg, & Whitehouse, 2006). Further, the vowels of older adults between 68 and 82 years of age are more variable than younger speakers (Benjamin, 1982), and the variability increases with individuals over 87 years of age (Liss, Weismer, & Rosenbek, 1990). These findings are consistent across cross-sectional and longitudinal studies (Endres, Bamback, & Flosser, 1971; Harrington, Palethorpe, & Watson, 2007; Linville, 1987; Linville & Fisher, 1985).

Cognitive, anatomical and physiological effects of aging may affect the individual's speech production. The differences in vowel productions between elderly and young adults could be due to the limited tongue and lip movements in older adults, decreased accuracy of motor control, reduced sensory feedback (e.g. hearing loss), and/or decreased cognitive-linguistic function (Benjamin, 1997; Torre & Barlow, 2009; Zraick, Gregg, & Whitehouse, 2006). Other physiological and anatomical changes may include increased dimensions of the vocal tract and oral cavity, lowering of the larynx, stretching of ligaments, and/or muscle atrophy in the pharynx and tongue (Benjamin, 1997; Linville & Fisher, 1985; Xue & Hao, 2003).

A hearing loss may also affect vowel production. Frequently reported vowel production errors are neutralization towards a central "schwa" vowel (Cowie & Douglas-Cowie, 1983; Ozbic & Kogovsek, 2010; Plant, 1984; Smith, 1975), less differentiation between vowels (Ozbic & Kogovsek, 2010), and/or substitutions of target vowels with neighbouring vowels in the vowel quadrilateral (Coughlin, Kewley-Port & Humes, 1998; Dorman *et al.*, 1985; Owens, Talbott & Schubert, 1968; Richie, Kewley-Port, & Coughlin, 2003), increased vowel duration (Cowie & Douglas-Cowie, 1983; Plant, 1984), substitution of diphthongs for vowels and nasalization of vowels (Richardson *et al.*, 1993). However, the effects of hearing loss vary depending on when the hearing loss was acquired, the type of amplification the individual uses, and the degree of hearing loss (Kosky & Boothroyd, 2001). For example, postlingually deafened speakers usually continue to produce intelligible speech for years following their hearing loss, probably due to the robustness of somatosensory goals and feedforward commands they acquired while they could hear (Cowie & Douglas-Cowie, 1983; Menard *et al.*, 2007). Nevertheless, such speakers do experience degradation of their speech; the degree of degradation varies considerably from one individual with hearing loss to another (Langereis *et al.*, 1997; Menard *et al.*, 2007).

Sensorineural hearing losses will most likely affect auditory sensitivity in the high frequencies. As a result, it is expected that a person with a hearing loss may have more difficulty perceiving and producing high and less audible vowel formants, [e.g. second formants (F2) and third formants (F3)] than low, more audible vowel characteristics [e.g.

fundamental frequency (F0) and first formant (F1)] (McCaffrey & Sussman, 1994; Nicolaidis & Sfakiannaki, 2007). Further, Ozbic and Kogovsek (2010) analyzed F1 and F2 vowel productions of individuals with hearing loss and normal hearing. They found individuals with hearing loss differed more in F2 productions than F1 productions compared to individuals with normal hearing. They also found more production differences between individuals with hearing loss and normal hearing for front vowels (e.g. /i/) that have F2 formants that are high in frequency and low in intensity than back vowels (e.g. /u/) that have F2 formants that are lower in frequency and high in intensity. The current study examined speech production in three vowels /ɛ, ɪ, i/ with different formant frequencies. The vowel /i/ has the highest F2 within the vowel space which may be more affected by a hearing loss compared to the other vowels. Whereas, the vowel /ɛ/ has a relatively low F2 that may not be affected by a high frequency hearing loss.

Auditory feedback, the hearing of one's voice, is involved with the regulation of speech production and speech error detections. Studies have shown that perturbations in auditory feedback will result in changes in speech that corrects for the perturbation. For example, Borden *et al.* (1994) have shown that if the sound of a talker's voice is amplified, the talker will reduce their vocal intensity, whereas, if the talker's voice is attenuated, the talker will increase their vocal intensity. Also, if the auditory feedback is filtered, talkers may change their speech by modifying the characteristics of vocal tract resonances so that the target speech sound could be attained (Garber *et al.*, 1981). Speech compensation has been found in studies that manipulated F0 in talkers. Talkers compensated to the F0 manipulation by compensating in the opposite direction (Burnett *et al.*, 1998; Jones & Munhall, 2000). A pattern has occurred across manipulated auditory feedback studies: Talkers change their productions of speech in the opposite direction of the manipulation.

This compensation pattern has also been found in vowel manipulation studies (MacDonald, Goldberg, & Munhall, 2010; Mitsuya *et al.*, 2015; Munhall *et al.*, 2009; Purcell & Munhall, 2006ab). These studies are generally comprised of four stages where the talker is continually repeating single utterances of a vowel in /hVd/ context: baseline, ramp, hold, and end. There are no perturbations to the formants in the baseline phase. During the ramp phase, F1 or F2 of a vowel is either gradually increased or decreased.

For example, if F1 of the vowel /ɛ/ in "head" is gradually shifted down, it would progressively sound more like /ɪ/ in "hid"; if F1 is gradually shifted up, it would progressively sound more like /æ/ in "had". The perturbation rate during the ramp phase does not affect speech compensation but the magnitude of the perturbation change affects the magnitude of the speech compensation (MacDonald *et al.*, 2010). In the hold phase, the formant manipulation is held constant at the maximum amount. The maximum magnitude of F1 perturbations is usually 200 Hz as this amount of manipulation would change the target vowel sound to another vowel category (MacDonald *et al.*, 2010; Mitsuya *et al.*, 2015). An F2 perturbation requires a larger magnitude of F2 change to elicit F2 compensation, such as 250 Hz for the vowel /ɛ/ (MacDonald, Purcell, & Munhall, 2011). This magnitude of F2 perturbation may change across different vowels as the F2 distances between vowels vary. Finally, in the end phase, the perturbation is removed and the talker receives normal auditory feedback. The current study used these four phases to study F1 and F2 compensations across the vowels /ɛ, ɪ, and i/.

Formant perturbation studies have mainly been studied with young adults with normal hearing (MacDonald, Goldberg, & Munhall, 2010; Mitsuya *et al.*, 2015; Villacorta, Perkell & Guenther, 2007). A study by Mollaei, Shiller and Gracco (2013) manipulated F1 of /ɛ/ and measured formant compensation behaviours in older adults with Parkinson's Disease and age-matched healthy controls. They found that the magnitude of F1 compensation was reduced in older adults with Parkinson's Disease compared to the age-matched healthy controls. However, it is uncertain if the compensation of healthy older adults would be similar to younger adults.

Further, a hearing loss may affect a person's ability to monitor their own voice quality because they may not be able to hear themselves. This may prevent the person from detecting speech errors when they are unaided. However, if the person is aided with hearing aids, sounds are now audible and their auditory feedback system may be able to use these cues. In other words, when a person with hearing loss is aided with hearing aids, they may be able to detect speech errors in their own voice and make appropriate corrections. However, these corrections may be limited because the hearing aids act as an amplification device and do not fix any deterioration within the speech motor control system or auditory system that may have occurred due to the hearing loss, such as broadened auditory filters (Bernstein & Oxenham, 2006; Carney & Nelson, 1983; Dubno & Dirks, 1989; Glasberg & Moore, 1986; Lutman, Gatehouse, & Worthington, 1991). It is uncertain if the detection of speech errors in individuals with hearing aids is different or similar to the detection of speech errors for individuals with normal hearing. Three groups of participants: older adults aided with binaural hearing aids, and older and younger adults with normal hearing were included in the current study to examine speech compensation behaviours to formant perturbations in an altered auditory feedback paradigm.

The purpose of the proposed study was to identify differences in the use of auditory feedback between individuals with normal hearing and those with hearing loss. We also investigated how hearing aids affect the use of auditory feedback to reveal information about the maintenance of speech production and perception in individuals with hearing loss. Older adults who acquire a hearing loss can maintain intelligible speech as their speech motor control system is able to use the available sounds they are still able to hear, somatosensory goals and feedforward commands (Cowie & Douglas-Cowie, 1983; Menard *et al.*, 2007). When these individuals are aided, sounds are amplified and the speech motor control system may be able to detect speech errors better. However, the effects of hearing loss on the auditory system, such as broadened auditory filters, may still effect the perception of speech errors. It was hypothesized that older adults with hearing aids may have less speech compensation than older adults with normal hearing. Further, the speech motor control system may not be as efficient in older adults with normal hearing or hearing loss compared to younger adults because hearing loss and aging effect cognition, anatomy and auditory systems (Benjamin, 1997; Torre & Barlow, 2009; Kosky & Boothroyd, 2001; Zraick, Gregg, & Whitehouse, 2006). It was therefore also hypothesized that the older adults may have less compensation than younger adults.

3.2 Method

3.2.1 Participants

Ninety-two speakers were recruited from the city of London, Ontario and were divided into three groups: older adults with hearing aids (HAs), control group: older adults and control group: younger adults. Routine audiometry using a Grason-Stadler 61 audiometer was completed in a double-walled sound treated booth on all participants. Air conduction thresholds were obtained using Etymotic Research ER-3A insert earphones coupled to foam tips and measured bilaterally at all octave and interoctave frequencies between 250 and 8000 Hz. To be included in the older adults with hearing aids group, participants had to be within 55-80 years of age, have interaural differences less than or equal to 20 dB at each frequency, have a sensorineural hearing loss and binaural hearing aid use of at least one year. To be included in the control group: older adults, participants had to be within 55-80 years of age, have thresholds less than 40 dB HL between 500-4000 Hz, and interaural differences less than or equal to 20 dB at each frequency. To be included in the control group: younger adults, participants had to be within 18-35 years of age, have thresholds less than 20 dB HL between 250-8000 Hz and interaural differences less than or equal to 20 dB at each frequency.

Routine otoscopy was completed on all participants to rule out any contraindications, such as foreign bodies, discharge, or occluding wax in the ears. Tympanograms were obtained binaurally using a Madsen Otoflex 100 immittance meter and all participants except two had static compliance and tympanometric peak pressure within normal limits. Two participants (male: 75 years and male: 70 years) presented a tympanogram with peak pressure of -165 and -133 daPa, respectively. Since previous records also indicated consistent negative peak pressure, these participants were not excluded from the study.

All participants considered themselves native English speakers. All but five speakers acquired Canadian English as their first language. Three participants were from Texas, Ohio and Colorado and two participants came to Canada at less than five years of age. No participants had known language, neurological or speech impairments. Sixteen participants were not included in the study because they were older than 80 years of age, had full dentures, English was not their first language, or did not fit the hearing requirements of their respective group. In total, sixty-two speakers met all of the criteria for inclusion of their respective groups and are reported in the analyses that follow.

Participants completed the study in two sessions of approximate three hours in total. In the first session, participants completed the audiologic assessments and questionnaires. In the second session, participants completed the speech perturbation experiment. At the end of each session, participants were compensated $5 for every half hour for their time.

3.2.2 Hearing aid and hearing aid fitting

Research grade Phonak Audeo V90-13 receiver-in-canal hearing aids were chosen to fit the range of hearing losses included in the study. The Phonak Audeo hearing aids with standard receivers were suitable for mild-to-moderate-severe hearing losses, whereas the Phonak Audeo with power receivers were suitable for moderate to severe hearing losses. Closed domes were attached to the receivers and the participants' ears were occluded with silicone earmold impression material (Hal-Hen, Per-Form H/H) at each session.

A research version of Phonak Target v4.1 programming software was used for programming. The hearing aids consisted of one program for direct audio input (DAI) only. The microphone in this program was activated in omnidirectional mode during hearing aid verification and deactivated prior to testing. The volume control and other digital signal processing features in the hearing aid such as noise reduction were deactivated. The compression of the hearing aids was set to linear.

Test-box hearing aid verification was carried out using the Speechmap feature of an Audioscan® RM500SL hearing aid analyzer (Audioscan, Dorchester, ON, Canada) in a sound booth. Real ear to coupler difference was measured binaurally using a RE770 transducer coupled to the foam tip used for audiometry on the Audioscan® Verifit2. The output of the hearing aids was verified to meet Desired Sensation Level v5 targets (Scollie *et al.*, 2005) for adults at input levels of 55, 65 and 75 dB SPL for digitized speech passages found in the hearing aid analyzer. Hearing aid gain was adjusted using the Phonak Target v4.1 programming software.

Using the live-speech function of the hearing aid analyzer, a running speech passage and speech-shaped noise from the audiometer (Madsen Itera) was also verified. The running speech passage were verified to input level around 80 dB SPL and the speech-shaped noise was verified to input level around 50 dB SPL. This verification was completed to ensure equivalency of signal output levels between the hearing aids used by the older adults with hearing loss and insert earphones used by the control groups.

3.2.3 Equipment

Equipment used in the current study was similar to that reported in Mitsuya and Purcell (2016). Participants were seated in a sound attenuated booth (Eckel Industries of Canada, Ontario, Canada; model C26). Participants wore a headset microphone (Shure WH20) and were prompted to speak when the target word appeared on a computer screen at rate of approximately once every four seconds. The microphone signal was amplified with a microphone amplifier (Tucker-Davis Technologies MA3), low pass filtered with a cut-off frequency of 4500 Hz (Frequency Devices type 901), digitized at a 10 kHz sampling rate with 18-bit precision and filtered in real time to produce formant shifts (National Instruments PXI-6289M input/output board). The processed signal was then amplified to 80 dBA and mixed with speech-shaped noise at 50 dBA (Madsen Itera) through foam tip insert earphones (Etymotic Research ER2) for the control groups. The older adults with hearing aids had the processed signal and speech-shaped noise through the hearing aids via DAI input.

3.2.4 Online acoustic analyses and model order estimation

A statistical amplitude threshold technique was used to detect voicing and an infinite impulse response filter previously described by Purcell and Munhall (2006a) was used to shift formants in real-time. An iterative Burg algorithm was used to estimate formants every 900 μs (Orfanidis. 1988). Filter coefficients were calculated based on these formant estimates such that a pair of spectral zeros was placed at the existing formant frequency and a pair of spectral poles was placed for the new formant to de-emphasize and emphasize existing voice harmonics, respectively.

The number of coefficients needed for the autoregressive analysis is called the model order. This was estimated by collecting six tokens of each English vowel /i, ɪ, e, ɛ, æ, ɔ, o, u, ʊ, ʌ/ in /hVd/ context (“heed”, “hid”, “hayed”, “head”, “had”, “hawed”, “hoed”, “who’d”, “hood”, and “hud”, respectively). The words were presented on a computer screen for 2.5 s with an intertrial interval of 1.5 s. Speakers were instructed to speak in their normal voice without pitch gliding. The best model order for the target vowel was chosen based on minimum variance of F1 and F2 frequencies over the middle portion of the vowel.

3.2.5 Offline formant analysis

The method for offline formant analysis is the same method reported in Munhall *et al.* (2009). The harmonicity of the power spectrum was used to estimate the vowel boundaries. The boundaries were inspected and corrected if necessary. Vowel formants (F1, F2, and F3) were estimated from the middle 40-80% of the vowel’s duration, with a 25 ms window that was shifted in 1 ms increments until the end of the middle portion of the vowel segment. A single average value for each of the formants was calculated from these sliding window estimates. Formant estimates were examined and were relabeled if incorrect (e.g. F2 being labelled as F1) or removed if the formant under examination was well beyond the distribution of other tokens.

3.2.6 Experimental phases

The session began with the model order estimation segment followed by the formant perturbation conditions. There were five formant perturbation conditions: the F1 of head was manipulated in the positive and negative directions (head+, head-), the F2 of hid was manipulated in the positive and negative directions (hid+, hid-) and the F2 of heed was manipulated in the negative direction (heed-). The order of the perturbation conditions was randomly assigned to each participant. Participants were given a passage to read with a five minute break to normalize their speech productions after each condition.

For the head conditions (head+, head-), speakers produced 125 utterances of the word “head” when a visual prompt was presented. These 125 trials were divided into five experimental phases. In the Acclimatization phase (utterances 1-15), participants received normal feedback. These utterances were discarded during analyses. In the Baseline phase (utterances 16-35), participants received normal feedback. In the Ramp phase, (utterances 36-50), the F1 value was increased or decreased by 50 Hz every 15 utterances (Ramp±50, Ramp±100, Ramp±150). In the Hold phase (Hold±200, utterances 81-105), the maximum ±200 Hz F1 perturbation was held constant. At utterance 106, the perturbation was removed and the participants received normal feedback until the end of the condition (End phase, End0; utterances 106-125). A schematic of the experimental phases for /ɛ/ can be seen in Figure 6. The maximum ±200 Hz perturbation of /ɛ/ was chosen based on previous studies that manipulated F1 of /ɛ/ (Purcell & Munhall, 2006; MacDonald *et al.* 2015; Mitsuya *et al.,* 2015)

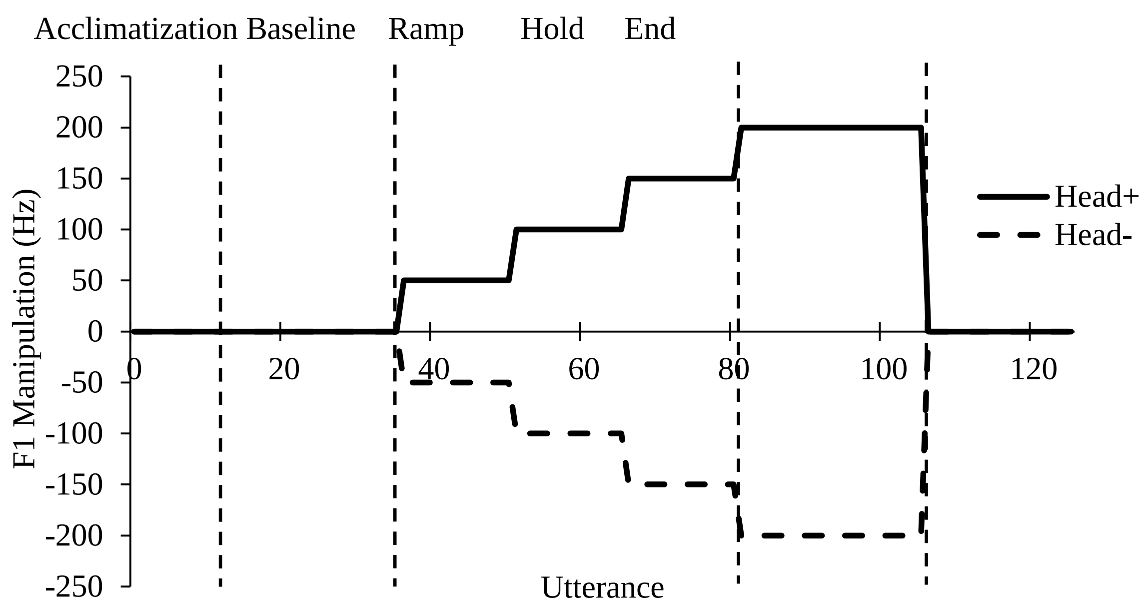


Figure 6. Schematic procedure of the feedback perturbation applied to the first formant of /ɛ/. The green line indicates the positive F1 manipulation. The red line indicates the negative F1 manipulation. The vertical dotted lines indicate the boundaries of the five experimental phases: Acclimatization, Baseline, Ramp, Hold, and End (from left to right).

For the hid conditions (hid+, hid-), speakers produced 125 utterances of the word "hid" when a visual prompt was presented. In the Acclimatization phase (utterances 1-15), participants received normal feedback. These utterances were discarded during analyses.

In the Baseline phase (utterances 16-35), participants received normal feedback. In the Ramp phase, (utterances 36-50), the F2 value was increased by 150 Hz or decreased by -100 Hz every 15 utterances (increased: Ramp+150, Ramp+300, Ramp+450; decreased: Ramp-100, Ramp-200, Ramp-300). In the Hold phase (Hold+600 or Hold-400, utterances 81-105), the maximum F2 perturbation was held constant by +600 Hz or -400 Hz for the positive and negative manipulations, respectively. At utterance 106, the perturbation was removed and the participants received normal feedback until the end of the condition (End phase, End0; utterances 106-126). A schematic of the experimental phases for /ɪ/ can be seen in Figure 7. The maximum increase (+600 Hz) and decrease (-400 Hz) manipulations of /ɪ/ were based on pilot studies that were completed to determine the maximum compensation magnitude observable for F2 manipulations. Through the pilot studies, it was determined that F2 compensation was asymmetrical, such that a positive manipulation required a larger manipulation to elicit maximum compensation.

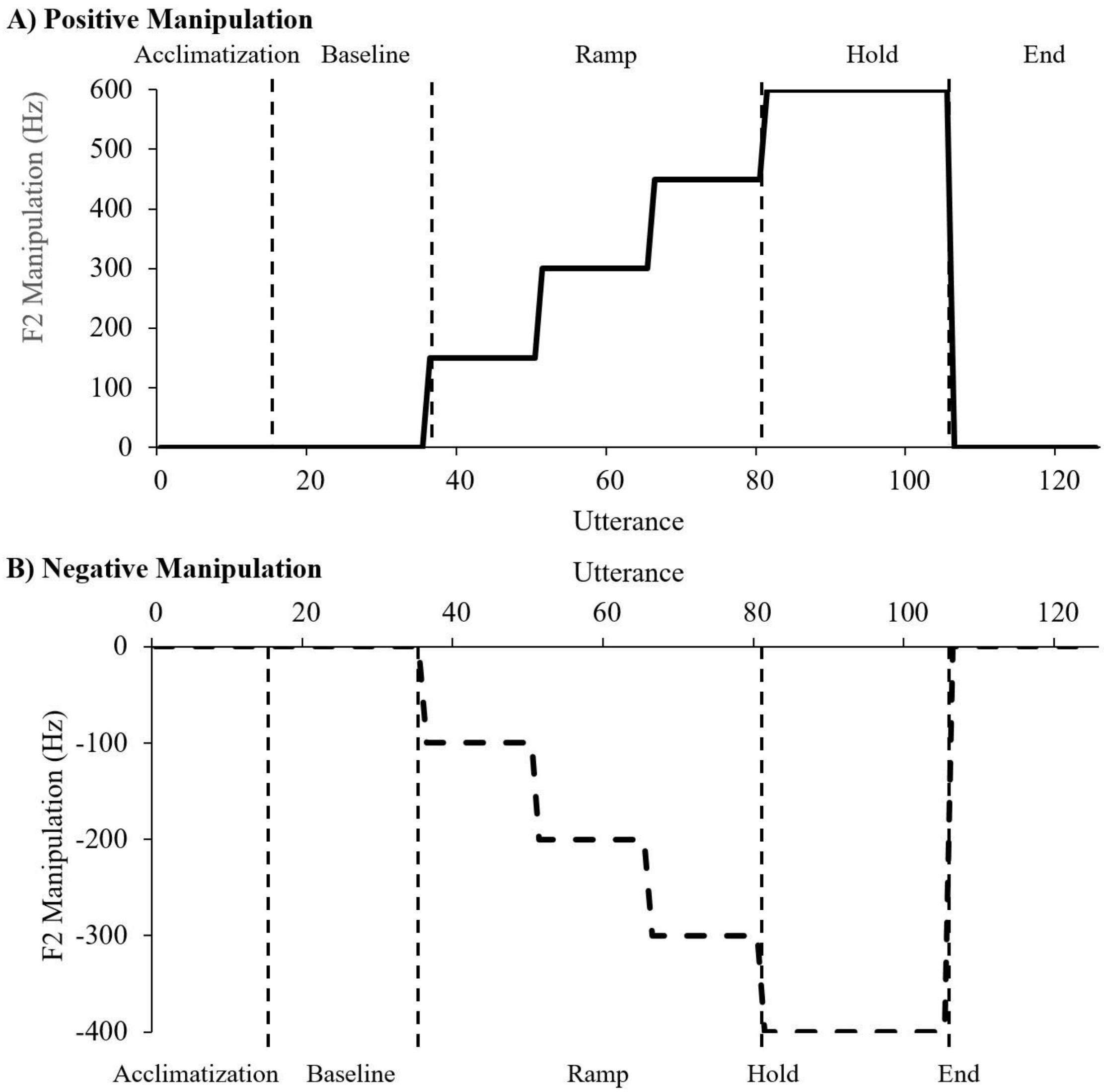


Figure 7. Schematic procedure of the feedback perturbation applied to the second formant of /ɪ/: A) Positive F2 manipulation; B) Negative F2 manipulation. The vertical dotted lines indicate the boundaries of the five experimental phases: Acclimatization, Baseline, Ramp, Hold, and End (from left to right).

For the heed condition, speakers produced 125 utterances of the word “heed” when a visual prompt was presented. These 125 trials were divided into five experimental phases. In the Acclimatization phase (utterances 1-15), participants received normal feedback. These utterances were discarded during analyses. In the Baseline phase (utterances 16-35), participants received normal feedback. In the Ramp phase, (utterances 36-50), the F2 value was decreased by -175 Hz every 15 utterances (Ramp-175, Ramp-350, Ramp-525). In the Hold phase (Hold-700, utterances 81-105), the maximum -700 Hz perturbation was held constant. At utterance 106, the perturbation was removed and the participants received normal feedback until the end of the condition (End phase, End0; utterances 106-125). A schematic of the experimental phases for /i/ can be seen in Figure 8. The maximum -700 Hz manipulation of /i/ was based on pilot studies that were completed to determine the maximum compensation magnitude. The study by Mitsuya *et al.* (2015) showed that /i/ was manipulated in the positive direction there were minimal changes in speech production. Therefore, in the current study /i/ was only manipulated in the negative direction.

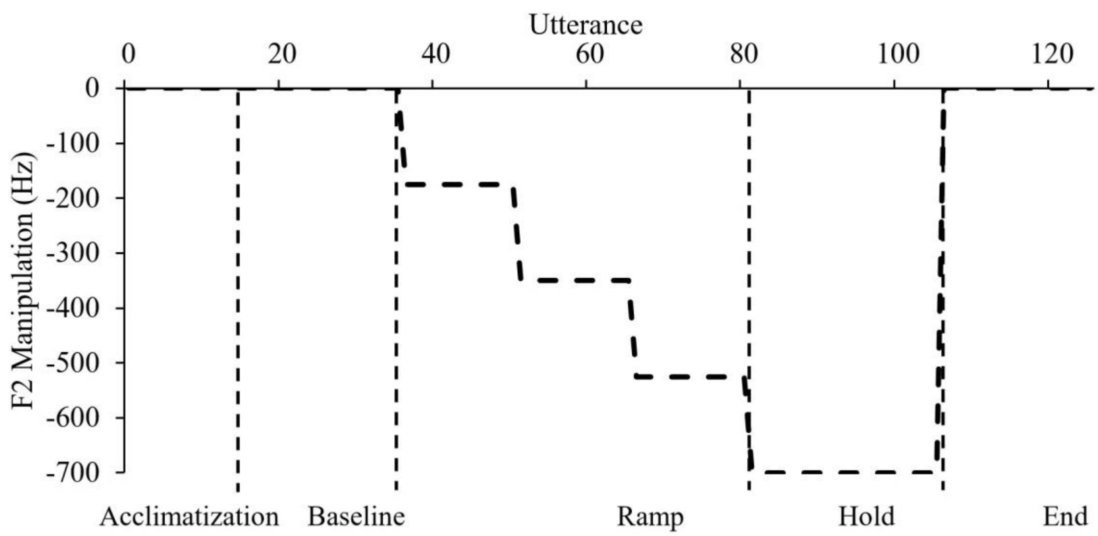


Figure 8. Schematic procedure of the feedback perturbation applied to the second formant of /i/. The vertical dotted lines indicate the boundaries of the five experimental phases: Acclimatization, Baseline, Ramp, Hold, and End (from left to right).

3.3 Results

3.3.1 Demographics

Table 1 includes the demographic information for the participants included in the analyses.

Table 1. Group demographics characteristics

Group	n		Age		BEPTA		HA experience		HL confirmed	
	Male	**F**	**M**	**SD**	**M**	**SD**	**M**	**SD**	**M**	**SD**
Older adults with hearing loss	10	10	71.55	6.25	47.80	11.80	12.62	11.90	12.90	11.94
Control: Older adults	7	12	69.06	5.79	16.33	7.82				
Control: Younger adults	8	15	25.13	3.32	2.59	3.97				

Notes: Age in years; BEPTA = better ear pure tone average between 500-4000 Hz in dB HL; HA experience = bilateral hearing aid experience in years; HL confirmed = hearing loss confirmed by a hearing professional in years; F = female; M = mean; SD = standard deviation

The mean and range of audiometric thresholds in dB HL as a function of frequency for the older adults with hearing aids are plotted in Figure 9. Figures 10 and 11 are the audiometric thresholds for the control groups: older adults and younger adults, respectively.

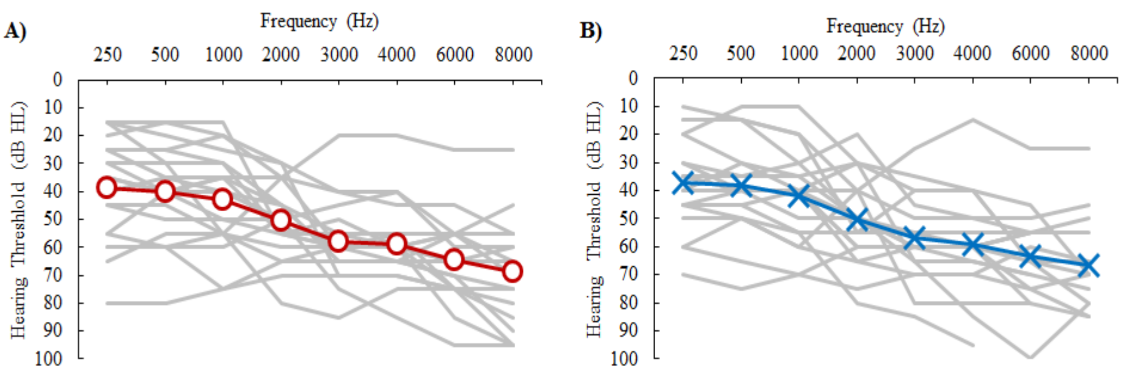


Figure 9. Air conduction thresholds in dB HL as a function of audiometric test frequency for older adults with hearing aids: A) Right Ear: Circle symbols indicate mean thresholds; B): Left Ear: X symbols indicate mean thresholds. Grey lines indicate individual thresholds.

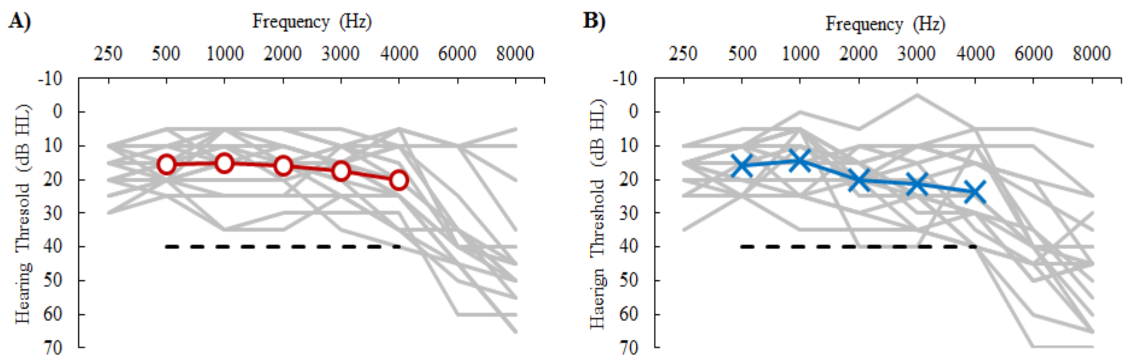


Figure 10. Air conduction thresholds in dB HL as a function of audiometric test frequency for control group with older adults: A) Right Ear: Circle symbols indicate mean thresholds; B) Left Ear: X symbols indicate mean thresholds. Grey lines indicate individual thresholds. Dashed line represents hearing threshold criteria for inclusion in group.

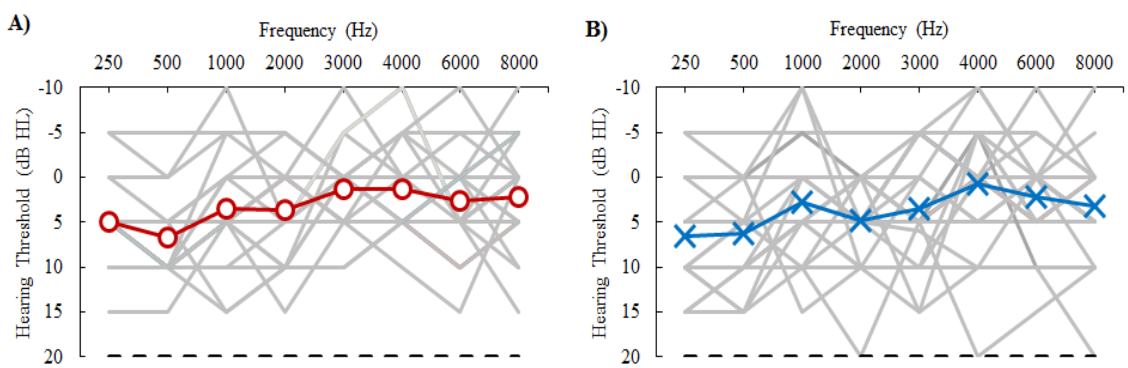


Figure 11. Air conduction thresholds in dB HL as a function of audiometric test frequency for control group with younger adults: A) Right Ear: Circle symbols indicate mean thresholds; B): Left Ear: X symbols indicate mean thresholds. Grey lines indicate individual thresholds. Dashed lines indicate hearing threshold criteria for group.

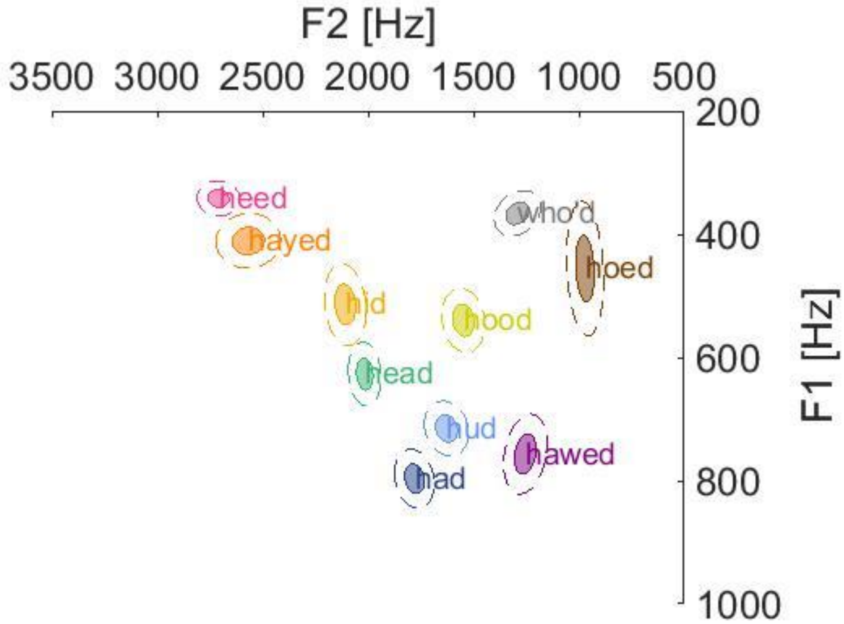
3.3.2 Vowel space

English vowel spaces were collected from all participants. In the vowel space figures, the center of each ellipse represents the mean F1 and F2 frequencies for that vowel, while the solid and dashed ellipses represent one and two standard deviations, respectively. The

F1/F2 values of the older adults with hearing aids are plotted in Figure 12A (female talkers) and Figure 12B (male talkers). The F1/F2 values of the control group with older adults are plotted in Figure 13A (female talkers) and Figure 13B (male talkers). The F1/F2 values of the control group with younger adults are plotted in Figure 14A (female talkers) and Figure 14B (male talkers). The F1 and F2 results for the groups are also found in Table 2 (vowels /i, ɪ, e, ɛ, æ/) and Table 3 (vowels /ɔ, u, o, ʊ, ʌ/).

Older adults with hearing aids:

A) Female talkers



B) Male talkers

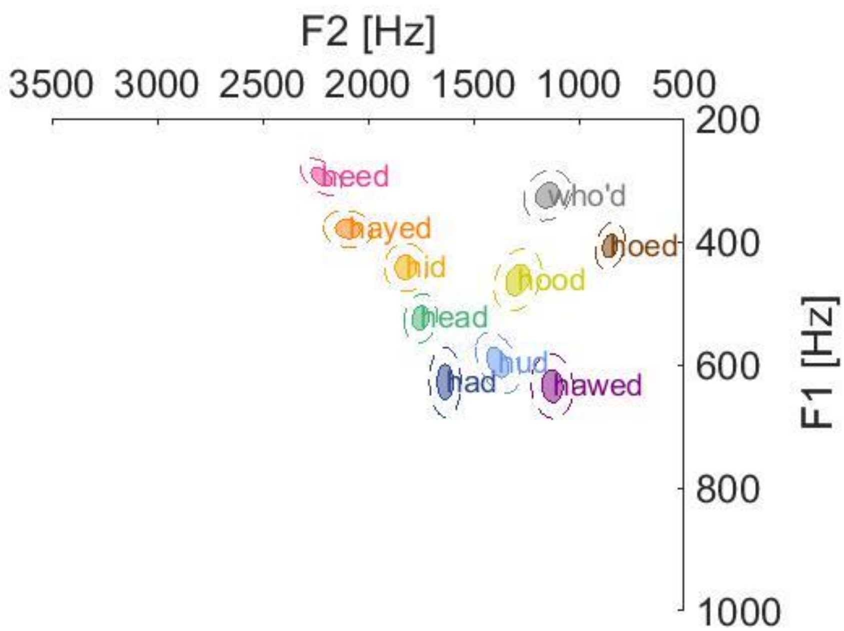
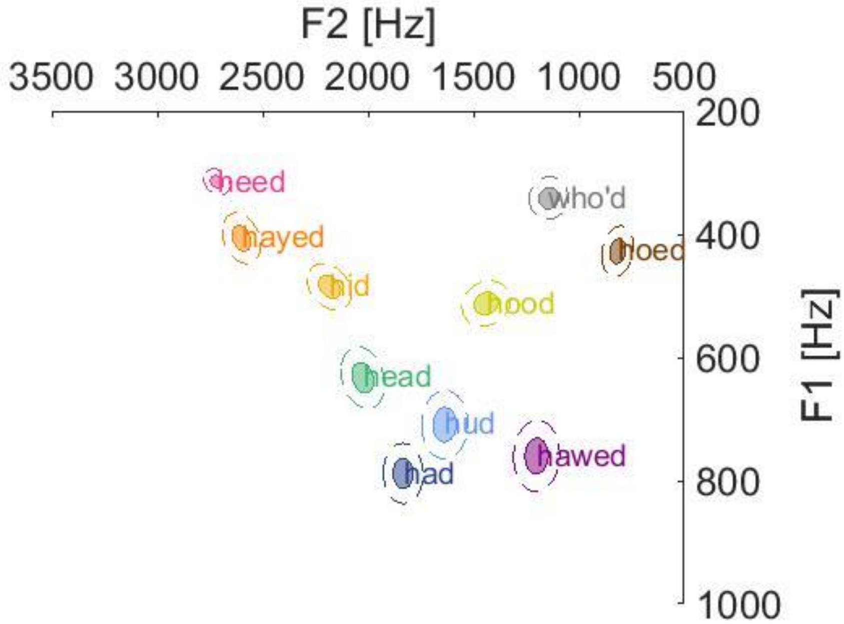


Figure 12. Vowel spaces in an /hVd/ context for older adults with hearing aids. A) Female talkers (n = 10); B) Male talkers (n = 10). The center of each ellipse represents the mean F1 and F2 frequencies. The solid and dashed ellipses represent one and two standard deviations, respectively.

Control group with older adults:

A) Female talkers



B) Male talkers

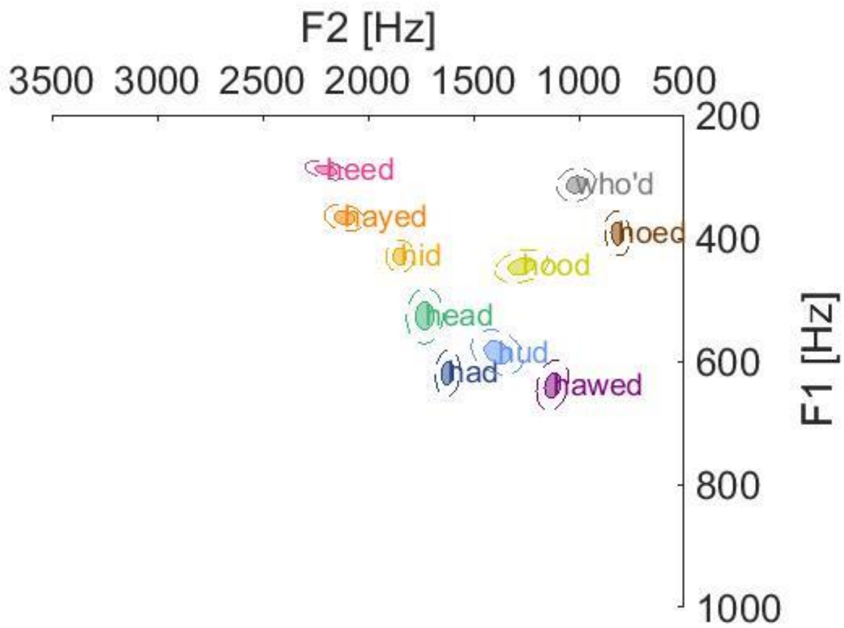
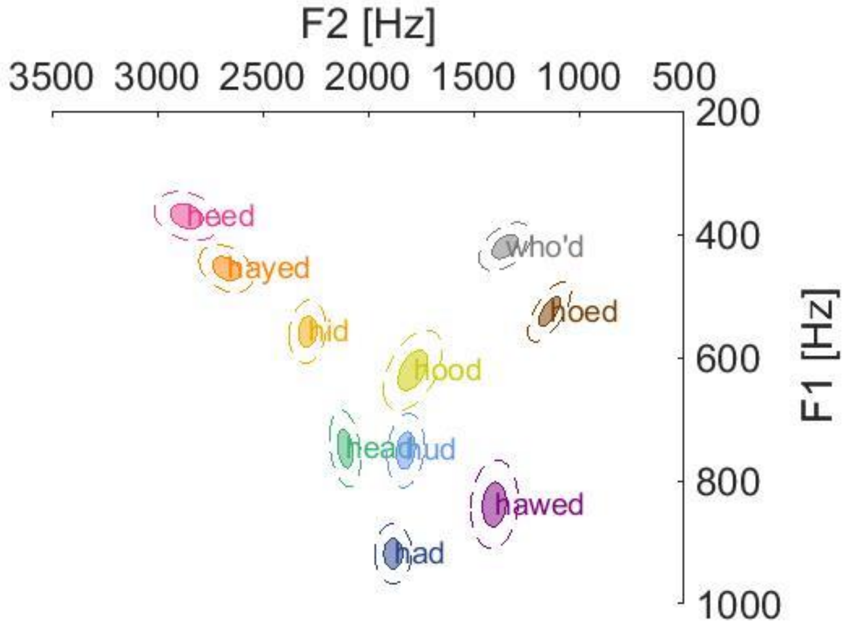


Figure 13. Vowel spaces in an /hVd/ context for the control group with older adults. A) Female talkers (n = 12); B) Male talkers (n = 7). The center of each ellipse represents the mean F1 and F2 frequencies. The solid and dashed ellipses represent one and two standard deviations, respectively.

Control group with younger adults:

A) Female talkers



B) Male talkers

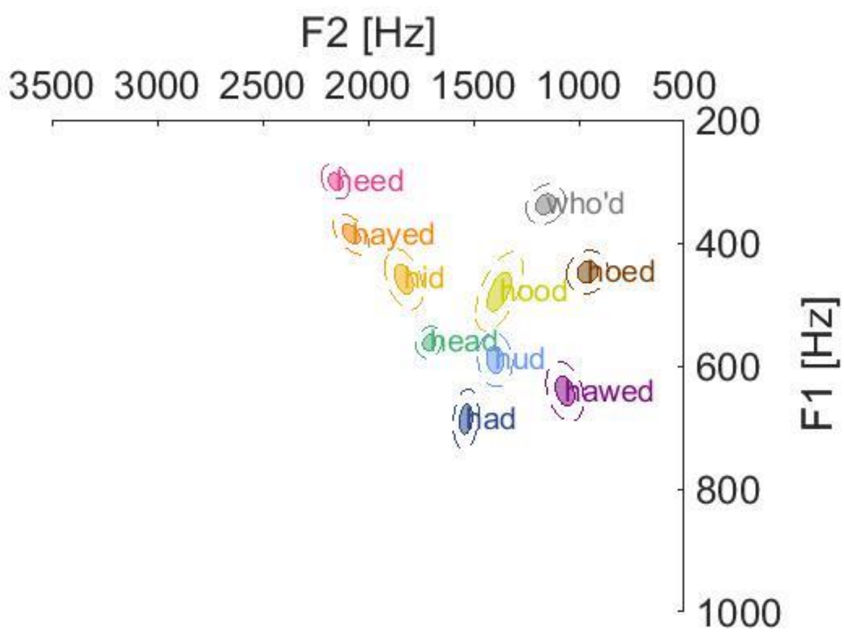


Figure 14. Vowel spaces in an /hVd/ context for the control group with younger adults. A) Female talkers (n = 15); B) Male talkers (n = 8). The center of each ellipse represents the mean F1 and F2 frequencies. The solid and dashed ellipses represent one and two standard deviations, respectively.

Table 2. Average F1 and F2 values for vowels with standard deviations in parentheses for /i. ɪ, e, ɛ and æ/. Vowels produced in an /hVd/ context for each group. Formants are reported in Hz.

	/i/		/ɪ/		/e/		/ɛ/		/æ/	
	F1	**F2**	**F1**	**F2**	**F1**	**F2**	**F1**	**F2**	**F1**	**F2**
Older adults with hearing aids										
Female	342.85 (32.17)	2709.74 (121.08)	520.63 (51.83)	2108.11 (110.88)	410.70 (41.52)	2567.60 (164.10)	626.40 (60.51)	2014.96 (128.30)	797.03 (66.31)	1779.44 (111.23)
Male	295.00 (45.55)	22221.40 (128.72)	442.04 (38.51)	1820.70 (140.68)	379.80 (44.64)	2095.60 (142.53)	524.62 (54.17)	1750.50 (142.37)	628.22 (55.48)	1633.70 (136.82)
Control group: older adults										
Female	315.10 (34.60)	2716.33 (194.06)	486.17 (50.09)	2183.21 (159.47)	405.10 (38.53)	2599.43 (180.33)	633.491 (56.36)	2024.09 (147.03)	788.05 (59.01)	1832.26 (162.86)
Male	289.94 (20.08)	2197.20 (213.83)	429.49 (29.57)	1848.30 (174.80)	367.04 (30.49)	2111.70 (217.96)	523.08 (52.54)	1730.50 (167.22)	619.59 (74.82)	1622.20 (127.71)
Control Group: younger adults										
Female	371.01 (49.21)	2860.26 (183.65)	558.48 (67.38)	2288.05 (146.85)	455.32 (39.94)	2671.88 (194.65)	745.81 (82.56)	2108.78 (125.13)	918.47 (86.12)	1881.47 (92.46)
Male	300.31 (30.91)	2153.89 (161.27)	457.88 (32.28)	1827.72 (126.51)	385.93 (40.13)	2685.53 (162.51)	561.76 (30.26)	1709.34 (125.71)	686.59 (45.57)	1535.35 (147.88)

Table 3. Average F1 and F2 values for vowels with standard deviations in parentheses for /ɔ, u, o, ʊ, and ʌ/. Vowels produced in an /hVd/ context for each group. Formants are reported in hertz.

	/ɔ/		**/u/**		**/o/**		**/ʊ/**		**/ʌ/**	
	F1	**F2**	**F1**	**F2**	**F1**	**F2**	**F1**	**F2**	**F1**	**F2**
Older adults with hearing aids										
Female	761.56 (65.78)	1250.49 (132.41)	365.72 (29.70)	1290.65 (179.63)	456.20 (59.27)	970.48 (103.63)	538.30 (57.38)	1547.78 (130.59)	715.21 (53.01)	1629.50 (155.45)
Male	631.16 (68.59)	1128.10 (123.65)	325.20 (30.87)	1148.40 (85.94)	406.89 (50.66)	850.73 (61.48)	462.81 (54.99)	1288.90 (70.84)	596.51 (66.43)	1384.30 (82.27)
Control group: older adults										
Female	762.36 (63.57)	1201.72 (109.76)	341.59 (24.23)	1142.56 (79.53)	428.60 (42.87)	815.48 (149.65)	512.33 (41.79)	1437.22 (146.01)	709.14 (67.18)	1636.00 (164.27)
Male	639.10 (48.34)	1123.20 (123.05)	313.26 (15.80)	1012.20 (71.74)	392.93 (23.23)	816.30 (73.02)	445.66 (23.49)	1270.90 (83.08)	585.59 (54.11)	1385.60 (124.74)
Control group: younger adults										
Female	838.71 (66.20)	1401.09 (124.65)	418.80 (43.32)	1349.34 (179.95)	528.39 (45.29)	1141.71 (103.57)	620.76 (68.42)	1784.43 (155.91)	750.67 (68.01)	1819.55 (128.13)
Male	641.41 (48.09)	1065.55 (117.12)	352.88 (54.98)	1157.38 (153.78)	446.06 (58.68)	956.94 (135.85)	479.20 (21.39)	1377.17 (143.00)	590.13 (38.00)	1396.67 (144.95)

Statistical analyses in this study were completed using SPSS (version 24; IBM, Armonk, NY). Separate mixed ANOVAs for F1 and F2 were conducted for males and females. Each mixed ANOVA had one within-subject factors (vowels: /i, ɪ, e, ɛ, æ, ɔ, o, u, ʊ/) and one between-subject factor (group; three levels: Older adults with hearing aids, Control: Older adults, Control: Younger adults). For all statistical analyses, Greenhouse-Geisser corrected degrees of freedom were used for interpretation of significant effects. Post hoc analyses, corrected for multiple comparison using Bonferroni corrections, were completed only when there was a significant interaction effect between vowels and group. Post-hoc analyses were not conducted when vowels were significant because vowels would have different formants based on their location in the vowel space. Results of the ANOVAs were interpreted at an α of 0.05.

F1 Differences in Males: The main effect of vowels was significant [$F(1.63, 34.15) = 98.31$, $p < 0.001$, $\eta_p^2 = 0.82$]. The main effect of group was non-significant [$F(2, 21) = 2.14$, $p = 0.14$, $\eta_p^2 = 0.17$). The interaction between vowels and group was non-significant [$F(3.25, 34.15) = 1.11$, $p = 0.36$, $\eta_p^2 = 0.10$]. There were no F1 differences across groups for males within each vowel.

F2 Differences in Males: The main effect of vowels was significant [$F(2.73, 54.51) = 431.197$, $p < 0.001$, $\eta^2_p = 0.96$]. The main effect of group was non-significant [$F(2,20) = 0.429$, $p = 0.66$, $\eta^2_p = 0.04$]. The interaction between vowels and group was non-significant [$F(5.45, 54.51) = 1.710$, $p = 0.14$, $\eta^2_p = 0.15$]. There were no F2 differences across groups for males within each vowel.

F1 Differences in Females: The main effects of vowels and group were significant [vowels: $F(3.93, 133.70) = 756.18$, $p < 0.001$, $\eta^2_p = 0.96$; group: $F(2, 34) = 13.69$, $p < 0.001$, $\eta_p^2 = 0.446$]. The interaction between vowels and group was significant [$F(7.87, 133.70) = 3.25$, $p = 0.002$, $\eta_p^2 = 0.160$]. Post hoc analyses revealed no significant F1 differences between female older adults with hearing aids and the control group with female older adults within each vowel. The older adults with hearing aids had lower F1 compared to younger adults for /e/ ($p = 0.001$), /ɛ/ ($p = 0.03$), /æ/ ($p = 0.001$). The older adults in the control group had lower F1 compared to younger adults for /i/ ($p = 0.004$),

/ɪ/ (p = 0.009), /e/ (p = 0.008), /ɛ/ (p = 0.001), /æ/ (p < 0.001), /o/ (p < 0.001), and /ʊ/ (p < 0.001).

F2 Differences in Females: The main effects of vowels and group were significant [vowels: $F(3.23, 109.80) = 942.60$, $p < 0.001$, $\eta^2_p = 0.97$; group: $F(2, 34) = 12.05$, $p < 0.001$, $\eta^2_p = 0.42$]. The interaction between vowels and group was significant [$F(6.46, 109.80] = 3.25$, $p = 0.005$, $\eta^2_p = 0.16$]. Post hoc analyses revealed the older adults with hearing aids had similar F2 values across vowels as the control group with older adults except for /o/ (p < 0.001). The older adults with hearing aids had lower F2 values than younger adults for /ɪ/ (p = 0.01), /ɔ/ (p = 0.01), /ʌ/ (p = 0.01), /ʊ/ (p = 0.001), and /o/ (p = 0.004). The control group with older adults had lower F2 values than younger adults for /ɔ/ (p = 0.001), /u/ (p = 0.005), /ʌ/ (p = 0.009), /ʊ/ (p < 0.001), and /o/ (p < 0.001).

3.3.3 Formant Manipulations

Statistical analyses in this study were completed using SPSS (version 24; IBM, Armonk, NY). The baseline average of F1 was calculated using the utterances from the Baseline phase (i.e. trials 21-40). The F1 values were then normalized by subtracting a speaker's baseline average from each utterance. To quantify a change in formant production, the average normalized F1 values during each phase was calculated. Separate ANOVAs were conducted for each vowel /ɛ, ɪ, i/ because the formant manipulations were different. For all statistical analyses, Greenhouse-Geisser corrected degrees of freedom were used for interpretation of significant effects. Results of the ANOVAs were interpreted at an α of 0.05. Post hoc analyses, adjusted for multiple comparisons by using Bonferroni corrections, were performed when there were significant results.

3.3.4 Manipulations of /ɛ/

The average F1 magnitude of compensation for /ɛ/ across phases and direction of manipulation for each group is shown in Figure 15.

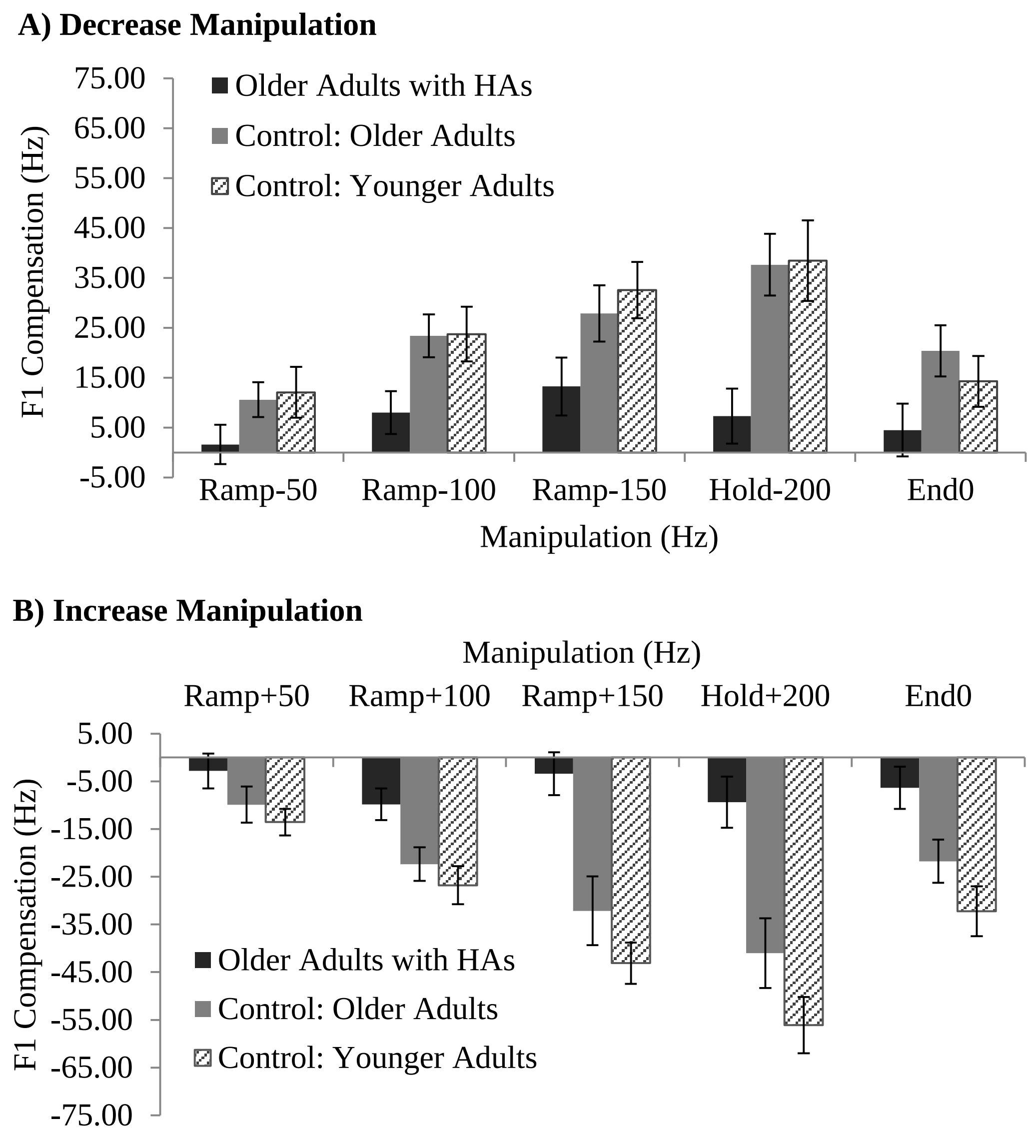


Figure 15. Average compensation of F1 across F1 manipulation phases for each group for /ε/ manipulation: A) Decrease manipulation; B) Increase manipulation. The error bars indicate ±1 standard error.

To compare the two directions of perturbation for /ε/, the magnitudes of compensations for each speaker in the increase condition for F1 and F2 were multiplied by -1. The multiplication factor was applied to both formants for consistency. A mixed ANOVA was performed with three within-subject factors (direction: increase, decrease; formants: F1, F2; phases: Ramp50, Ramp100, Ramp150, Hold200, End0) and one between-subject factor (group; three levels: Older Adults with hearing aids, Control: Older adults, Control: Younger adults). The main effects of direction and group were non-significant [direction: $F(1, 59)=2.88$, $p = 0.10$, $\eta_p^2 = 0.05$; group: $F(2, 3724.67) = 1.13$, $p = 0.33$, $\eta_p^2 = 0.04$]. The main effects for formants and phases were significant [formants: $F(1, 59) = 82.83$, $p < 0.001$, $\eta_p^2 = 0.58$; phases: $F(3.40, 200.67) = 5.33$, $p < 0.001$, $\eta_p^2 = 0.08$]. The four way interaction was non-significant [$F(5.33, 157.14) = 0.91$, $p = 0.48$, $\eta_p^2 = 0.30$]. The three way interaction between direction, formants and phases was non-significant [$F(2.66, 101.51) = 0.16$, $p = 0.90$, $\eta_p^2 = 0.00$]. The three way interaction between direction, phases and group was non-significant [$F(5.92, 174.69) = 0.77$, $p = 0.60$, $\eta_p^2 = 0.03$]. The three way interaction between direction, formants and group was non-significant [$F(2,59) = 1.35$, $p = 0.27$, $\eta_p^2 = 0.04$]. The only three way interaction that was significant was between formants, phases and group [$F(5.36, 158.03) = 4.31$, $p < 0.001$, $\eta_p^2 = 0.13$]. Post hoc analyses were completed to assess the effect of group across phases in F1 and F2, as well as the effect of phases across groups in F1 and F2.

Effect of group on phases. The older adults with hearing aids had smaller F1 changes than the control groups: older and younger adults at each of the phases. There were no group differences for F2 changes across phases.

Effect of phases on groups. Figure 16 illustrates the average magnitude of compensation across phases for each group for F1 and F2. The older adults with hearing aids had similar F1 changes across phases. In contrast, for both control groups, as the manipulation increases (i.e. as phases progressed), the magnitude of F1 compensation increased. The older adults with hearing aids and the control group with older adults had similar F2 changes across phases. The control group with younger adults had statistical significant differences, in which the manipulation phases (Ramp50, Ramp150, Hold200) had more F2 changes than End0.

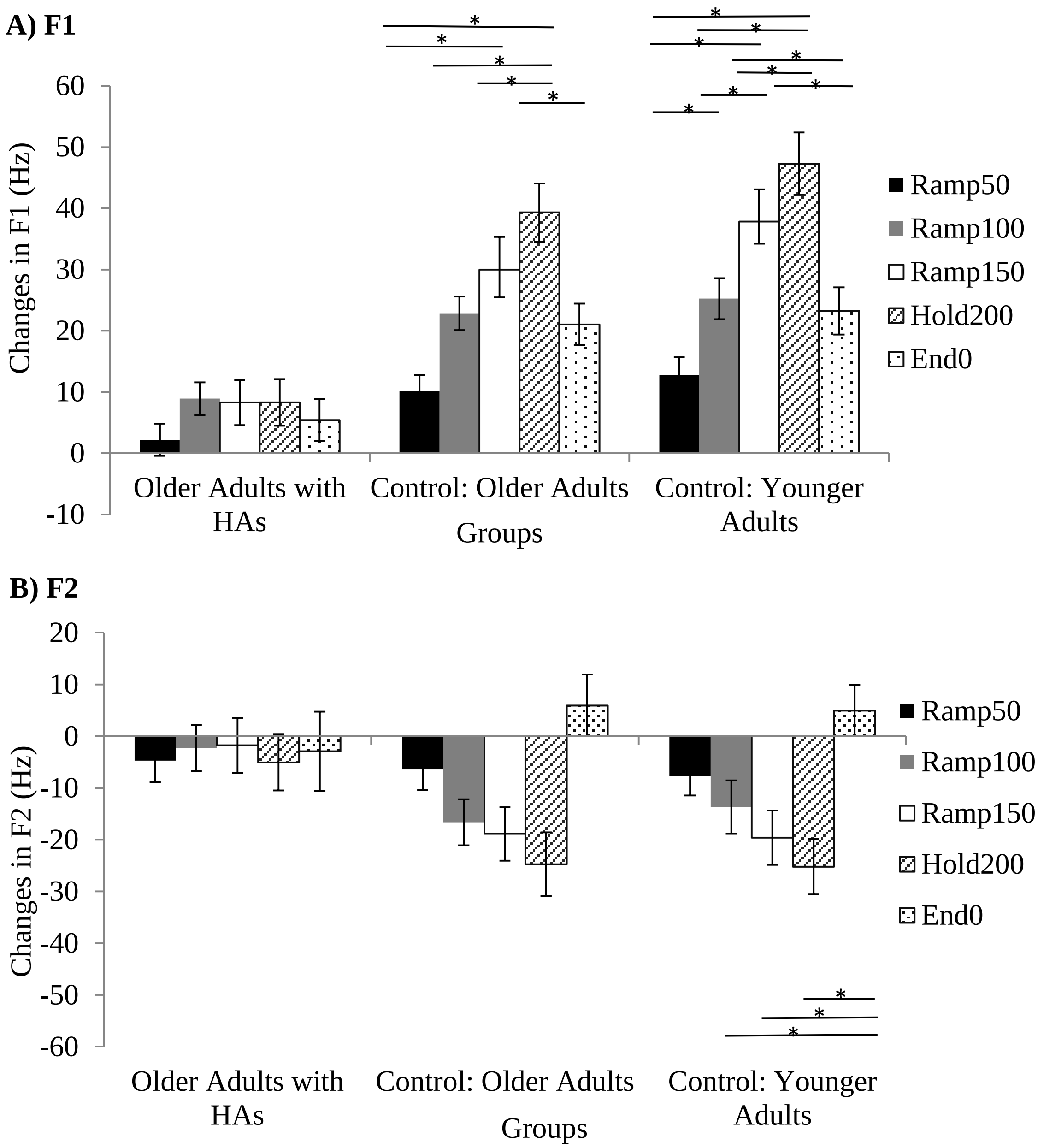


Figure 16. Average formant changes across F1 manipulation phases for each group for /ɛ/ manipulation: A) F1; B) F2. * indicates significant difference ($p \leq 0.05$). The error bars indicate ±1 standard error.

3.3.5 Manipulations of /ɪ/

The average F2 magnitude of compensation across phases and direction of manipulation for each group is shown in Figure 17. The F2 increase and decrease manipulations of /ɪ/

had different magnitudes across the phases so two mixed ANOVAs were performed. Post hoc analyses were completed when there were significant results.

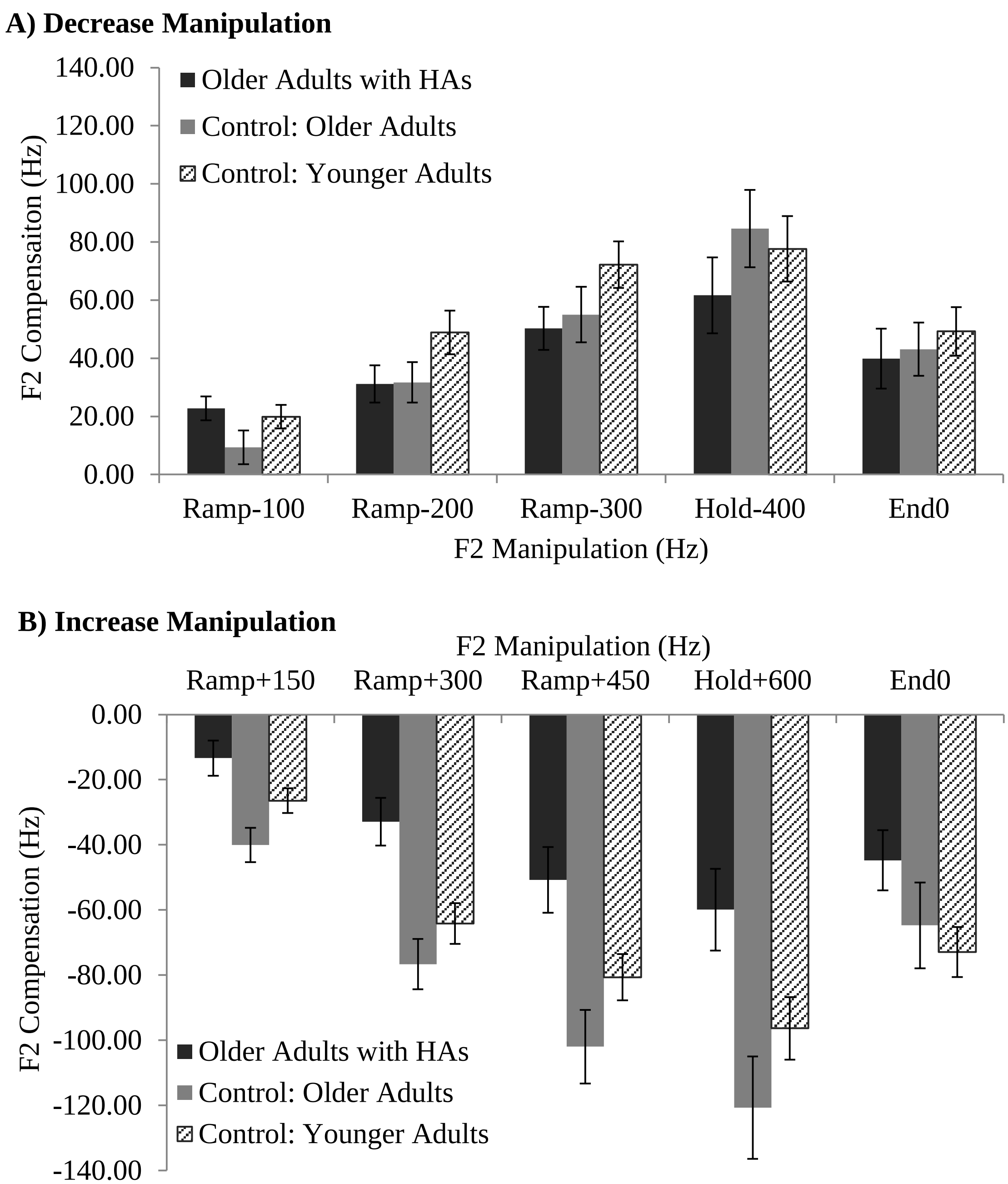


Figure 17. Average compensation in F2 across F2 manipulation phases for each group for /ɪ/: A) Decrease manipulation; B) Increase manipulation. The error bars indicate ±1 standard error.

3.3.5.1 Increase manipulation

The magnitude of change for each speaker in the increase condition for F1 and F2 was multiplied by -1. The multiplication factor was applied to both formants for consistency. A mixed ANOVA was performed with formants (two levels: F1, F2) and phases (five levels: Ramp+150, Ramp+300, Ramp+450, Hold+600, End0) as the two within subject factors and group (three levels: Older Adults with hearing aids, Control: Older adults, Control: Younger adults) as the between-subject factor. The main effects of formants, phases and group were significant [formants: $F(1, 59) = 168.46$, $p < 0.001$, $\eta_p^2 = 0.74$; phases: $F(2.66, 156.73) = 43.00$, $p < 0.001$, $\eta^2{}_p = 0.42$; group: $F(2, 59) = 6.04$, $p < 0.001$, $\eta_p^2 = 0.17$]. The three way interaction of formants, phases, and group was non-significant [$F(5.32, 156.96)$, $p = 0.07$, $\eta_p^2 = 0.07$].

The two way interaction between formants and phases was significant [$F(2.66, 156.96) = 53.74$, $p < 0.001$, $\eta_p^2 = 0.48$]. Post hoc analysis was completed to assess the effect of phases in F1 and F2 changes. Figure 18 illustrates the average magnitude of compensation across each phase for F1 and F2 and significant differences in the post-hoc analysis are illustrated within the figure. Overall, as F2 manipulation increased in size, greater F2 compensation occurred. There were non-significant F1 changes across phases.

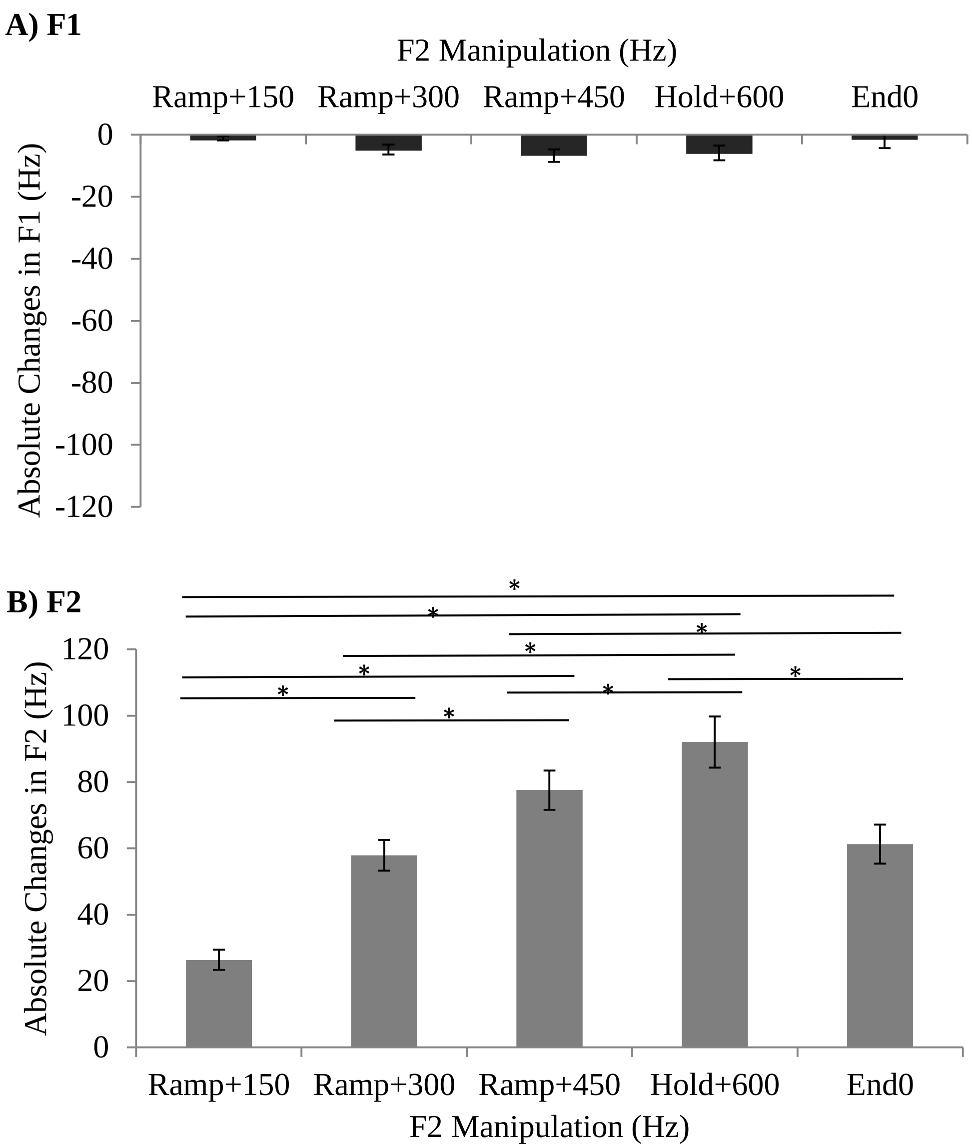


Figure 18. Average changes across F2 manipulation phases for /ɪ/ increase condition: A) F1; B) F2. * indicates significant difference ($p \leq 0.05$). The error bars indicate ±1 standard error.

The two way interaction between formants and group was significant [$F(2, 59) = 6.51$, $p < 0.001$, $\eta_p^2 = 0.18$]. Post hoc analysis was completed to assess the effect of group in F1 and F2 changes. There were no significant differences across groups in F1 changes. In comparison, the older adults with hearing aids had less F2 compensation than the control groups of older and younger adults.

The two way interaction between phases and group was significant [$F(5.31, 156.73) = 2.67$, $p = 0.02$, $\eta_p^2 = 0.08$]. Post hoc analysis was completed to assess the effect of phases across groups on formant changes. Overall, older adults with hearing aids had less formant changes than the control groups ($p \leq 0.05$) across formant manipulation phases. There were no group differences in the End0 phase.

3.3.5.2 Decrease manipulation

A mixed ANOVA was performed with formants (two levels: F1, F2) and phases (five levels: Ramp-100, Ramp-200, Ramp-300, Hold-400, End0) as the two within subject factors and group (three levels: Older adults with hearing aids, Control: Older adults, Control: Younger adults) as the between-subject factor. The main effects of formants and phases were significant [formants:

$F(1, 59) = 84.37$, $p < 0.001$, $\eta_p^2 = 0.59$; phases: $F(2.60, 153.16) = 31.21$, $p < 0.001$, $\eta_p^2 = 0.35$]. The main effect of group was non-significant [$F(2, 59) = 0.37$, $p = 0.70$, $\eta_p^2 = 0.01$]. The three way interaction between formants, phases and group was non-significant [$F(5.09, 150.22) = 1.54$, $p = 0.18$, $\eta_p^2 = 0.50$]. The two way interaction between formants and group was non-significant [$F(2, 59) = 0.82$, $p = 0.44$, $\eta_p^2= 0.03$]. The two way interaction between phases and group was non-significant [$F(5.19, 153.16) = 1.33$, $p = 0.25$, $\eta_p^2 = 0.04$].

The only significant two way interaction was between formants and phases [$F(2.55, 150.22) = 35.46$, $p < 0.001$, $\eta_p^2 = 0.38$]. Post hoc analysis was completed to assess the effect of phases in F1 and F2 changes. Figure 19 illustrates the average magnitude of compensation across each phase for F1 and F2. Overall, as F2 manipulation increased in size, greater F2 compensation occurred. There were no significant F1 differences across phases.

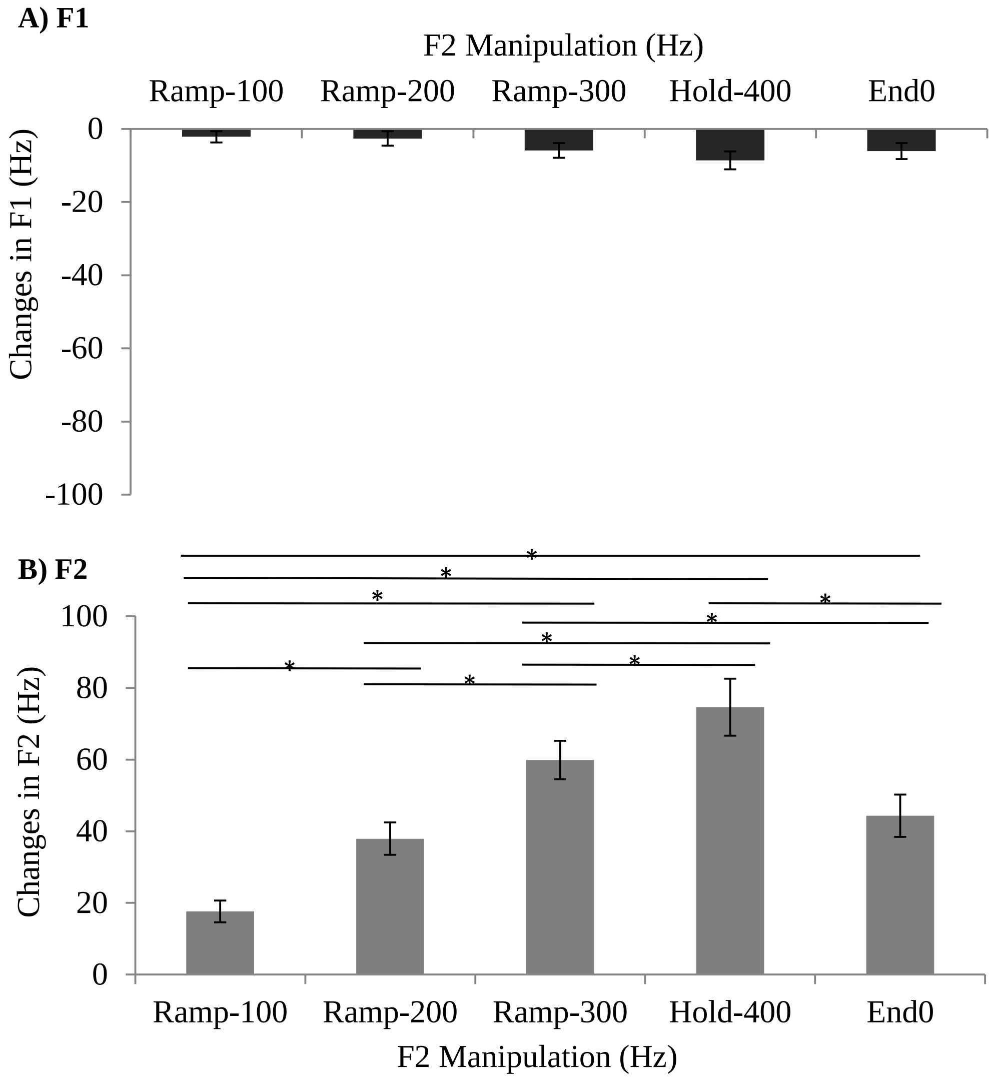


Figure 19. Average changes across F2 manipulation phases for /ɪ/ decrease condition: A) F1; B) F2. * indicates significant difference (*p* $\leq$ 0.05). The error bars indicate ±1 standard error.

3.3.6 Manipulation of /i/

The average F2 magnitude of compensation across phases for each group is shown in Figure 20.

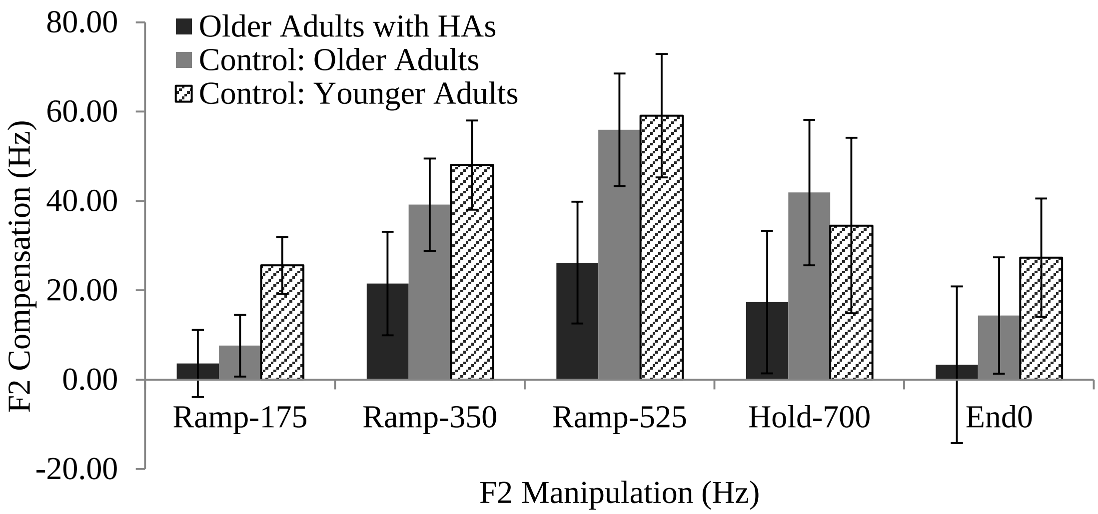


Figure 20. Average compensation in F2 across F2 manipulation phases for each group for /i/ manipulation. The error bars indicate ±1 standard error.

A mixed ANOVA was performed with formants (two levels: F1, F2) and phases (five levels: Ramp-175, Ramp-350, Ramp-525, Hold-700, End0) as the two within subject factors and group (three levels: Older adults with hearing aids, Control: Older adults, Control: Younger adults) as the between-subject factor. The main effects of formants and phases were significant [formants: $F(1, 59) = 17.74$, $p < 0.001$, $\eta_p^2 = 0.23$; phases: $F(2.09, 123.57) = 12.30$, $p < 0.001$, $\eta_p^2 = 0.17$]. The main effect of group was non-significant [$F(2, 59) = 0.91$, $p = 0.41$, $\eta_p^2 = 0.03$]. The three way interaction between formants, phases and group was non-significant [$F(4.12, 121.58) = 0.55$, $p = 0.71$, $\eta_p^2 = 0.02$]. The two way interaction between phases and group was non-significant [$F(4.19, 123.57) = 0.94$, $p = 0.45$, $\eta_p^2 = 0.03$]. The two way interaction between formants and group was non-significant [$F(2, 59) = 2.142$, $p = 0.13$, $\eta_p^2 = 0.07$].

The only two way interaction that was significant was between formants and phases [$F(2.06, 121.58) = 9.73$, $p < 0.001$, $\eta_p^2 = 0.14$]. Post hoc analysis was completed to assess the effect of phases in F1 and F2 changes. Figure 21 illustrates the average magnitude of compensation across each phase for F1 and F2. Overall, as F2 manipulation increased in size, greater F2 compensation occurred, except Ramp-525 had greater F2 changes than

Hold-700. Even though there was some significant F1 change differences across phases, it was very small (less than 10 Hz) compared to the F2 compensation.

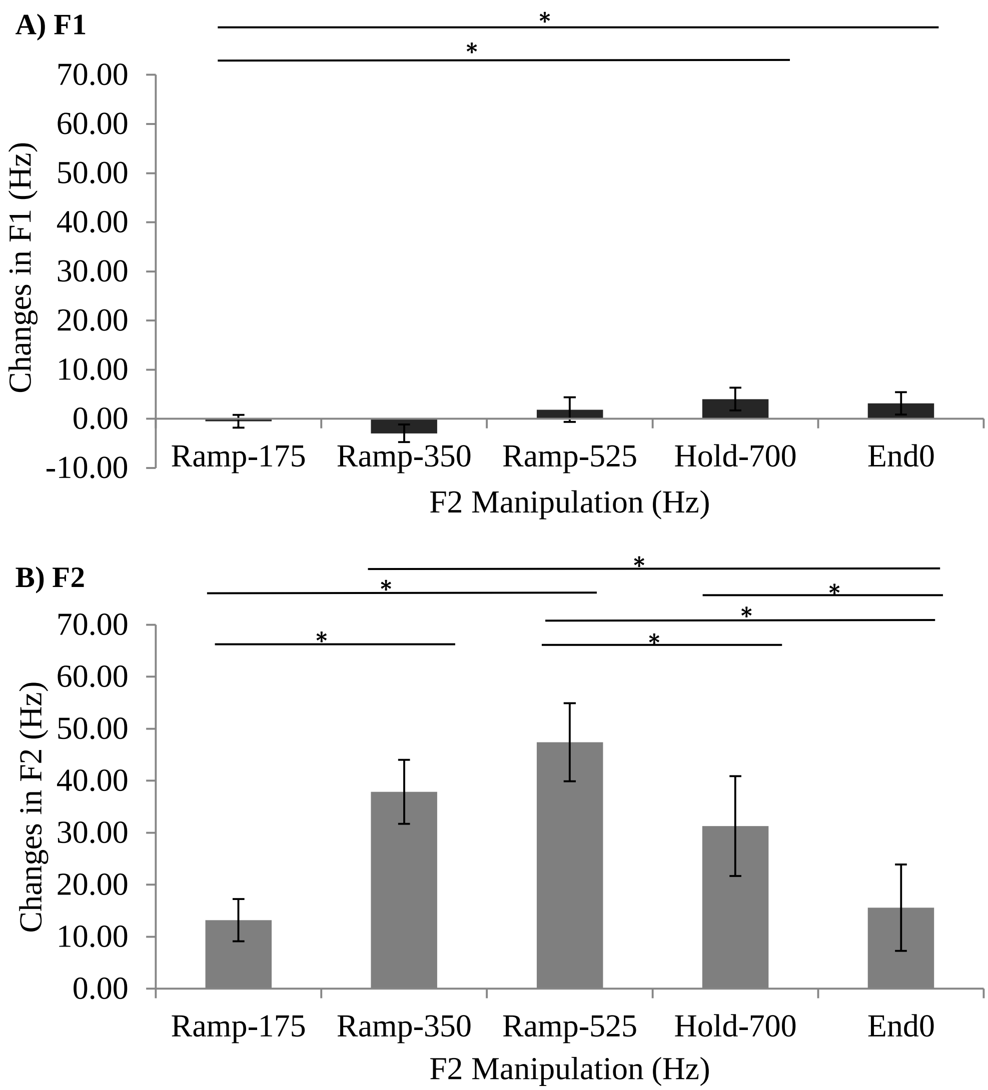


Figure 21. Average changes across F2 manipulation phases for /i/ manipulation: A) F1; B) F2. * indicates significant difference ($p \leq 0.05$). The error bars indicate ±1 standard error.

3.4 Discussion

The speech motor control system uses auditory feedback to detect errors and regulate speech production. If a person has an impairment in the auditory system, they may not be able to use their auditory feedback cues effectively. With the use of hearing aids, people with hearing loss should be better able to use these auditory cues and possibly able to detect errors in their speech. The purpose of the study was to compare talkers' compensatory formant production in response to formant manipulations between older and younger adults with normal hearing and older adults with hearing loss when aided with binaural hearing aids.

3.4.1 Aging effects

There are physiological and anatomical effects of aging that may affect older adult's vowel productions, such as lowering of the larynx and muscle atrophy in the pharynx and tongue (Benjamin, 1997; Linville & Fisher, 1985; Xue & Hao, 2003). A common effect from these changes is lowering of formant values (Benjamin, 1982; Endres *et al.*, 1971; Harrington *et al.*, 2007; Linville & Fisher, 1985; Liss *et al.*, 1990; Torre & Barlow, 2009; Xue & Hao, 2003; Zraick *et al.*, 2006). In the current study, the three groups of male participants: older adults with hearing aids, older adults with normal hearing and young adults with normal hearing did not significantly differ in F1 and F2 values across all vowels. In contrast, the older females had significantly lower F1 and F2 values than younger females for some of the vowels. Females may be more susceptible to aging effects in vowel productions because females have higher formants and have more hormonal changes that occur with aging (Sataloff *et al.*, 1997). As females age, estrogen levels decrease which causes changes in the mucous membrane linings of the vocal tract and other muscles. These muscle changes are reflected in the voice characteristics of older females, like masculinization of the voice. Thus, the physical changes that occur with aging influence vowel productions.

It was hypothesized that the cognitive and anatomical changes that occur with aging may affect an older adult's ability to respond to formant perturbations compared to younger adults. The current results had no significant differences between younger and older adults with normal hearing for all vowel perturbation conditions. The older adults responded to the formant perturbations similar to the younger adults, such that their compensation was in the opposite direction of the manipulation and displayed similar magnitudes of compensation. These results suggest that the speech motor control system may not be affected by aging. Aging effects in the current study may not have been found due to the large age range in the control group from 55 to 80 years of age. Future studies may want to separate the group into smaller age categories (e.g. 50 to 59 years, 60 to 69 years, and so on) and include older adults above 80 years of age.

3.4.2 Effects of hearing loss and hearing aids

The speech motor control system is affected by an impairment in the auditory system. The results showed that hearing aid users had less formant compensation than talkers with normal hearing for some vowel conditions, specifically the positive and negative F1 manipulations of /ɛ/ and the positive manipulation of /ɪ/. These results suggest that the hearing aid users may be using a different feedback system to detect errors, such that auditory feedback errors may not play as an important role compared to individuals with normal hearing. Nasir and Ostry (2008) studied speech learning in cochlear implant recipients with their implants turned off by altering somatosensory feedback. They used a robotic device to change the position of the jaw while the participant said /s/-initial words. The cochlear implant users showed compensation to the sensorimotor perturbation similar to individuals with normal hearing. Furthermore, the study by Laugesen *et al.* (2009) suggested that some hearing aid users use their sensorimotor feedback to monitor and change their speech intensity. These results suggest that individuals with hearing loss use feedforward commands or sensorimotor feedback to regulate speech production.

The use of auditory feedback may be different between hearing aid users and individuals with normal hearing. Hearing aid users had less F1 compensation compared to the normal hearing individuals for /ɛ/. This suggests for the vowel /ɛ/, hearing aid users may be using a different or altered feedback system to monitor for speech production accuracy compared to normal hearing individuals. In comparison, for the vowel /i/, there were no group differences. This suggests that the hearing aid users used their auditory feedback similarly as normal hearing individuals. The weighting of the auditory feedback system on detecting speech errors may be different across different vowels for hearing aid users compared to normal hearing individuals. This may be due to a relationship between absolute formant frequencies and hearing loss. Future studies should extend the current study to other vowels to determine how the speech motor control changes with hearing loss.

Another possible explanation why hearing aid users have less compensation may be due to limitations of hearing aid technology. Hearing aids generally have more impact on sounds above 1000 Hz because more hearing loss occurs above 1000 Hz (Nicolaidis & Sfakiannaki, 2007). As well, the main goal of hearing aids is to amplify sounds from the environment (Dillion. 2012). Digital signal processing algorithms in hearing aids classify the auditory scene into four main categories: music, noise, speech in quiet or speech in noise (Büchler *et al.*, 2005; Kerckhoff, Listenberger, & Valente, 2008). There are also digital signal processing algorithms that help with high-frequency audibility, such as extended bandwidth or frequency lowering technology (Brennan *et al.*, 2014; Glista *et al.*, 2009; Kreisman *et al.*, 2010). All these digital signal processing algorithms may not be directly improving the person's ability to hear their own voice. The speech coding strategies in hearing aids may have limitations that prevent hearing aid users from detecting auditory feedback errors, especially in the lower frequency region. For example, the F1 of head is relatively low (around 600 Hz) and hearing aid users in the current study were unable to compensate for the F1 perturbations of /ɛ/ during the Ramp and Hold phases (see Figure 16). Thus, there is a need for digital signal processing in hearing aids to better focus on the hearing aid user's own voice, especially since approximately 30% of hearing aid users are not satisfied with their own voice (Kochkins, 2010).

The auditory system is complex such that an impairment cannot be easily fixed with amplification devices. Hearing aids cannot restore the auditory system of an individual with hearing loss to make it similar to an individual with normal hearing. For example, outer hair cells in the cochlea are usually damaged in individuals with hearing loss. This causes the auditory filters in the cochlea to be broader and flatter, which results in a reduction in frequency selectivity (Dubno & Dirks, 1989, Peters & Moore, 1992; Glasberg & Moore, 1986). As well, cochlear damage can also affect loudness and pitch perception, frequency discrimination and/or temporal processing [see Moore (1996) for review]. Hearing aids cannot restore outer hair cells or other damage, and hearing aid users will still receive degraded speech input from their hearing aids. This may have caused the hearing aid users to have less compensation than individuals with normal hearing. It will be interesting for future studies to use the current altered auditory feedback paradigm to include unaided conditions to compare the differences in compensation between aided and unaided.

The perturbation paradigm has shown that speech motor control is sensitive to impairments. For example, Mollaei and colleagues (2013) found older adults with Parkinson's Disease had less compensation to a F1 manipulation of /ɛ/ compared to healthy, age-matched adults. Results from the current study showed that a hearing loss also affects the speech motor control system. Talkers with Parkinson's Disease and hearing aid users have reduced capacity to adapt to a change in auditory feedback. Future studies may want to include other talkers with different diseases or different configurations or degrees of hearing loss to determine how the speech motor control system uses auditory feedback cues.

3.4.3 Interaction of feedback and other systems

Place of constriction for vowel production may play an important role in compensatory patterns. There were no significant group differences for /i/. Also, the highest percent compensation for the F2 manipulation of /i/ was 9%, whereas, the highest percent compensations for the F2 manipulation of /ɪ/ were 15% and 18.5% for the increase and decrease directions, respectively. This suggests that the vowel /i/ is unique. The vowel /i/ is a high, front, and closed vowel. The articulatory movements for /i/ may be limited because the body of the tongue is already in a high position and the mouth is relatively closed (Perkell & Nelson, 1985). Motor commands and somatosensory feedback may play a stronger role for this vowel. The study by Mitsuya *et al.* (2015) also suggests that /i/ may be more controlled by somatosensory feedback as their F1 manipulation of /i/ in the negative direction resulted in minimal F1 changes. Thus, a hearing loss may have minimal effects on the detection of acoustic feedback errors for /i/ as somatosensory feedback plays a stronger role. This may provide a possible reason why there were no group differences and less percent compensation for /i/ compared to other vowels in the current study. As well, Mitsuya *et al.* (2015) also found other corner vowels /u/, /ɔ/ and /æ/ had less F1 compensations compared to /ɛ/. This suggests that corner vowels may be less controlled by auditory feedback. Future studies may want to look at other corner vowels to see if there are group differences between individuals with hearing loss and normal hearing.

The perturbations and compensations for the increase and decrease conditions of /ɪ/ are asymmetric. To elicit large F2 compensations for /ɪ/ in the current study, a perturbation of magnitude of 600 Hz in the increase condition and 400 Hz for the decrease direction had to be used. Further, Mitsuya *et al.* (2015) found when they perturbed F1 of /ɪ/ by 200 Hz, they had ~10 Hz of F1 compensation in the decrease condition and ~45 Hz of F1 compensation in the increase direction. This suggests that the direction of error in the vowel /ɪ/ matters. Studies by Bohn and Polka (2001), Polka and Bohn (1996, 2003), Polka and Werker (1994), Swoboda *et al.* (1978) have shown that the direction of change within the vowel space plays a key role on salience of the perceived change. For example, the direction of change from /ɪ/ to /i/ is easier to discriminate than /i/ to /ɪ/ (Swoboda *et al.*, 1978). Further investigation with the vowel /ɪ/ is required to understand how this vowel is perceived and controlled.

Manipulation of F1 resulted in significant changes in F1 productions and non-significant changes in F2. Similarly, a manipulation of F2 resulted in significant changes in F2 productions and non-significant changes in F1. This pattern of independent formant control by the speech motor control system is similar to other studies by Villacorta *et al.* (2007) and MacDonald *et al.* (2011). The speech motor control system is able to parse out F1 and F2 and detect perturbations within each formant and correct for it, without affecting the other formants. This occurred for all groups of participants: hearing aid users and older and younger adults with normal hearing. This suggests that this ability of the speech motor control system is not affected by hearing loss or hearing aids. Studies by MacDonald *et al.* (2010) and Munhall *et al.* (2009) have manipulated F1 and F2 concurrently in young adults with normal hearing. Their results have also showed that F1 and F2 compensations are independent from each other. Concurrent manipulations may require more cognitive or other processes for the speech motor control system. Future studies may also want to manipulate F1 and F2 concurrently to determine if there are aging and hearing loss effects.

3.4.4 Conclusion

Research using altered auditory feedback may offer insight on robust and critical acoustic cues that are important for speech production. Formant compensation patterns across the different vowels: /ɛ/, /ɪ/ and /i/ suggest each vowel may be regulated differently. The speech motor control system in older adults and younger adults with normal hearing reacted similarly to formant perturbations, even though there was an aging effect on actual vowel productions. This suggests that the speech motor control system might be robust to aging effects. However, results found that older adults with hearing aids have less formant compensation than their age-matched peers and younger adults with normal hearing. Hearing aid users may be relying on other feedback systems, such as somatosensory feedback. As well, there may be limitations of hearing aids and permanent effects of hearing loss on the auditory system that may prevent the user to hear their voice adequately. Future studies are needed to further investigate how the speech motor control system regulates speech production and how hearing loss and hearing aids affect the speech motor control system.

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Chapter 4

4 Compensatory responses to intensity-shifted auditory feedback in individuals with normal hearing and hearing loss

4.1 Introduction

Being able to control one's own speech intensity is important for speech production. For instance, a talker may modulate their speech intensity to attract or reduce attention to themselves when in conversations with others (Bauer *et al.*, 2006). A person may also need to increase or decrease their speech intensity output based on environmental noises or distances from other talkers (Lane & Tranel, 1991). In combination with fundamental frequency, speech intensity helps with conveying meaning in speech by segmenting messages or placing emphasis on certain syllables or words (Hafke-Dys, Preis, & Kaczmarek, 2013). Clinically, it is important to understand the mechanisms involved with intensity control to prevent and treat disorders that are characterized by abnormal speech intensity. For example, individuals with Parkinson's Disease usually have monotonous speech with low intensity (Logemann *et al.*, 1978, Ramig, 1994) and individuals with spastic dysarthria or laryngeal dystonia have speech with unstable or variable speech intensity (Griffiths & Bough, 1989).

4.1.1 Hearing loss and own-voice intensity control

Being able to control one's own speech intensity is dependent on feedback mechanisms, such as auditory feedback. Hearing loss and the use of amplification devices may introduce problems in the intensity-related auditory feedback system. Leder *et al.* (1987) had postlingually deafened men with profound bilateral sensorineural hearing losses and men with normal hearing read the Rainbow Passage. Their results showed that the speech intensity was higher and the intensity fluctuated more in men with hearing loss. After a cochlear implant, Leder and Spitzer (1990) showed there was a reduction in speech intensity in a similar task as Leder *et al.* (1987). Similarly, Perkell *et al.* (1992) showed a reduction in speech intensity of vowels following activation of cochlear implants. Lane *et* *al.* (1997) showed a reduction in variability for amplitude contours after cochlear implantation. The role of hearing appears to be important in the regulation of self-monitoring of speech intensity.

Even with amplification devices, hearing impaired users may still have concerns with their own intensity. New hearing aid users may have difficulties with their own speech intensity level control as they perceive their speech to be too loud (Laugesen *et al.*, 2008). However, around 25% of experienced hearing aid users are unsatisfied with the sound of their own speech and this may affect their usage and satisfaction with the hearing aids (Kochkin, 2010). Laugesen and colleagues (2011) developed the Own Voice Qualities questionnaire and showed that hearing aid users that have open fitting or small vent fittings with at least one year's experience with hearing aids have more own-voice concerns than individuals with normal-hearing. The hearing aid users had concerns with determining the correct own-voice level for different conversation situations, being able to hear and speak at the same time and whispering. Digital signal processing in the different listening programs of the hearing aids, such as dynamic range compression or noise management algorithms may change how the hearing aid users perceive their speech loudness. Thus, the altered auditory feedback created by hearing aids may be responsible for concerns the hearing aid users have with their own speech intensity and loudness.

There is very little research on how hearing aid users control and perceive their own speech intensity levels. At the time of writing, a study by Laugesen *et al.* (2009) was the only study that measured speech intensity in different talking situations in hearing aid users. The study showed that hearing aid users increased their speech intensity level as the distance with their conversation partner increased. However, differences in intensity levels across the hearing aid users occurred. Some hearing aid users increased their speech intensity similarly to individuals with normal hearing. While other hearing aids users differed in their speech intensity growth rates with different distances from individuals with normal hearing. The authors suggest that the latter group of hearing aid users may have not relied on their auditory feedback system but developed another strategy for controlling their own speech intensity levels with proprioceptory feedback.

These hearing aids users may not have used their auditory feedback system because their hearing loss and processed sounds from their hearing aids have changed their auditory feedback so much that they unconsciously do not use the auditory cues anymore. Further investigations are needed to determine how hearing aid users control their own speech intensity.

4.1.2 Lombard and sidetone amplification effects

In the literature, there are two main methods to study own speech intensity control: Lombard effect and side tone amplification. The Lombard effect occurs when there is noise present in the auditory environment, and an individual will typically raise their speech intensity [Lombard (1911) as cited in Lane and Tranel (1971)]. The sidetone amplification effect occurs when the individual's speech intensity is amplified, and the individual will typically lower their speech intensity. Even though the Lombard and sidetone amplification effects both affect the self-perception of speech loudness, the underlying processes of the effects are different from each other. Siegel *et al.* (1976) showed that sidetone amplification effect was affected by age, such that adults and 4 year old children responded to the sidetone amplification but children who were 3 year old did not. In contrast, the Lombard effect occurred in all age groups. Additionally, a study by Siegel *et al.* (1982) showed that performance with the Lombard effect did not predict performance with sidetone amplification.

Early Lombard and sidetone amplification studies asked participants to read a passage or have a conversation such that participants were continuously talking throughout the manipulations. Lane and Tranel (1971) reviewed early studies that depicted the Lombard and sidetone amplification effects. They have summarised that in previous studies, individuals respond to the perturbation or environmental noise. Specifically, in sidetone amplification studies, subjects would compensate in the opposite direction of the perturbation, such that for every 2 dB increase in speech intensity feedback, talkers attenuated their vocal intensity by 1 dB. In Lombard effect studies, for every 2 dB of noise presented, talkers increased their speech intensity by 1 dB. Lane and Tranel (1971) concluded that talkers would adjust their speech intensity by about 50% of the imposed stimuli. Moreover, Siegel and Pick (1974) showed that changes to speech intensity ranged between 10-40% of the imposed manipulation for sidetone amplification. They also found that if the sidetone amplification included the addition of noise, compensation to the manipulation may change as well. The effect of sidetone amplification is also stable such that when participants were tested across five days, their speech intensity responses were consistent (Chang-Yit, Pick & Siegel, 1975). The Lombard and sidetone amplification effect studies that used ongoing speech as their stimuli manipulated the environmental noise or speech intensity feedback with large changes in intensity, such as 10 dB (Siegel *et al.*, 1982; Siegel & Pick, 1974) or 20 dB (Siegel *et al.*, 1982; Siegel & Pick, 1974; Chang-Yit, Pick, Siegel, 1975).

In the recent literature, intensity perturbation studies had participants repeat isolated vowels, syllables or words (Bauer *et al.*, 2006; Heinks-Maldonado & Houde, 2005; Larson, Sun & Hain, 2007; Liu *et al.*, 2007; Therrien, Lyons & Balasubramaniam, 2012). In comparison to ongoing speech, the perturbations are smaller in magnitude (usually less than 10 dB) and duration (usually less than 500ms). A study by Heinks-Maldonado and Houde (2005) had participants say the vowel sound /α/ for 5 seconds and shifted the sound either louder or quieter by 10 dB twice within the prolonged duration. They found that participants could respond to sudden amplitude perturbations by compensating in the opposite direction. Further, studies by Bauer *et al.* (2006), Hafke (2009) and Larson, Sun and Hain (2007) found that participants had more compensation to the perturbation if the vocal feedback amplification was higher level than if it was lower level. A study by Bauer and colleagues (2006) found the magnitude of compensation increase as the perturbation increased, however, the proportion of compensation to perturbation was larger with smaller perturbations than larger perturbations. This suggested that greater compensation occurred for smaller changes in vocal intensity because smaller changes occur naturally in speech. Whereas, larger perturbations in vocal intensity is unnatural and it requires perception to determine if appropriate compensations are needed or if it should be ignored. The robustness of responses to intensity perturbations have been found in older adults between 45 to 89 years of age (Liu *et al.*, 2012), individuals with Parkinson’s Disease (Liu *et al.*, 2012) and in two syllable utterances in Mandarin talkers (Liu *et al.*, 2007).

4.1.3 Real-time perturbation paradigm

The pattern of compensation to perceived vocal amplitude changes are similar to studies that manipulate vowel formants (Purcell & Munhall, 2006ab; MacDonald, Goldberg & Munhall, 2010; Mitsuya *et al.*, 2015) using the altered auditory feedback paradigm. The paradigm generally involves four stages, in which the talker is repeating a single vowel in /hVd/ context, such as /ɛ/ in "head". The first stage is the baseline phase, where there are no manipulations to the formants. In the second stage, the ramp phase, the formants are perturbed. For example, if the first formant (F1) of the vowel /ɛ/ in "head" is decreased, it would slowly sound like /ɪ/ in "hid"; if F1 is increased, it would slowly sound like /æ/ in "had". The next stage is called the hold phase where the maximum perturbation is held constant. The maximum magnitude of F1 perturbations is usually 200 Hz as this change would shift the target vowel sound to another category (MacDonald, Goldberg & Munhall, 2010; Mitsuya *et al.*, 2015). Finally, the last stage is called the end phase. In the end phase, the perturbation is removed and the participant receives normal auditory feedback. This type of paradigm has also been used to manipulate fundamental frequency (Burnett *et al.*, 1998; Jones & Munhall, 2000) and spectral noise of fricatives (Casserly, 2011; Shiller *et al.*, 2009).

In the altered auditory feedback paradigm, compensations to formant manipulations are robust. A main pattern is that compensations occur in the opposite direction of the perturbation. For example, a positive manipulation of F1 would cause the participant to decrease their F1 (Purcell & Munhall, 2006, Mitsuya *et al.*, 2015). Another pattern is that compensation is usually around 25-50% of the formant perturbation (Houde & Jordan, 1998; MacDonald, Golberg, & Munhall, 2010; Munhall *et al.*, 2009; Purcell & Munhall, 2006; Villacorta, Perkell, & Guenther, 2007). However, this proportion of compensation may change based on the vowel and the direction of the manipulation (Mitsuya *et al.*, 2015). Partial compensation reflects that speech production is controlled by other systems than audition, such as the feedforward system and somatosensory feedback (Nasir & Ostry, 2008; Tremblay, Shiller, & Ostry, 2003). Thus, this type of paradigm may be adaptable to study the regulation of self monitoring of vocal amplitude in individuals with normal hearing and hearing loss.

The altered auditory feedback paradigm is generally used to study how the speech motor control system detects errors in speech by manipulating vowel formants or pitch. The purpose of the present study was to extend the existing literature on how the speech motor control system regulates speech by using the altered auditory feedback paradigm to manipulate vocal intensity. It is hypothesized that the altered auditory feedback paradigm would have comparable results to sidetone amplification studies. Compensation to changes in speech is robust, independent of the paradigm. For example, speech compensations occur in the opposite direction of the perturbation in sidetone amplification studies and the altered auditory feedback paradigm with vowel formants. The current study also examines how self-regulation of vocal intensity differs between individuals with normal hearing and those who wear binaural hearing aids. It is hypothesized that hearing aid users should be able to compensate to perceived vocal intensity changes, however, their compensation may be less than individuals with normal hearing. This may be due to hearing aid users using another strategy to regulate their vocal intensity and the impairment in the auditory system due to the hearing loss.

4.2 Method

4.2.1 Participants

All participants from Chapter Three participated in this study. The two experiments were counterbalanced within a session. Three groups of participants were included in the study: (1) 20 older adults with hearing aids (n = 20, M = 71.55 yrs, SD = 6.25) (2) control group with older adults (n = 19, M = 69.06 yrs, SD = 5.79) and (3) control group with younger adults (n = 23, M = 25.13 yrs, SD = 3.32). The older adults with hearing aids were required to have bilateral, symmetrical, sensorineural hearing loss and a history of binaural hearing aid use for at least one year prior to data collection. The control group with older adults were required to have hearing thresholds less than or equal to 40 dB HL between 500-4000 Hz. The control group with younger adults were required to have hearing thresholds less than or equal to 20 dB HL between 250-8000 Hz. All participants had English as their first language and no known language, neurological or speech impairments. Further details of participant information are provided in Chapter Three.

4.2.2 Hearing aid and hearing aid fitting

Details of hearing aids and fitting methods are provided in Chapter Three, described briefly here. The older adults wore bilateral Phonak Audeo V90-13 receiver-in-the-canal hearing aids during the study. Closed domes were attached to the receivers and the participants' ears were occluded with silicone earmold impression material (Hal-Hen, Per-Form H/H). The hearing aids had one program for direct audio input (DAI). The volume control and other digital signal processing features in the hearing aid were deactivated. The compression of the hearing aids was set to linear. A coupler-based verification strategy was used to fit the hearing aids to Desired Sensation Level (DSL) v5-adult targets (Scollie *et al.*, 2005) that incorporated the participants' real-ear-to-coupler-difference values in the Audioscan® RM500 SL hearing aid analyzer (Audioscan, Dorchester, ON, Canada). The hearing aid gain was adjusted using a research version of Phonak Target v4.1 programming software.

4.2.3 Equipment

Equipment used in the current study was similar to that reported in Mitsuya and Purcell (2016). Participants were seated in a sound attenuated booth (Eckel Industries of Canada, Ontario, Canada; model C26). Participants wore a headset microphone (Shure WH20) and were prompted to speak when the target word appeared on a computer screen at rate of approximately once every four seconds. The microphone signal was amplified with a microphone amplifier (Tucker-Davis Technologies MA3), low pass filtered with a cut-off frequency of 4500 Hz (Frequency Devices type 901), digitized at a 10 kHz sampling rate with 18-bit precision and amplified or attenuated in real time to produce feedback level shifts (National Instruments PXI-6289M input/output board). The processed signal was amplified to a nominal level of 60 dBA for the increase condition and 80 dBA for the decrease condition. The signal was presented through foam tip insert earphones (Etymotic Research ER2) for the control groups with older and young adults and through the hearing aids via DAI input for the older adults with hearing aids.

4.2.4 Online intensity manipulation

A single gain was applied to each trial. This gain varied between trials when there was a change in the phase of the experiment. The gain was one (i.e., no gain) during trials where no manipulation was applied, which occurred during the Acclimatization, Baseline, and End phases described below. For these phases, nominal feedback levels of 60 and 80 dBA were presented for the increase and decrease conditions, respectively. In other phases, the gain was adjusted to apply a phase-specific relative increase or decrease in utterance feedback intensity.

4.2.5 Offline intensity manipulation

Identifying vowel nuclei for offline intensity analysis was similar to the method used for offline formant analysis reported in Munhall *et al.* (2009). The harmonicity of the power spectrum was used to estimate the vowel boundaries. The boundaries were inspected by the author (L.V.) and corrected if necessary to align with the beginning and end of clear periods of harmonicity. Vowel intensity was calculated as the root-mean-square (RMS) voltage at the digitizer input for the entire vowel nucleus. It is possible to calculate absolute sound pressure levels using the microphone calibration and known microphone amplifier gain, but the relative change in voice intensity is the variable of interest for the purposes of this study. Therefore, relative changes in voice intensity were obtained by observing changes relative to the Baseline voice intensity.

Each RMS voltage value was converted to a decibel value for result analyses with the following formula:

$$dB = 20 * log10\left(\frac{utterance\ (volts)}{utterance\ average\ of\ baseline\ (volts)}\right)$$

4.2.6 Experimental phases

There were two intensity conditions: increase and decrease with the target word “head”. The order of the conditions was counterbalanced. Participants were given a passage to read with a five minute break to normalize their speech productions between conditions.

In each condition, speakers produced 115 utterances when a visual prompt was presented. These 115 trials were divided into four experimental phases. In the Acclimatization phase (utterances 1-10), participants received normal feedback. These utterances were discarded during analyses. In the Baseline phase (utterances 11-25), participants received normal feedback. In the Ramp phase (utterances 26-100), the intensity value was increased or decreased in decibel steps every 15 utterances (±2.5, ±5, ±10, ±15, ±20 dB). At utterance 101, the perturbation was removed and the participants received normal feedback until the end of the condition (End phase, End0; utterances 101-115). A schematic diagram of the experimental phases can be seen in Figure 22.

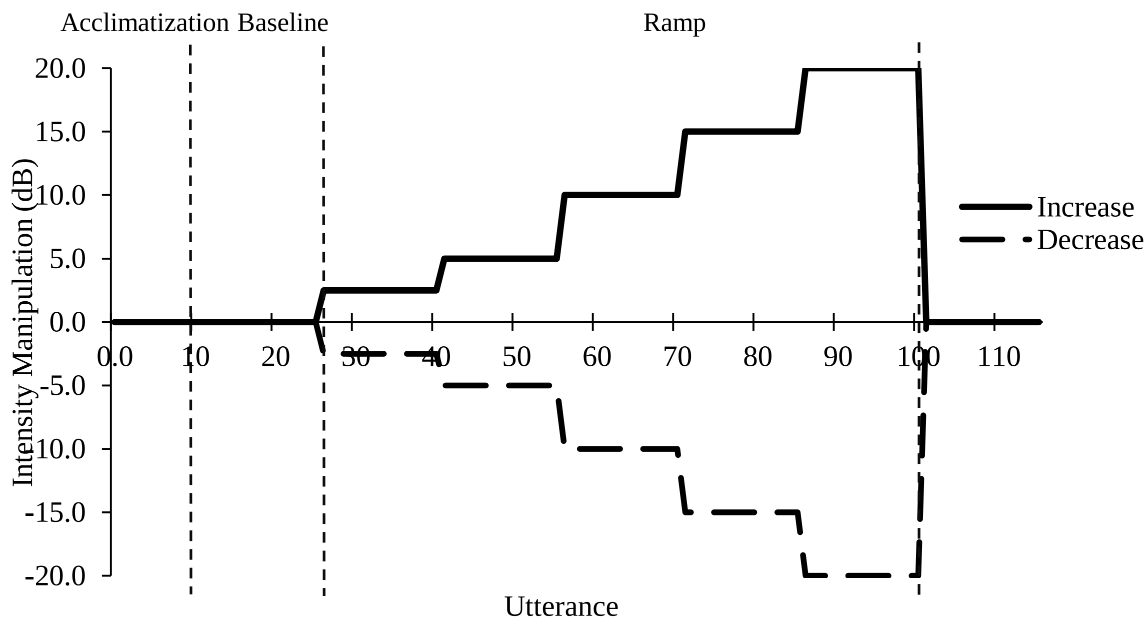


Figure 22. Schematic procedure of the intensity perturbation. Black solid line indicates the increase condition. The dashed line indicates the decrease condition. The vertical dotted lines indicate the boundaries of the four experimental phases: Acclimatization, Baseline, Ramp and End (from left to right).

4.3 Results

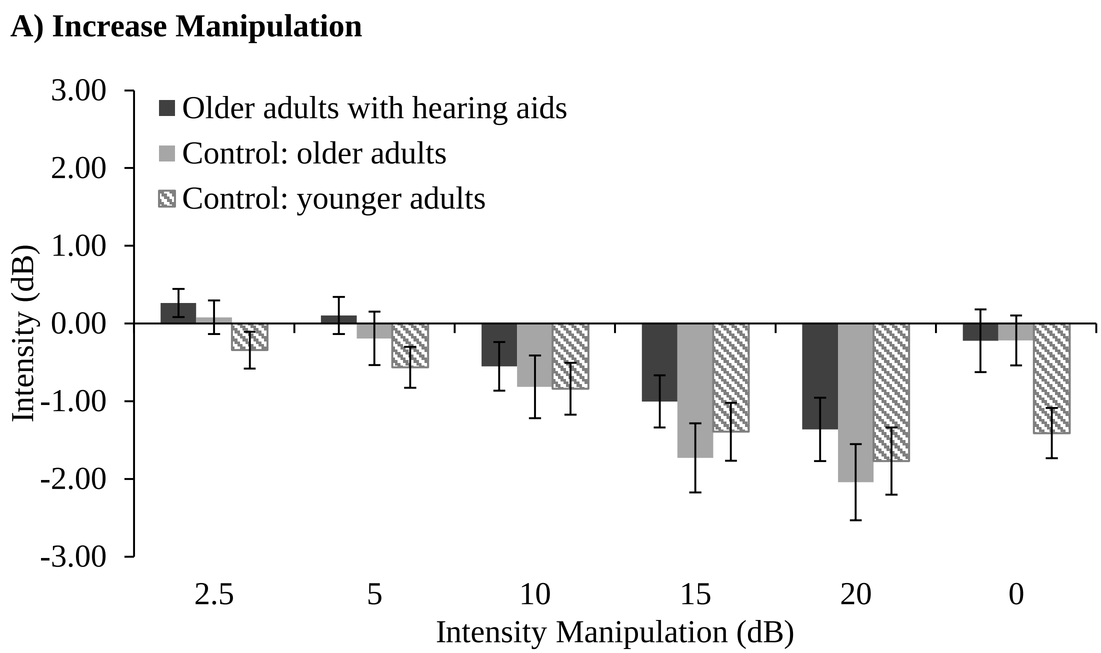
Statistical analyses in this study were completed using SPSS (version 24; IBM, Armonk, NY). The baseline average intensity was calculated using the utterances from the

Baseline phase (i.e. trials 21-40). The intensity values were then normalized by subtracting a speaker's baseline average from each utterance. To quantify a change in intensity production, the average normalized intensity values during each phase were calculated. For all statistical analyses, Greenhouse-Geisser corrected degrees of freedom were used for interpretation of significant effects. Results of the ANOVAs were interpreted at an α of 0.05.

Figure 23 illustrates the average normalized change in speech intensity across manipulation phases and direction of manipulation for each group.

Figure 24 illustrates the average normalized speech intensity change across utterances in each direction of manipulation for each group.

Table 4 shows the average change in normalized speech intensity across manipulation phases in decibels and percent compensation for each group.



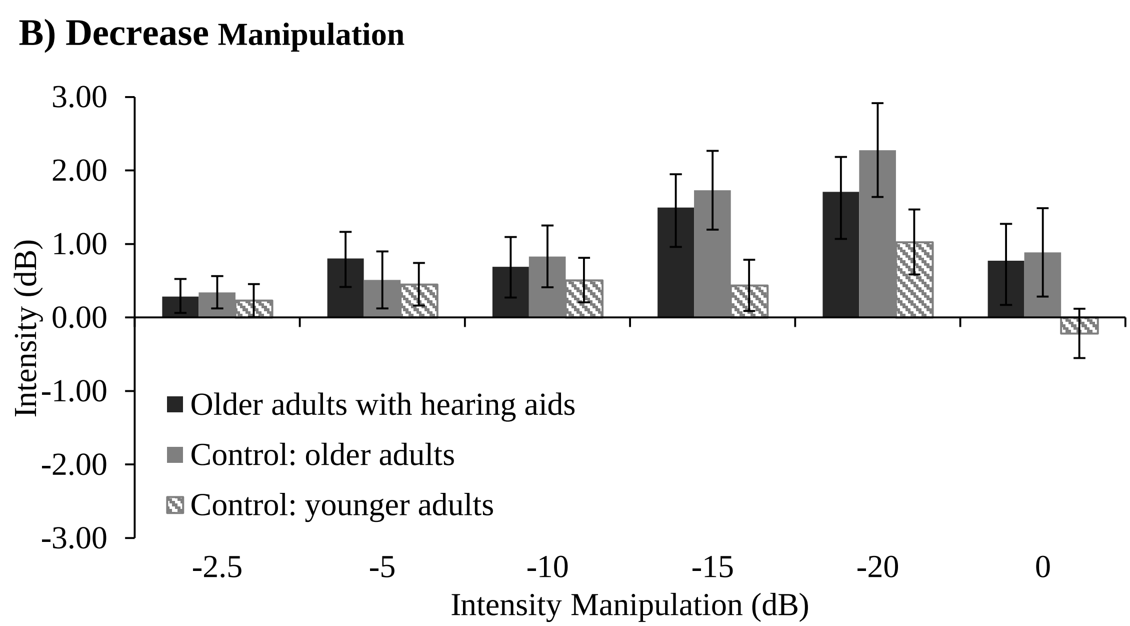


Figure 23. Average change in speech intensity across manipulation phases: A) Increase manipulation; B) Decrease manipulation. Black bars indicate older adults with hearing aids. Grey bars indicate control group with older adults. Striped bars indicate control group with younger adults. The error bars indicate ±1 standard error.

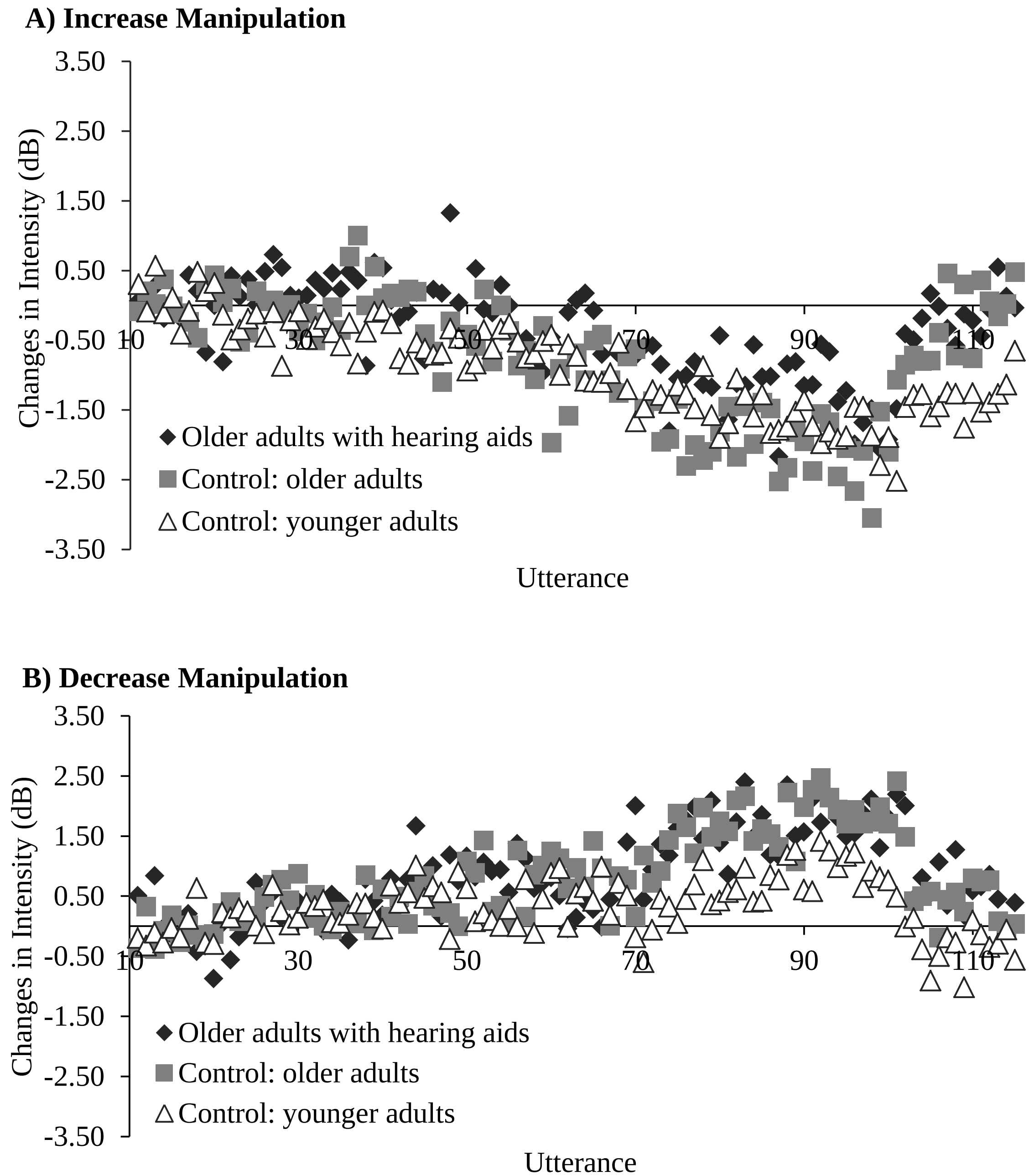


Figure 24. Average change in speech intensity values across participants for each utterance: A) Increase manipulation; B) Decrease manipulation. Black diamonds indicate older adults with hearing aids. Grey squares indicate control group with older adults. White triangles indicate control group with younger adults.

Table 4. Average changes in speech intensity across manipulation phases

Manipulation Phases	2.5		5		10		15		20	
Compensation	**dB**	**%**	**dB**	**%**	**dB**	**%**	**dB**	**%**	**dB**	**%**
Increase Manipulation										
Older adults with hearing aids	0.30	12.14	0.07	1.47	-0.53	5.29	-1.01	6.73	-1.34	6.71
Control: Older adults	0.06	2.42	-0.28	5.67	-0.91	9.11	-1.80	12.02	-2.09	10.47
Control: Younger adults	-0.35	13.82	-0.58	11.57	-0.85	8.55	-1.38	9.21	-1.77	8.87
Decrease Manipulation										
Older adult with hearing aids	0.31	12.51	0.74	14.72	0.71	7.14	1.49	9.91	1.71	8.62
Control: Older adults	0.35	15.09	0.35	7.04	0.82	8.17	1.56	10.38	1.83	9.17
Control: Younger adults	0.22	8.94	0.36	7.22	0.46	4.57	0.42	2.81	0.97	4.84

Notes: dB = average speech intensity changes in decibels; % = average intensity changes in percent compensation

To compare the two directions of intensity manipulation, the magnitude of compensation for each speaker in the increase condition was multiplied by -1. A mixed ANOVA was performed with two within-subject factors (direction: increase, decrease; phases: Ramp2.5, Ramp5, Ramp10, Ramp15, Ramp20, End0) and one between-subject factor (group: older adults with hearing aids, control group with older adults, control group with younger adults). The main effects of direction and group were non-significant [direction: $F(1.00, 0.07) = 0.004$, $p = 0.95$, $\eta_p^2 = 0.00$; group: $F(2, 59) = 0.21$, $p = 0.81$, $\eta_p^2 = 0.01$]. The main effect of phases was significant [$F(3.58, 54.42) = 41.32$, $p < 0.001$, $\eta_p^2 = 0.41$].

All interactions found were non-significant ($p > 0.05$). A post hoc analysis on the effect of phases, adjusted for multiple comparisons using Bonferroni corrections, is shown in Figure 25. Overall, as the magnitude of manipulation increased, the change in intensity increased. When the manipulation was removed (End0), there were smaller changes in intensity compared to Ramp15 and Ramp20.

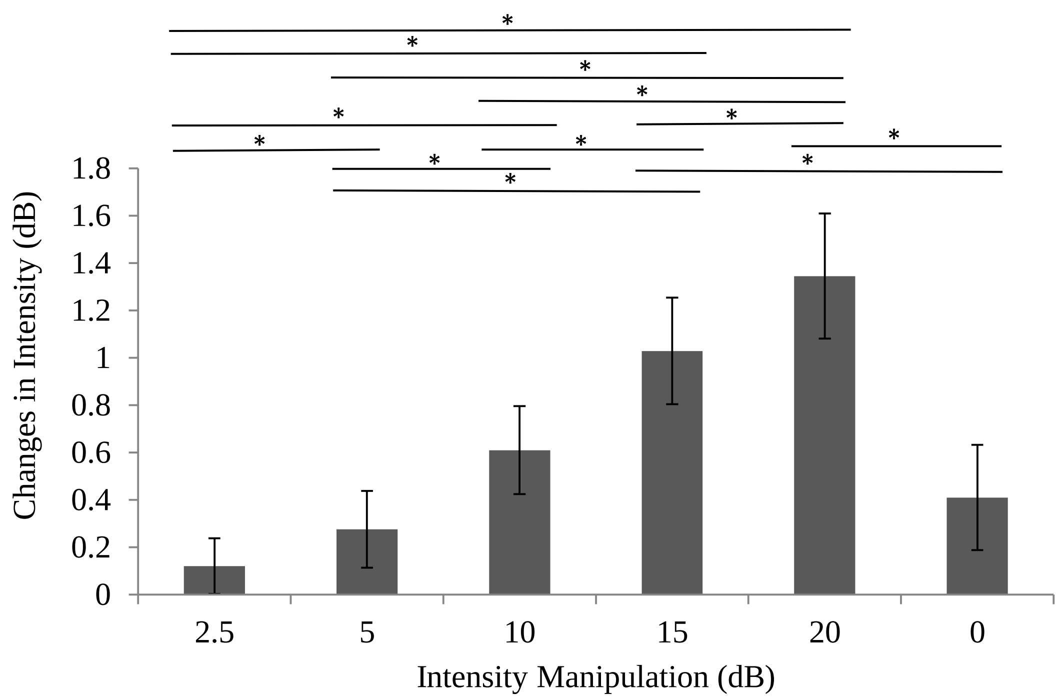


Figure 25. Average change in speech intensity across phases. The error bars indicate ±1 standard error. * indicates significant difference ($p \leq 0.05$).

4.4 Discussion

Numerous studies have shown that intensity perturbations elicit changes in speech intensity output that correct for the induced errors (Chang-Yit *et al.*, 1975; Larson *et al.*, 2007; Bauer *et al.*, 2006; Liu *et al.*, 2012; Hafke, 2009; Heinks-Maldonado & Houde, 2005). These studies have shown that auditory feedback is important to speech motor control and controlling speech intensity output is important for speech production. The current study used a new paradigm to study the sidetone amplification effect where intensity perturbations were elicited in single syllable utterances in younger and older adults with normal hearing and older adults with hearing loss.

The perturbation paradigm that has usually been studied with vowel formant perturbations was adapted to study intensity perturbations. All groups of talkers in the current study responded to the intensity perturbations by compensating in the opposite direction of the perturbations, increased the magnitude of compensation with larger perturbations, and the magnitudes of compensations were smaller than the magnitudes of the perturbations. These patterns of results are similar to the perturbation paradigms that manipulated vowel formants (Purcell & Munhall, 2006b; Mitsuya *et al.*, 2015, MacDonald *et al.*, 2010, 2011; Villacorta *et al.*, 2007). This perturbation paradigm is an effective experimental methodology to study the influence of auditory feedback on the regulation of speech production.

The only significant effect that the current findings had was on phases, such that as the magnitude of manipulation increased from 2.5 to 20 dB, the magnitude of vocal intensity increased. This showed that the speech motor control system is able to detect various intensity changes and partially compensate. Similarly, Chang-Yit and colleagues (1975) had a gradual manipulation where they increased or decreased the sidetone in 2 dB steps until the maximum perturbation of 20 dB. They also found the magnitude of compensation increased with the magnitude of the perturbation.

Sensorimotor learning may have taken place during the current study. In the End phase, when the intensity feedback was returned to baseline, the voice levels of the participants were similar to the voice levels when the manipulations were between 2.5 to 10 dB. The intensity levels did not return to baseline level at the End phase and compensation persisted when feedback was returned to normal. This is similar to other perturbation studies where they have manipulated vowel formants (Purcell & Munhall, 2006b; MacDonald *et al.*, 2011; Villacorta *et al.*, 2007). For example, MacDonald *et al.* (2011) showed that F2 did not return to baseline levels when the formant manipulation was returned to normal. The present results show the speech motor control system adapted to the perturbation paradigm and continued to anticipate changes in intensity.

The current results had no compensation differences between the directions of the manipulation, such that the increase and decrease manipulations were not significantly different from each other. This was similar to the studies by Heinks-Maldonado and Houde (2005) and Chang-Yit *et al.* (1975), in which they also did not find directional differences in compensation behaviours. However, studies by Larson *et al.* (2007), Bauer *et al.* (2006), Liu *et al.* (2012) and Hafke (2009) had directional differences, such that the compensations in the increase direction were larger in magnitude than compensations in the decrease direction. The differences in the literature could be due to differences in the magnitude of perturbations. The studies that had a significant difference between the increase and decrease manipulations used perturbations that were less than 6 dB. In contrast, studies that had no significant differences between the two directions of manipulations had magnitude of perturbations that were greater than 10 dB. The current study was the only study that manipulated intensity across a wide range of magnitudes from 2.5 to 20 dB. This wide range of magnitudes may not have been able to differentiate between the directions of manipulations. Future studies may want to separate the perturbation levels into smaller groups (i.e. less than 6 dB versus 10 dB or more) to have comparable results with the literature.

The current results showed partial compensation where the magnitude of compensation was not equivalent to the magnitude of the perturbation. This demonstrates that the talker's speech intensity is not completely controlled by auditory feedback. Furthermore, the current findings had higher percent compensations with smaller perturbations and lower percent compensations with larger perturbations. Studies by Bauer *et al.* (2006), Hafke (2009) and Larson *et al.* (2007) also found proportionally larger compensation for smaller perturbations than those for larger perturbations. This suggests the speech motor control system may be able to differentiate between natural and unnatural occurring intensity changes in speech. The speech motor control system anticipates small fluctuations in intensity in natural speech production and can make appropriate corrections. If the intensity perturbation is larger than naturally occurring speech intensity changes, the speech motor control system may rely on other feedback systems to determine if the error is real or not. By having other regulatory feedback systems, it would prevent the talker's speech from excessively changing based on environmental background sounds. For example, when talkers were provided with visual feedback to monitor how loud their voice was, the amount of speech intensity increase when auditory feedback masked was smaller than when talkers were not provided with visual feedback (Therrien *et al.*, 2012). As well, Erickson (2002) showed that changes in jaw and tongue movements can result in increased speech intensity without the need to change auditory feedback. If the talker opened their jaw more, it would emphasize their vowel sound and increase the intensity. Thus, responses to small intensity perturbations may be optimally monitored by the auditory feedback system, however, with larger intensity perturbations, other feedback systems may be more involved.

Overall, the proportions of compensations at each manipulation level in the current study are smaller than reported literature values. The only exception was that the current study and the study by Heinks-Maldonado & Houde (2005) at a 10 dB manipulation had similar percent compensations of around 5-9%. One possible explanation for the current study having smaller compensation values could be the starting level. The starting level of the increase condition was 60 dBA, increasing to 80 dBA. Most studies such as Chang-Yit *et al.* (1975), Larson *et al.* (2007), Hafke (2009), Siegel *et al.* (1982) had starting levels between 75-80 dB SPL, such that the maximum perturbations would be over 80 dB SPL. Thus, the current study may need to start at a higher level for the increase condition to compare with other literature values. However, the starting level for the decrease manipulation in the current study was similar to other studies and the current study still produced lower compensation levels.

Another possible reason for smaller compensation is the differences in experimental manipulations of speech between the current and past studies. In the current study, the entire word, "head" changed in intensity. Whereas, a common methodology in the literature is to manipulate intensity within a prolonged utterance of a vowel (i.e. /u/ or /α/) (Bauer *et al.*, 2006; Hafke, 2009; Larson *et al.*, 2007; Liu *et al.*, 2012). Another methodology is to change the intensity as the talker is speaking (i.e. reading the Rainbow passage aloud; Chang-Yit *et al.*, 1975; Siegel *et al.*, 1982). These differences in starting levels and types of speech chosen may have resulted in differences in compensation levels. Future studies that use the perturbation paradigm of the current study may want to use similar starting levels and vowels/words (i.e. "who'd" or "hawed") as past studies for comparable results.

There were no significant group differences in the current study: all groups of participants had similar compensation patterns to the intensity perturbations. The current paradigm and analyses were not able to differentiate between older and younger adults. This suggested that there were minimal or no aging effects on responding to intensity changes in speech production. Further, the current paradigm was not able to differentiate between individuals with normal hearing and individuals with bilateral hearing aids. This showed that the hearing aids translated the person's speech intensity level appropriately and the user relied on their auditory feedback system to detect the intensity perturbations. Whereas, the study by Laugesen *et al.* (2009) suggested that some hearing aid users did not rely on their auditory feedback for speech intensity level control. However, the study by Laugesen *et al.* (2009) measured voice level changes by asking the talker to speak to others at different distances. This difference in task may have tapped into another mechanism of speech intensity control. To further understand how a hearing loss and hearing aids may affect the speech motor control system for voice level control, future studies may want to have individuals with hearing loss perform the perturbation task without their hearing aids on.

The current experimental design may have limited the ability to detect group differences. The current study has many condition levels for each variable, such as six perturbation changes for each direction. In the literature, there are usually one (Chang-Yit *et al.*, 1975; Heinks-Maldonado & Houde, 2005; Larson *et al.*, 2007; Liu *et al.*, 2007), two (Liu *et al.*, 2012; Seigel & Pick, 1974; Seigel *et al.*, 1982), three (Bauer *et al.*, 2006) or four (Hafke, 2009) perturbation changes for each direction. As well, these perturbation changes are usually small perturbation changes (less than 6 dB) or large perturbation changes (more than 10 dB). For example, the study by Bauer *et al.* (2006) used three small perturbation changes (1, 3 and 6 dB) for each direction of manipulation. Whereas, Seigel and Pick (1974) and Siegel *et al.* (1982) used two large perturbation changes (10 and 20 dB) in their experiments. If the current experiment were to separate the perturbation changes into small and large perturbation changes, significant group differences may have occurred. In Figure 23b for the negative manipulation, there were no observable group differences for small perturbations (less than 5 dB). However, observable group differences for large perturbations (greater than 10 dB) were seen, such that the younger adults had less speech intensity changes than the older adults with normal hearing and hearing aids. However, these group differences were not significant within the ANOVA analysis. Future studies may want to replicate the current paradigm, however, separate the perturbation steps into smaller groups.

There are other limitations within the current study that could have reduced the probability of detecting group differences. One limitation is the variability of demographics within the participants. The group with hearing aids varied in degree and configuration of hearing loss. Future studies may want to separate the hearing aid users by degree of hearing loss: mild, moderate, and severe. As well, the older adults ranged from 55 to 80 years of age. Future studies may want to reduce the age range and add more age categories to determine if there are aging effects. Another limitation is the sample size. The sample size was approximately 20 participants per group in the current study. Increasing the sample size would increase power for detecting group differences.

The current paradigm was able to measure the sidetone amplification effect: as the talker's speech intensity was amplified, the talker decreased their speech intensity and vice versa. Differences that occurred between the current study and other intensity-shifted feedback studies may have been due to differences in experimental methodologies. This suggests that compensation to intensity perturbations may be task sensitive and the regulation of speech intensity is modulated by many parameters. Further, the speech motor control system, when controlling for intensity perturbations, may not be affected by hearing loss (mediated by the amplification of hearing aids) and aging effects. Future studies with the perturbation paradigm will help further understand how much auditory feedback modulates the control of speech production.

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Chapter 5

5 Speech production changes across different parameters of non-adaptive and adaptive non-linear frequency compression

5.1 Introduction

Satisfaction with hearing instruments is, in part, dependent on sound clarity, natural sounding, and fidelity or richness of sound (Kochkin, 2005). Specifically, 77% of hearing instrument users are satisfied with the sound of their hearing aids and 73% of hearing instrument consumers are satisfied with the sound of their voice (Kochkin, 2010). The reasons for dissatisfaction, particularly with the sound of a user's own voice, are not well understood and the literature review presented here reveals few studies on this issue. It is important to determine the possible reasons for poor satisfaction with one's own voice during hearing aid use, in order to understand the nature of the problem and to potentially improve satisfaction, acceptance and use.

A common complaint in hearing aid users have that they perceive they are talking in a barrel, their voice sounds hollow, or that when they are chewing their food it sounds loud (Chung, 2004; Kuk & Ludvigsen, 2002). These unnatural perceptions of their voice are caused by the occlusion effect. The occlusion effect occurs when bone-conducted sounds are trapped in the ear canal because the opening of the ear canal is blocked by the hearing aid (Chung, 2004; Stender & Appleby, 2009; Winkler, Latzel, & Holube, 2016). It affects the perception of intensity and timbre of the hearing aid user's own voice (Sweetow & Pirzanski, 2003). Another way a hearing aid can affect the perception of the hearing aid user's voice is ampclusion (Kuk & Ludvigsen, 2002; Painton, 1993; Sweetow & Pirzanski, 2003). Ampclusion occurs because the person's voice when talking is closest to the hearing aid microphone. As a result, the sound of the hearing aid user's voice is perceived to be about 15 dB louder than other voices (Sweetow & Pirzanski, 2003). In addition, if the hearing aid has a delay of more than 20 milliseconds, the quality of the hearing aid user's voice can also be affected (Stone & Moore, 1999, 2002). It is important for hearing aids to not distort the user's voice so that it does not affect the perception of other sounds or perceived errors in their speech.

The ability to hear one's own voice uses the auditory feedback system. This system helps the individual to monitor and maintain accurate speech production. In talkers who have normal hearing, studies have indicated that talkers will make changes in their speech to correct for experimental perturbation to the auditory feedback. For instance, if the feedback signal is amplified, the speaker decreases his/her vocal intensity, whereas if the sound is attenuated, the talker increases his/her vocal intensity (Borden, Harris, & Raphael, 1994). As well, if the sound is filtered, the talker would make speech adjustments to modify some of his/her vocal tract resonance characteristics so that the perceived speech sound is closer to a perceptual target sound (Garber, Seigel & Pick, 1981). Speech compensation effects have been studied in frequency-shifted studies that perturbed the fundamental frequency (F0) (Burnett *et al.*, 1998; Jones & Munhall, 2000), vowel formants (Purcell & Munhall, 2006; MacDonald *et al.*, 2010, Mitsuya *et al.*, 2015) and spectral noise of fricatives (Casserly, 2011; Shiller *et al.*, 2009). There have also been studies that examined speech compensations using the altered auditory feedback paradigm in older adults (Liu *et al.*, 2011; Liu, Russo, & Larson, 2010) and individuals with Parkinson's Disease (Chen *et al.*, 2013; Liu *et al.*, 2012; Mollaei, Shiller, & Gracco, 2013). These altered auditory feedback studies have shown that changes in an individual's perception of speech sounds can alter changes in speech production. One advantage of the perturbation method is that listeners are often unaware of the perturbation or their response to it, allowing study of the auditory feedback mechanism using changes that are subtle and not associated with complaints of hearing aid wearers. However, perturbation studies in hearing aid users are generally lacking.

5.1.1 Frequency lowering technology: non-linear frequency compression

It is plausible that one possible reason for poor sound quality of one's own speech during hearing aid use is the perturbation that may be caused by the hearing aid signal processing. Intended to assist the listener in hearing externally-produced sounds, hearing aid signal processing manipulates the incoming signal in level, shape, envelope, and many other parameters. The impact of these on speech production is largely unknown. One interesting example of a signal processor that may perturb the feedback path is frequency lowering. Frequency lowering programs are used when high frequency audibility is poor in hearing aids. Poor amplification in the high frequency regions is due to the limited bandwidth of hearing aids, which provide amplification only to 6 kHz (Stelmachowicz *et al.*, 2008). Studies by Boothroyd and Medwetsky (1992) and Stelmachowicz *et al.* (2001) found that the frequency range of fricatives was from 2 to 4 kHz in male talkers and from 2 to 8.9 kHz in females and children. Thus, individuals with hearing loss may not adequately perceive high frequency sounds such as /s/ and /z/, especially when listening to female or child speakers or, for some talkers, their own speech production. Such poor audibility in high frequency input may delay production of fricatives, understanding and use of morphological rules (Lane & Webster, 1991; Moeller *et al.*, 2007; Stelmachowicz *et al.*, 2002, 2004). As well, speech understanding in noise and sound localization can be affected (Bohnert, Nyffeler, & Keilmann, 2010; Dubno, Ahlstrom, & Horwitz, 2002).

The goal of frequency lowering technology is to provide audibility to high frequency information regions by moving high frequency sounds to a lower frequency range where audibility is more likely (Kuk *et al.*, 2009; Wolfe *et al.*, 2010). One of the types of frequency lowering technology in current commercial hearing aids is non-linear frequency compression (NLFC). With NLFC, input frequencies are compressed above a cut-off frequency by a specified ratio so that high frequency inputs are shifted to a lower frequency range. Inputs below the cut-off frequency are not compressed and do not overlap with the lower frequency region, so natural formant ratios of vowels and F0 are maintained (Wolfe *et al.*, 2010). Currently, there are four manufacturers that incorporate NLFC into their hearing aids: Phonak and Unitron in their SoundRecover program, GN Resound in their Sound Shaper program, and Siemens in their Frequency Compression program. Reviews of older NLFC technology are found in Auriemmo *et al.* (2009), Simpson (2009) and McCreery *et al.* (2012).

Phonak has two types of NLFC in their SoundRecover program. In the SoundRecover1 (SR1) program, a specified cut-off frequency (CT) is determined. With this cut-off point, frequencies above are compressed into a reduced range and frequencies below are preserved. In the program, the clinician can change the cut-off frequency and the compression ratio, such that weaker settings are created with a higher cut-off frequency (1500 to 6000 Hz) and a lower compression ratio (1.5:1 to 4:1). This SR1 program is non-adaptive, so the cut-off frequency and compression ratio remain constant for all sound inputs. The newest iteration of SoundRecover is adaptive and is referred to as SoundRecover2 (SR2). SR2 is similar to SR1, such that low frequency sounds (i.e. vowels) receive little or no compression and high-frequency sounds are compressed (i.e. fricatives). The difference between SR1 and SR2 is that the cut-off frequencies in SR2 changes based on the energy distribution of the incoming signal (Glista *et al.*, 2016c). There are two cut-off frequencies, CT1 and CT2, in which CT1 has a lower cut-off frequency and CT2 has a higher cut-off frequency. SR2 rapidly applies CT1 or CT2 based on the incoming signal; if the spectrum of the signal is high frequency dominant, CT1 is selected and if the incoming signal is low frequency dominant, CT2 is selected. With a higher cut-off frequency (CT2) for low frequency signals, it lessens the effects of NLFC where frequency lowering may not be needed to improve audibility. The clinician can adjust CT1, CT2 and the strength of the compression ratio in Phonak's Target fitting software, see Glista *et al.* (2016c) for fitting protocols and verifications for SR2. SR2 was recently included in commercial Phonak hearing aids and minimal research has been conducted to compare SR1 and SR2. At the time of writing, the current literature on NLFC provides information on the effects of SR1 on speech perception and production, but studies of SR2 are lacking.

5.1.2 Non-adaptive non-linear frequency compression

Some studies have found that children and adult HA users benefit from NLFC technology. Glista, Scollie and Sulkers (2012) evaluated the effect of frequency compression in children on speech perception abilities through speech detection of /s/ and /ʃ/, /s-ʃ/ discrimination and plurals and consonant recognition. Findings suggest that frequency compression provided varying outcomes in benefit and acclimatization across listeners. Some participants demonstrated large changes in speech perception ability with NLFC hearing aids and others demonstrated little change. Similarly, studies by Wolfe and colleagues (2010, 2011) evaluated the use of NLFC in children with moderate to moderate-severe HL. The results showed that NLFC improves audibility for and recognition of high-frequency speech sounds. Wolfe *et al.* (2011) also found continued improvement in recognition of speech sounds in quiet after six months of NLFC use and improvement in speech recognition in noise after several weeks to several months of use. The studies by Glista *et al.* (2009) and Simpson, Hersbach and McDermott (2005) also had adults using NLFC technology show improvements in their speech recognition abilities. Glista *et al.*, (2009) demonstrated that benefit with frequency lowering may be correlated with degree and slope of hearing loss, with users who have more hearing loss in the high frequencies tending to receive more benefit.

In contrast, there are studies that found NLFC was neither detrimental nor advantageous in the hearing aid user's speech perception abilities. Hillock-Dunn *et al.* (2014) had children with NLFC enabled and disabled for various speech identification tasks in quiet and noise. Results found that there were no significant performance differences, on average, between the enabled and disabled conditions for all speech perception measures. However, subjects with greater difference in audible bandwidth between NLFC on and off were more likely to demonstrate improvements in high-frequency consonant identification in quiet and improvements in spondee identification in noise. Other studies by Perreau, Bentler, and Tyler (2013) and McDermott and Henshall (2010) have shown no significant differences with NLFC enabled on speech perception measures compared to NLFC disabled. Simpson, Hersbach and McDermott (2006) found similar results in which they found subjective benefits with NLFC in quiet and in noise, however no significant differences were measured by speech recognition testing with NLFC enabled or disabled. They speculated that incomplete acclimatization to the NLFC may have caused more confusion among fricative phonemes. Other authors have presented case studies in which suboptimal settings may disrupt benefit for individuals (Scollie, Glista, & Richert, 2014), leading to the development of validated fitting protocols (Scollie *et al.*, 2016). The need for both valid fitting of settings and an acclimatization period is supported by the findings of Glista *et al.*, (2012), who studied the time course of acclimatization with well-fitted settings, and found that for some listeners a six week use period was necessary before consonant recognition was optimized.

The frequency lowering scheme of NLFC should maintain or not affect the listener's normal vowel perception. Glista *et al.* (2009) and Simpson *et al.* (2005) found no significant differences in vowel identification and vowel phoneme scores using frequency compression enabled compared to frequency compression disabled. Perreau *et al.* (2013) found that vowel perception abilities in quiet using the frequency compression of Phonak Naida hearing aids were better than conventional amplification. However, Perreau *et al.* (2013) suggests that if the cut-off frequency in NLFC is set too low (e.g. 1500 Hz) and a high compression ratio is used, higher formants such as F2 and F3 might be too severely compressed or reduced and spectral smearing of the input signal might occur. The authors suggest using a higher cut-off frequency (e.g. 6000 Hz) and a lower compression ratio to prevent negative impacts on vowel perception.

Different NLFC parameters have different effects on speech recognition. A study by Ellis and Munro (2013) examined a low cut-off frequency (1600 Hz) with a compression ratio of 2 or 3 on sentence recognition in noise with young adults with normal hearing. Their results found that sentence recognition decreased with higher compression ratios. Similarly, Souza *et al.* (2013) used different cut-off frequencies (1000, 1500 and 2000 Hz) and compression ratios (1.5, 2.0 and 3.0) on sentence intelligibility in noise with adults with normal hearing or sensorineural hearing losses. There was a decrease in sentence recognition as cut-off frequency decreased and as compression ratio increased. Sentence recognition did not decrease when the cut-off frequency was 2000 Hz, regardless of compression ratio and when the compression ratio was 1.5, regardless of cut-off frequency. Overall, higher cut-off frequencies and lower compression ratios may result in higher sentence recognition.

Alexander (2016) also examined the impact of frequency compression parameters, studying the effect of six combinations of cut-off frequencies and input bandwidth (by varying compression ratios) on vowel and consonant recognition in noise with individuals who had moderately-severe and mild to moderately-severe hearing losses. The results found that a low cut-off frequency, 1600 Hz, had reduced vowel and consonant recognition, especially as compression ratio increased. In comparison, at higher cut-off frequencies (2800 Hz and 4000 Hz), phoneme recognition was unaffected.

Further, vowel recognition decreased when there was a larger change to the second formants at lower cut-off frequencies (1600 and 2200 Hz). An ideal setting for maximizing all vowel and consonant recognition cannot be achieved as phonemes vary in their acoustic properties. One setting may maximize the recognition of one phoneme (i.e. fricatives), however, it may decrease the recognition of other phonemes (i.e. vowels). Generally, if the cut-off frequency is less than 2200 Hz with high compression ratios, it may degrade speech recognition.

Sound quality ratings and subject preferences also vary with different NLFC parameters. A study by Parsa *et al.* (2013) examined the effects of different NLFC parameters on sound quality ratings of music and speech with adults and children with normal hearing and sensorineural hearing losses. The cut-off frequency varied from 1600 to 3150 Hz and compression ratio varied from 2.0 to 10.0. The results showed that sound quality ratings were more affected by cut-off frequencies than compression ratios, with ratings decreasing as cut-off frequency decreased below 3000 Hz. A study by Johnson and Light (2015) studied sound quality preference in three NLFC settings that was stronger than the manufacturer's default setting in older adults with severe high-frequency sensorineural hearing loss. The results found that the participants equally preferred the sound quality of the manufacturer's default setting and when NLFC was turned off. When NLFC was stronger than the manufacturer's default setting, participants preferred settings that had less NLFC, even if the stronger settings had an objective improvement in audibility. Overall, NLFC parameters with stronger frequency compression settings, via lower cut-off frequencies and higher compression ratios, may experience poorer sound quality ratings. If settings are too strong, it is possible that poor speech recognition abilities could result. Systematic fitting protocols attempt to minimize these unwanted side effects while maximizing improvement to high frequency audibility (Scollie *et al.*, 2016).

5.1.3 Non-adaptive vs. adaptive non-linear frequency compression

There are few studies that have compared non-adaptive and adaptive NLFC. Wolfe *et al.* (2017) evaluated audibility and speech recognition abilities in children with severe-to-profound high frequency hearing loss using both types of NLFC. Their results found after 4-6 weeks of acclimatization to adaptive NLFC, the children had better plural detection and word recognition scores than with non-adaptive NLFC. However, there were no differences between the two types of NLFC with phoneme detection thresholds and recognition scores. Glista *et al.* (2016b) presented two case studies that acclimatized to adaptive and non-adaptive NLFC. The results found that a benefit to using NLFC compared to when NLFC was turned off. As well, when there was a difference between adaptive and non-adaptive NLFC, the adaptive NLFC had better scores. Based on the two studies that compared adaptive and non-adaptive NLFC, when there was a difference between the two types of NLFC, adaptive NLFC showed greater benefit.

Different parameters of adaptive NLFC may result in different speech perception scores and sound quality ratings. Wolfe and colleagues (2017) compared two settings of non-adaptive NLFC. One setting (NLFC-2A) used the default settings of the adaptive NLFC and the second setting (NLFC-2B) was an adaptive NLFC setting that was closely matched to the parameters of non-adaptive NLFC. The two settings of the non-adaptive NLFC were not significantly different for plurals detection, however, NLFC-2B had better word recognition than NLFC-2A. Further, Glista *et al.* (2016a) had individuals with normal hearing and hearing loss rate sentences that were filtered with different adaptive NLFC settings from “very bad” to “very good”. Their results found that perceived sound quality varied with the strength of the adaptive NLFC such that with increasing strength of adaptive NLFC, sound quality ratings decreased. However, most of the sound quality ratings ranged between “average” and “good”. Thus, there are differences between the different parameters of adaptive NLFC, as a result, fine tuning and acclimatization may be needed when using adaptive NLFC.

5.1.4 Statement of purpose

In summary, the auditory feedback system is sensitive to changes in sound quality. These sound quality changes can be induced by different SR1 or SR2 settings. This may then result in changes in speech production as the talker’s auditory feedback system responds to the processed or distorted sounds. However, this proposed effect of frequency lowering on speech production has not yet been evaluated in an experimental context. The perturbation paradigm may be a valid method for exploring this issue. Therefore, the purpose of the current study was to measure changes in speech by manipulating the auditory feedback loop using SR1 and SR2. Several cut-off frequencies and compression ratios settings were used to vary the effects of SR1 and SR2. Vowels /ɛ, ɪ, i/ and consonant /s/ were selected for use in a perturbation task, because they have different energy distributions. The phoneme /s/ has the highest-frequency noise spectra distribution in this set, and the vowels range in second formant frequencies and span the upper-frequency portion of the vowel space in English. It was hypothesized that frequency lowering signal processing may perturb the auditory feedback loop, with more effects at stronger settings. Further, it was hypothesized that the two cut-off frequencies in SR2 may protect stimuli with low frequency information across different strengths of NLFC. Specifically, changes in formant productions for vowels may be less in SR2 than in SR1. Further, at stronger settings of SoundRecover, where cut-off frequencies are below 2200 Hz (Alexander, 2016), the sounds may be highly distorted or unnatural. The speaker's auditory feedback system may choose to ignore the distorted, processed sounds at stronger settings of SoundRecover. There might be minimal changes to speech at the stronger settings.

5.2 Method

5.2.1 Participants

All participants from Chapter Three participated in the current study. Participants came on a different session day from the studies in Chapter Three and Four. Three groups of participants were included in the study: (1) 20 older adults with hearing aids (*n* = 20, *M* = 71.55 yrs, *SD* = 6.25) (2) control group with older adults (*n* = 19, *M* = 69.06 yrs, *SD* = 5.79) and (3) control group with younger adults (*n* = 23, *M* = 25.13 yrs, *SD* = 3.32). The older adults with hearing aids were required to have bilateral, symmetrical, sensorineural hearing loss and a history of binaural hearing aid use for at least one year prior to data collection. The control group with older adults were required to have hearing thresholds less than or equal to 40 dB HL between 500-4000 Hz. The control group with younger adults were required to have hearing thresholds less than or equal to 20 dB HL between 250-8000 Hz. All participants had English as their first language and no known language, neurological or speech impairments. Further details of participant information are provided in Chapter Three.

5.2.2 Hearing aid and hearing aid fitting

The older adults with hearing aids used the same hearing aids as Chapter 3. A separate hearing aid program was created for the current study. The microphone in this program was set to omnidirectional mode. The volume control and other digital signal processing features in the hearing aid such as noise reduction were deactivated. Test-box hearing aid verification was completed using procedures similar to those described in Chapter 3.

The control groups with younger and older adults wore Phonak Audeo V90-13 receiver-in-canal hearing aids with standard receivers. The hearing aid program had an omnidirectional microphone and digital signal processing features were deactivated. The compression of the hearing aids was set to linear. The hearing aids were programmed to DSLv5-adult targets for a flat 40 dB HL audiogram from 250-8000 Hz with average real ear to coupler difference values on Audioscan® Verifit2 (Audioscan, Dorchester, ON, Canada).

When the participants arrived, closed domes were attached to the receivers and the participants' ears were occluded with silicone earmold impression material (Hal-Hen, Per-Form H/H). During the experiment, the hearing aids were attached to a hearing aid programmer (HI-PRO 2, Otometrics, Denmark) with programming cables (CS-44A, Phonak) binaurally. The hearing aid programmer was connected to a laptop with a research version of Phonak Target (v4.1) programming software. This allowed the researcher to change SoundRecover settings during the experiment.

5.2.3 SoundRecover1 and SoundRecover2 settings

A Phonak Audeo V90-13 receiver-in-canal hearing aid was programmed to a flat 40 dB HL audiogram from 250-8000 Hz with average real ear to coupler difference values on the Audioscan® Verifit2. The output of the hearing aids was verified to meet Desired Sensation Level v5-adult targets (Scollie *et al.*, 2005) for adults at input levels of 55, 65 and 75 dB SPL for running speech passages. SoundRecover settings were chosen by examining the aided peak of a calibrated /s/ stimulus that has been developed for use in verifying hearing aids (Scollie *et al.*, 2016). SoundRecover settings were chosen that spectrally separated the peak of /s/, across settings, by approximately 1000 Hz. Setting 1 of SR1 had the weakest parameters such that the centre peak of /s/ was within the bandwidth of the hearing aid and within the maximum power output cutoff. Setting 4 had the strongest parameters on the scale provided by Phonak Target (v4.1). Figure 26a illustrates the spectral peaks of /s/ across SR1 settings.

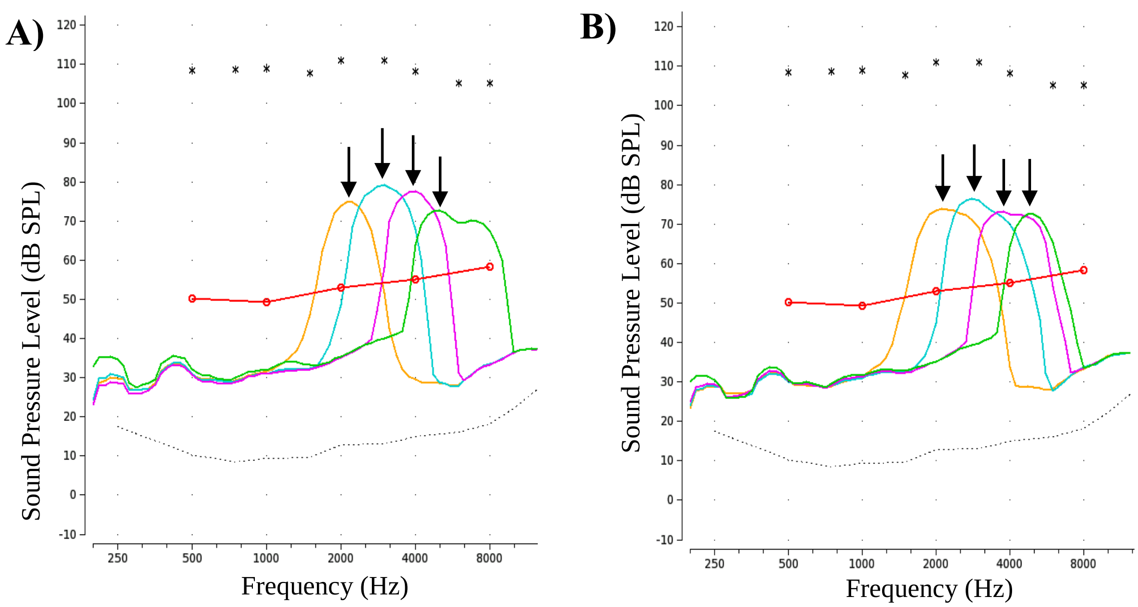


Figure 26. Spectral peaks of /s/ across different SoundRecover settings on Audioscan® Verifit2: A) SoundRecover1; B) SoundRecover2. Green curve indicates setting 1. Pink curve indicates setting 2. Blue curve indicates setting 3. Yellow curve indicates setting 4. Arrows indicate the center peak of /s/.

SR2 settings were chosen by closely matching the center peak of /s/ from the settings of SR1. This was accomplished by moving the slider on the “audibility-distinction” scale within Phonak Target. The “clarity-comfort” scale was set to “a” and was not changed across the different SR2 settings. Figure 26b illustrates the center peaks of /s/ across SR2 settings. Figure 27 illustrates the match between the center peaks of /s/ across the four settings of the SR1 and SR2 processors for control group listeners. Table 5 shows the center peak frequencies of each SoundRecover setting and the parameters for SR1 and SR2. The parameters determined were based on a flat 40 dB HL audiogram. For the hearing impaired listeners, similar procedures were followed, but were implemented for each listener’s individualized hearing aid setting.

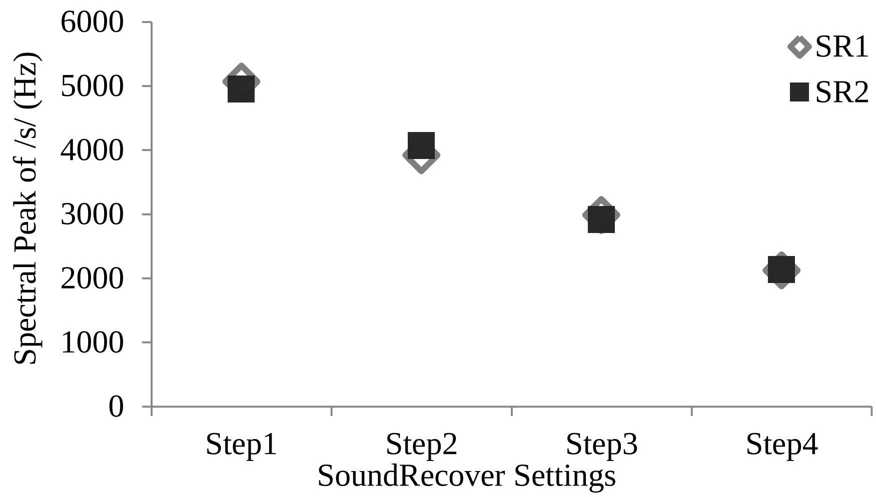


Figure 27. Spectral peaks of /s/ across SoundRecover1 and SoundRecover2 settings.

Table 5. Parameters for SoundRecover1 and SoundRecover2 settings.

SoundRecover phases	**SR1**			**SR2**			
	SP of /s/	**Parameters**		**SP of /s/**	**Parameters**		
		CT	**CR**		**CT1**	**CR**	**CT2**
SoundRecover off	5067	off	off	4953	off	off	off
Step1	5067	6.0	1.5	4953	4.4	1.3	5.8
Step2	3918	3.0	2.4	4069	3.4	1.1	5.1
Step3	2984	1.5	2.0	2917	2.2	1.2	3.9
Step4	2123	1.5	4.0	2139	1.1	1.4	2.5

Notes: SP of /s/ = spectral peak of /s/ in Hz; CT = cut-off frequency (kHz); CR = compression ratio harmonic protection (kHz)

5.2.4 Equipment

Equipment used in the current study was similar to that reported in Mitsuya and Purcell (2016). Participants were seated in a sound attenuated booth (Eckel Industries of Canada, Ontario, Canada; model C26). Participants wore a headset microphone (Shure WH20) and were prompted to speak when the target word appeared on a computer screen at rate of approximately once every four seconds. The microphone signal was amplified with a microphone amplifier (Tucker-Davis Technologies MA3), low pass filtered with a cut-off frequency of 13.5 kHz (Frequency Devices type 901), and digitized at a 28 kHz sampling rate with 18-bit precision (National Instruments PXI-6289M input/output board).

5.2.5 Offline formant analysis

The method for offline formant analysis is the same method reported in Munhall *et al.* (2009). The harmonicity of the power spectrum was used to estimate the vowel boundaries. The boundaries were inspected and corrected if necessary. Vowel formants (F1 and F2) were estimated from the middle 40-80% of the vowel's duration, with a 25 ms window that was shifted in 1 ms increments until the end of the middle portion of the vowel segment. A single average value for each of the formants was calculated from these sliding window estimates. Formant estimates were examined and were relabeled if incorrect (e.g. F2 being labelled as F1) or removed if the formant under examination was well beyond the distribution of other tokens.

5.2.6 Offline spectral mean analysis for /s/

Praat (v6.0.28, Boersma & Weenink, 2017) was used to segment the sibilant /s/ out of the word "see". For each /s/ file, the sound was processed through a first order high pass filter to attenuate low frequencies. Then, Welch's averaged modified periodogram was used for spectral estimation (Welch, 1967) with a window size of 1024 points and a 50% overlap multiplied by the Hamming window. The spectrum was normalized and the spectral mean was calculated according to Forrest *et al.* (1988).

5.2.7 Experimental phases

SR1 and SR2 had four conditions with target words: "head", "hid", "heed", and "see". In total, there were 8 SoundRecover conditions and the order of the SR1 and SR2 processors was randomly assigned to each participant. Participants were given a passage to read with a five minute break to normalize their speech productions after each condition.

In each condition, speakers produced 125 utterances of the target word when a visual prompt was presented. These 125 trials were divided into five experimental phases. In the

SoundRecover off phase (utterances 1-25), participants received normal feedback. After every 25 trials (Step1: utterances 26-50; Step2: utterances 51-75; Step3: utterances 76-100; Step4: utterances 101-125). As each phase progressed, SoundRecover settings were progressively stronger (lower cut-off frequencies and higher compression ratios). A schematic diagram of the experimental phases for SoundRecover1 and SoundRecover2 is illustrated in Figure 28.

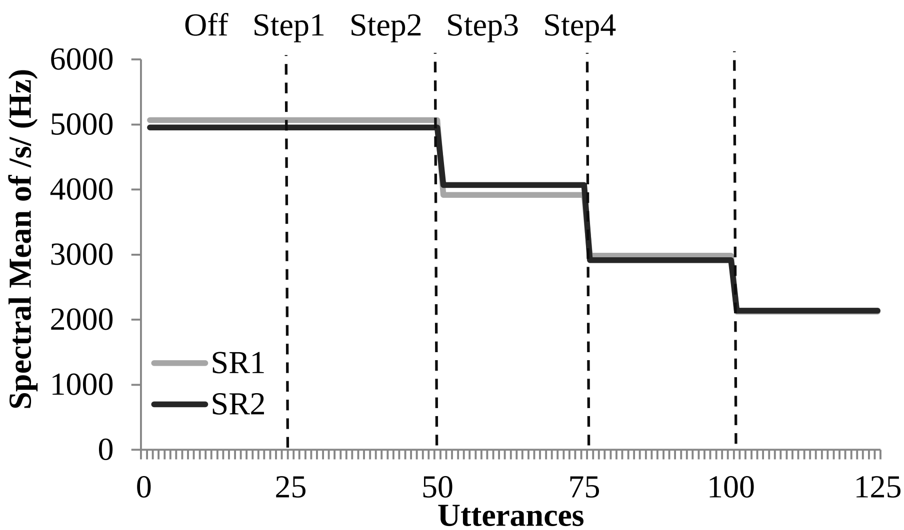


Figure 28. Schematic procedure of SoundRecover settings applied to the spectral peak of /s/. The vertical dotted lines indicate the boundaries of the five experimental phases: SoundRecover Off, Step1, Step2, Step3 and Step4 (from left to right).

5.3 Results

Statistical analyses in this study were completed using SPSS (version 24; IBM, Armonk, NY). The baseline average of F1 was calculated using the utterances made while the participants wore the SoundRecover off setting (i.e. trials 1-25). The F1 values were then normalized by subtracting a speaker's baseline average from each utterance. To quantify a change in F1 production, the average normalized F1 value during each SoundRecover setting was calculated. This analysis process was repeated for F2. Two separate repeated measures ANOVAs were conducted for vowels /ɛ, ɪ, i/ and the sibilant /s/. There were four within-subject factors (SoundRecover Type: SR1, SR2; vowels: ɛ, ɪ, i; formants: F1, F2; SoundRecover settings: Step1, Step2, Step3, Step4) and one between-subject factor (group: older adults with hearing aids, control group with older adults, control group with younger adults). The dependent variable was the change in formant production for vowels or change in spectral mean for /s/. For all statistical analyses, the Greenhouse-Geisser corrected degrees of freedom were used to adjust for lack of sphericity prior to interpretation of effects. Results of the ANOVAs were interpreted at an α of 0.05, with Bonferonni corrections for multiple comparisons.

5.3.1 Vowels

A mixed ANOVA was performed with four within-subject factors (SoundRecover Type: SR1, SR2; vowels: ɛ, ɪ, i; formants: F1, F2; SoundRecover settings: Step1, Step2, Step3, Step4) and one between-subject factor (group: older adults with hearing aids, control group with older groups, control group with younger adults). The main effects of SoundRecover Type, vowels and formants were significant [SoundRecover Type: $F(1.00, 59.00) = 8.90$, $p \leq 0.05$, $\eta_p^2 = 0.13$; vowels: $F(1.81, 106.59) = 8.56$, $p \leq 0.001$, $\eta_p^2 = 0.13$; formants: $F(1.00, 59.00) = 26.18$, $p \leq 0.001$, $\eta_p^2 = 0.31$]. The main effects of SoundRecover settings and group were non-significant [settings: $F(1.55, 91.45) = 4.33$, $p = 0.24$, $\eta_p^2 = 0.07$; group: $F(2.00, 59.00) = 0.09$, $p = 0.91$, $\eta_p^2 = 0.003$]. All remaining interactions were non-significant ($p > 0.05$).

The three way interaction between SoundRecover Type, vowels and group was significant [$F(3.88, 114.35) = 2.90$, $p = 0.03$, $\eta_p^2 = 0.09$]. Post hoc analyses were completed to assess the effect of SoundRecover Type across vowels in the different groups. Figure 29 illustrates the average change in formants across the vowels for each group. In older adults with hearing aids, there were no significant differences between SR1 and SR2 within each vowel. In the control group with older adults, the magnitude of formant changes was greater for SR2 than for SR1 for /i/ ($p = 0.05$). In the control group with younger adults, the magnitude of formant changes was greater for SR2 than for SR1 for /ɛ/ ($p = 0.04$) and /i/ ($p = 0.01$). Overall, these results show that the older adults with hearing aids responded similarly to SR1 and SR2, whereas, the control groups were sensitive to the differences between SR1 and SR2 for some vowels, such as /i/.

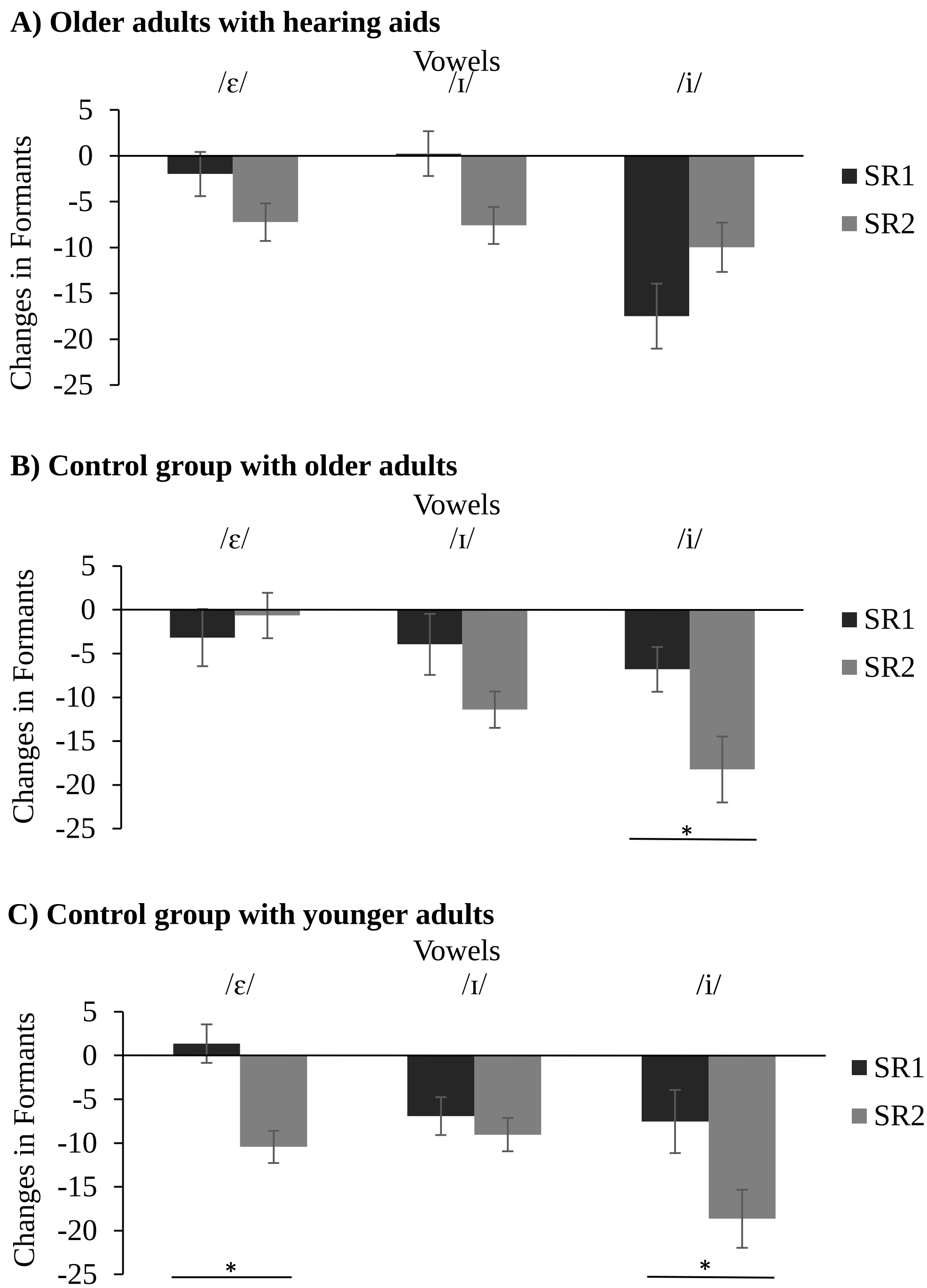


Figure 29. Average changes in formants across vowels for SoundRecover1 and SoundRecover2 for each group: A) older adults with hearing aids; B) older adults; C) younger adults. The error bars indicate ±1 standard error. * indicates significant difference ($p \leq 0.05$).

The three way interaction between formants, SoundRecover settings and group was also significant [$F(3.27, 96.68) = 4.87$, $p \leq 0.001$, $\eta_p^2 = 0.14$]. Post hoc analyses were completed to assess the effect of SoundRecover settings on changes in formants in the different groups. Figure 30 illustrates the average change in formants across the different SoundRecover settings for each group. In older adults with hearing aids, the weaker SoundRecover settings (Step1, Step2 and Step3) had larger F2 changes than F1 changes. In the control group with older adults, the weaker settings (Step1 and Step2) had more F2 changes than F1 changes. In the control group with younger adults, as the strength of SoundRecover increased, the magnitude of F2 change increased, whereas F1 did not. Overall, these results show that F1 did not vary across SoundRecover steps in the different groups and that specific SoundRecover steps elicit different F2 changes across the groups. For example, at Step4, younger adults had the most F2 change compared to both groups of older adults.

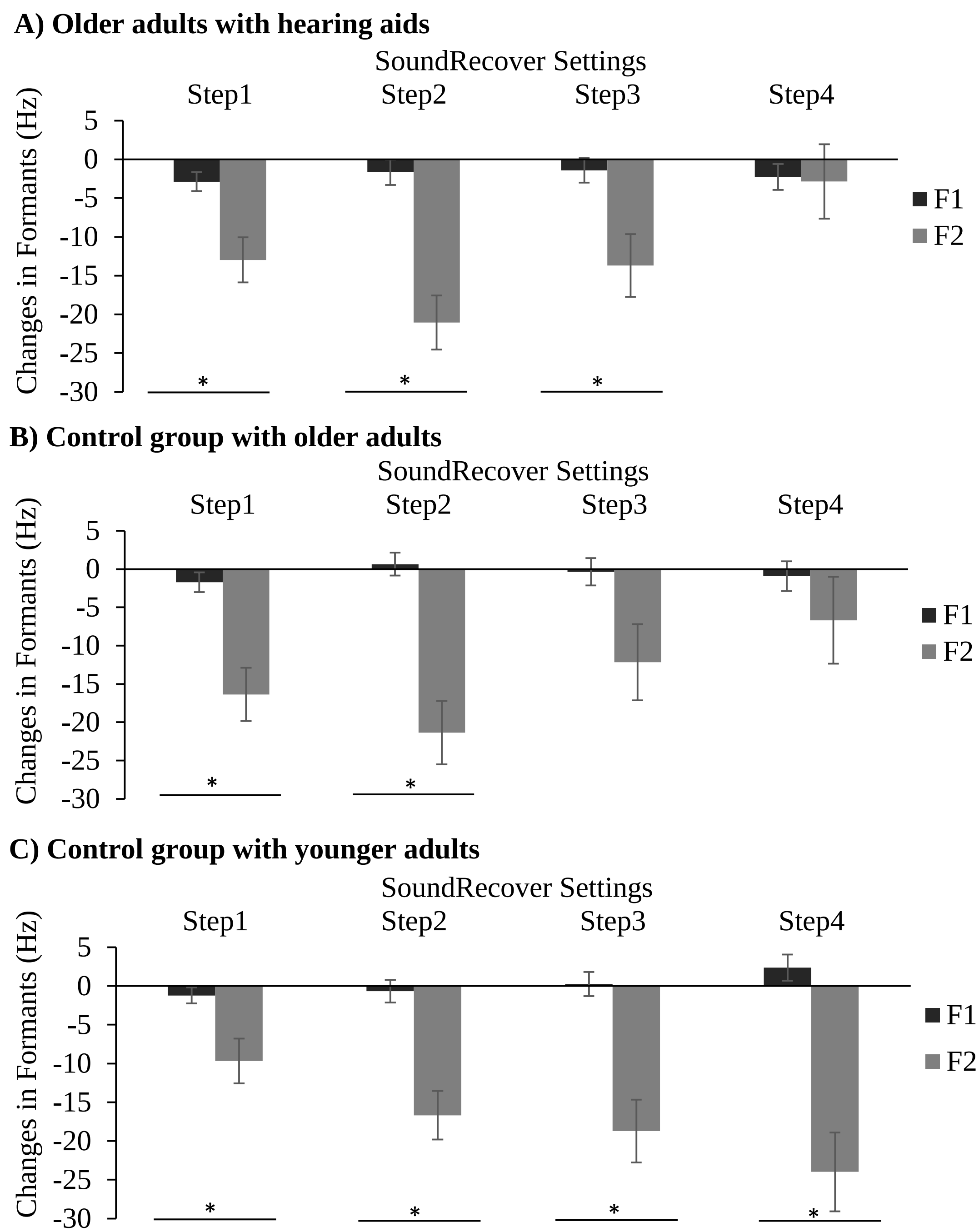


Figure 30. Average changes in formants across SoundRecover1 and SoundRecover2 settings for each group: A) Older adults with hearing aids; B) Older adults; C) Younger adults. The error bars indicate ±1 standard error. * indicates significant difference (*p* $\leq$ 0.05).

The three way interaction between vowels, formants and SoundRecover settings was also significant [$F(3.25, 191.77) = 11.16$, $p \leq 0.001$, $\eta_p^2 = 0.16$]. Post hoc analyses were completed to assess the effect of SoundRecover settings on changes in F1 and F2 in the different vowels. Figure 31 illustrates the average change in formants across the SoundRecover settings for each vowel. In /ɛ/ in "head" and /ɪ/ in "hid", all the steps of the SoundRecover settings had similar F1 and F2 changes, except for Step2. Step2 had a larger mean F2 reduction than was observed for F1 for /ɛ/ ($p = 0.002$) and /ɪ/ ($p = 0.006$). In /i/ in "heed", all the steps had significant differences between F1 and F2, in which F2 had a larger reduction than F1. Overall, when there was a significant difference between F1 and F2 within a SoundRecover setting, the changes in the F2 were larger, and the vowel /i/ had larger F2 mean changes than the other vowels across all SoundRecover settings.

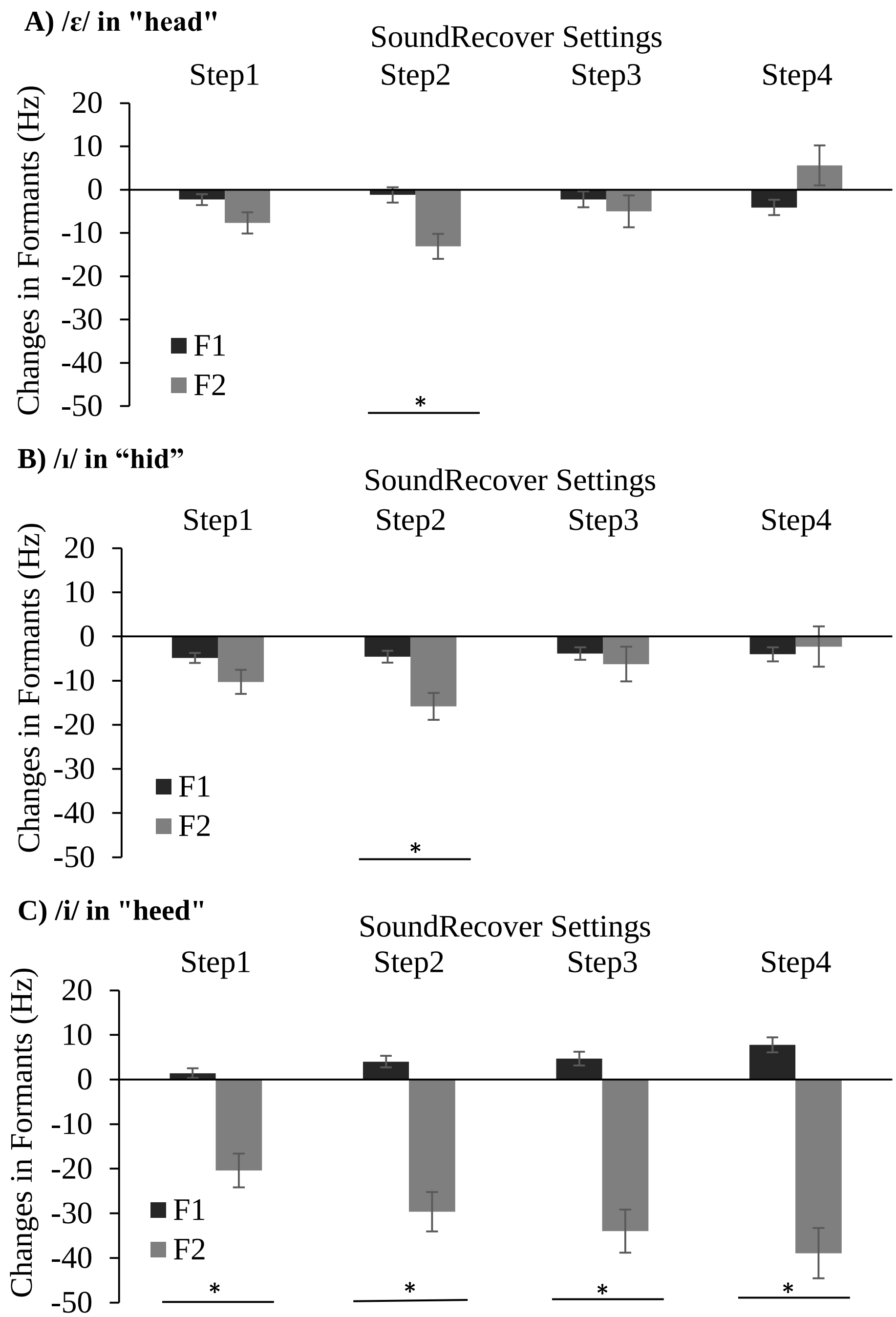


Figure 31. Average changes in formants across SoundRecover settings for each vowel: A) /ɛ/ in "head"; B) /ɪ/ in "hid"; C): /i/ in "heed". The error bars indicate ±1 standard error. * indicates significant difference ($p \leq 0.05$).

The three way interaction between SoundRecover Type, formants and SoundRecover settings was also significant and had the largest effect size of any of the observed interactions in this data set [$F(1.93, 113.81) = 15.81$, $p \leq 0.001$, $\eta_p^2 = .21$]. Post hoc analyses were completed to assess the differences between SR1 and SR2 across settings, collapsed across vowel type, at F1 and F2. Figure 32 illustrates the change in F1 and F2 across SoundRecover settings for SR1 and SR2. There were no significant differences in F1 across SoundRecover settings for SR1 and 2. In comparison, SR2 had more F2 lowering at the strongest two settings (Step3 and Step4) than SR1.

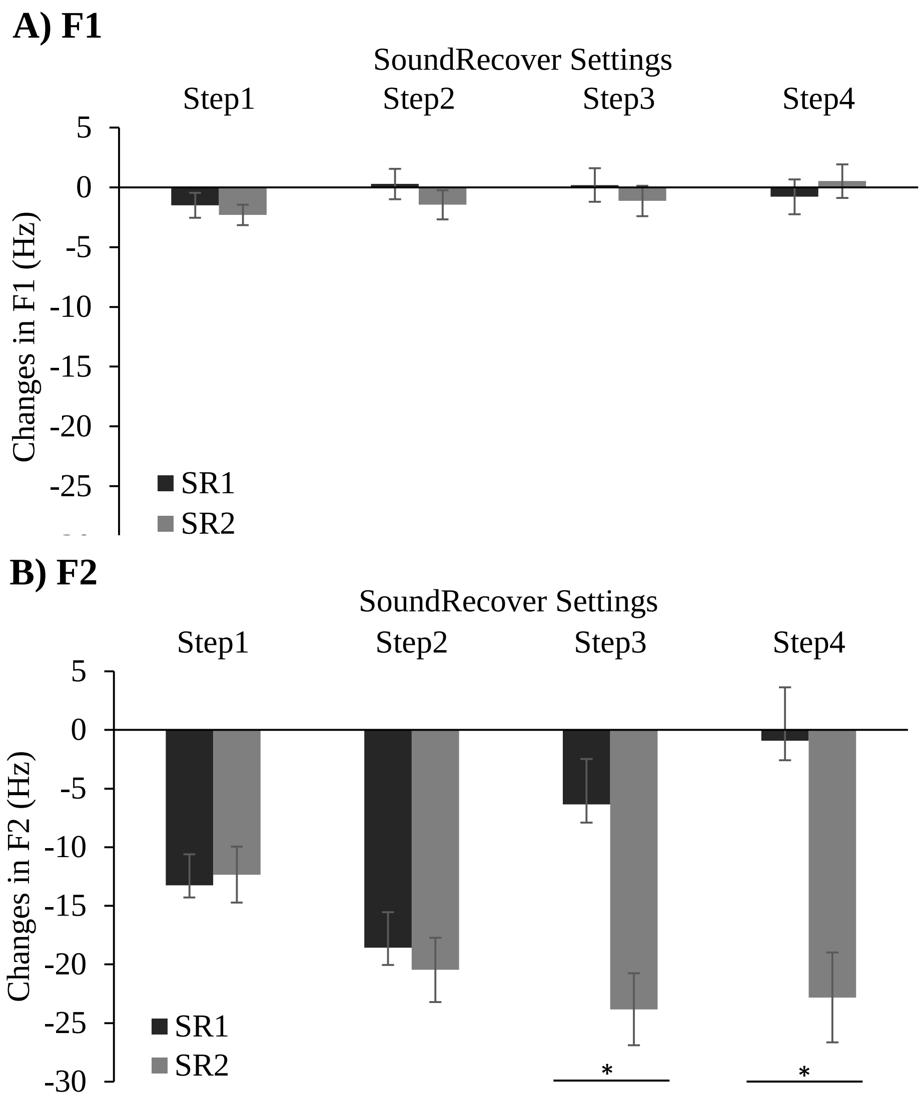


Figure 32. Average changes in formants across SoundRecover1 and SoundRecover2 settings: A) F1; B) F2. The error bars indicate ±1 standard error. * indicates significant difference ($p \leq 0.05$).

5.3.2 Sibilant /s/

A mixed ANOVA was performed with two within-subject factors (SoundRecover type: SR1, SR2; SoundRecover settings: Step1, Step2, Step3 and Step4) and one between-subject factor (group: older adults with hearing aids, control group with older adults, control group with younger adults). The main effects of SoundRecover Type and SoundRecover settings were non-significant [SoundRecover Type: $F(1.00, 59.00) = 3.50$, $p = 0.07$, $\eta_p^2 = 0.06$; SoundRecover settings: $F(2.38, 140.61) = 1.64$, $p = 0.19$, $\eta_p^2 = 0.03$]. The main effect of group was significant [$F(2.00, 59.00) = 7.37$, $p \leq 0.001$, $\eta_p^2 = 0.20$].

The three way interaction between SoundRecover type, SoundRecover settings, and group was significant. [$F(4.87, 143.64) = 3.67$, $p \leq 0.05$, $\eta_p^2 = 0.11$]. Figure 33 illustrates the average magnitude of spectral mean changes across different SR1 and SR2 settings for older adults with hearing aids, control groups with older adults and younger adults. There was no difference in spectral mean change for SR1 and SR2 for the control groups with older and younger adults. There was no difference in spectral mean change for SR2 in older adults with hearing loss. There were, however, significant differences between the steps of SR1 in older adults with hearing aids. Step1 (the weakest setting) had more lowering of spectral mean versus baseline than Step4 (the strongest setting) ($p = 0.03$). Steps 2 and 3 also had larger changes in spectral mean than Step4 (Steps 2 vs. 4: $p = 0.002$; Steps 3 vs. 4: $p = 0.30$).

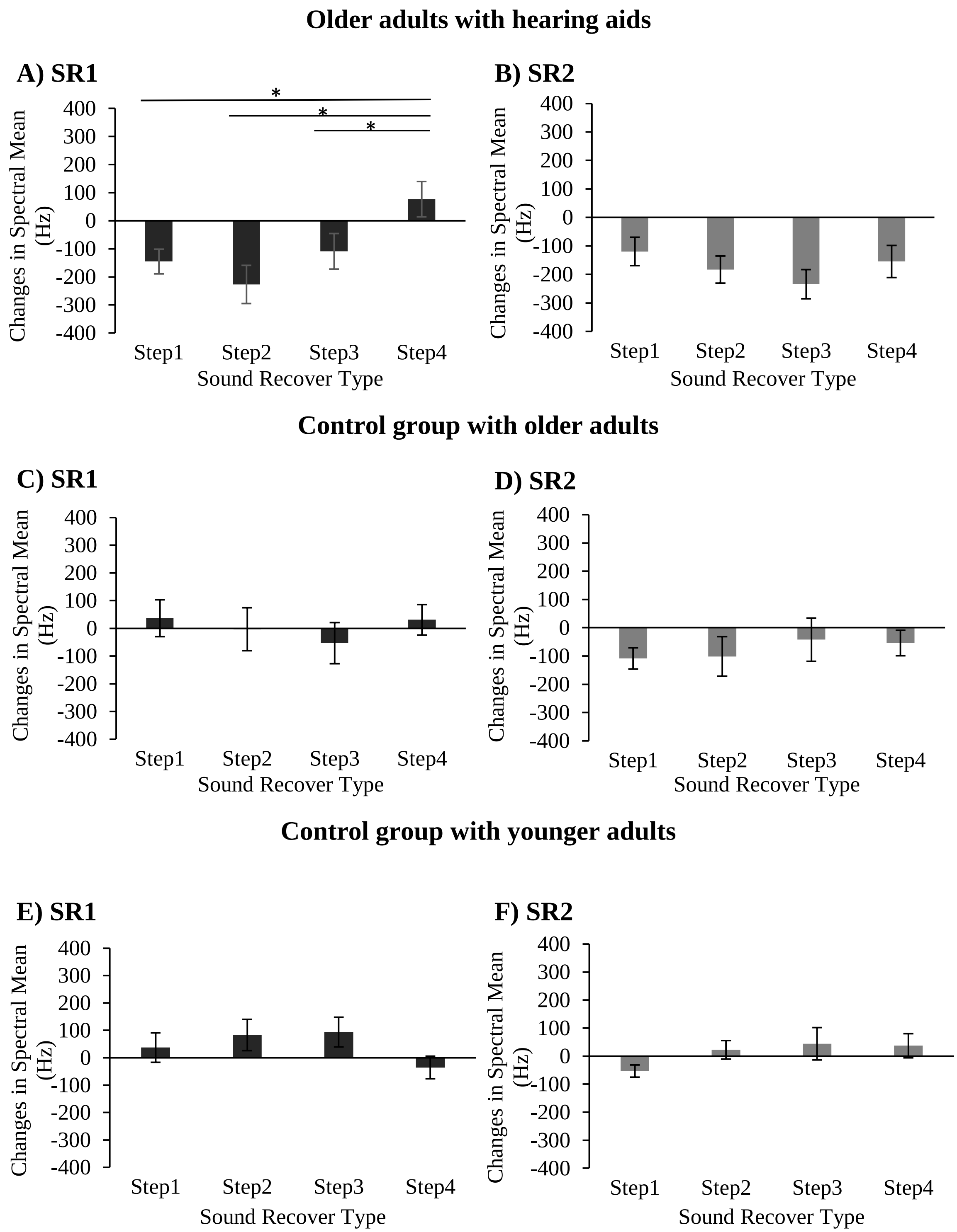


Figure 33. Average changes in spectral mean (Hz) of /s/ across SoundRecover1 and SoundRecover2 settings for each group. The error bars indicate ±1 standard error. * indicates significant difference (*p* ≤ 0.05).

5.4 Discussion

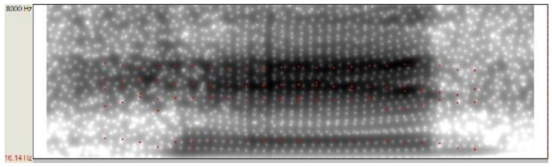
The purpose of the current study was to measure speech production changes between non-adaptive (SR1) and adaptive NLFC (SR2) across different cut-off frequencies and compression ratios settings. To interpret the speech production results of the current study, it was important to determine how speech feedback stimuli changed while listeners wore hearing aids with the different SoundRecover settings. Table 5 in the methods section shows the spectral peak changes of /s/ across different SoundRecover settings. Vowel recordings of an older female voice, who participated in the study, were chosen as stimuli and recorded through the hearing aids used in this study, to illustrate the changed auditory feedback that listeners received during this task. The sample listener was 63 years of age from Southwestern Ontario and had hearing thresholds less than 20 dB HL across all frequencies between 250 to 8000 Hz, bilaterally. Her productions of /ɛ/, /ɪ/ and /i/ during the Baseline phase of the experiment were recorded through hearing aids using a desktop anechoic chamber with recording microphones and a coupler (described in Scollie *et al.*, 2016). Table 6 shows her aided speech, with measured F0 and formant values for /ɛ/, /ɪ/ and /i/ across all SoundRecover settings. Figure 34, 35 and 36 show the aided spectograms of /ɛ/, /ɪ/ and /i/ across all SoundRecover settings, respectively. These illustrate several aspects of SR1 and SR2: F2 is more affected than F1, and is affected more so for the higher-frequency vowel /i/ (Table 6). Also, the formants above the F2 may show more of the effects of these processors, and may carry more of the differences between SR1 and SR2. Particularly for Steps 3 and 4 applied to /i/ in Figure 36, it is evident that SR2 is providing a weaker processing setting, with visible F3 structures that are essentially merged into F2 for the SR1 processor.

Table 6. Fundamental frequency, first and second formants for /ɛ/, /ɪ/, and /i/ across SoundRecover settings

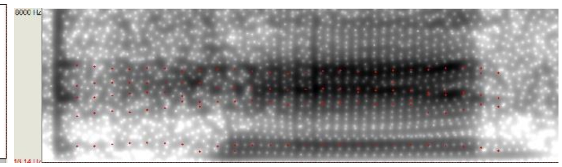
SR settings	/ɛ/			/ɪ/			/i/		
	F0	F1	F2	F0	F1	F2	F0	F1	F2
SR1									
SR1 off	204	585	2193	248	475	2026	204	408	2859
Step 1	204	583	2180	248	474	2001	204	407	2855
Step 2	204	564	1944	248	471	CNE	203	396	2286
Step 3	206	566	1590	248	469	1608	205	395	2000
Step 4	206	550	1284	247	473	1302	204	387	CNE
SR2									
SR2 off	204	584	2189	249	476	2034	204	407	2861
Step 1	203	587	2198	248	478	2090	202	408	2861
Step 2	204	589	2171	247	479	2071	204	414	2799
Step 3	204	570	1603	248	467	1590	204	395	2043
Step 4	206	562	1212	248	472	1242	205	404	1697

Notes: SR = SoundRecover; F0 = fundamental frequency (Hz); F1 = first formant (Hz); F2 = second formant (Hz); CNE = could not extract

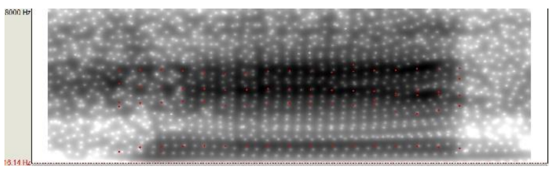
AA) SR1 off



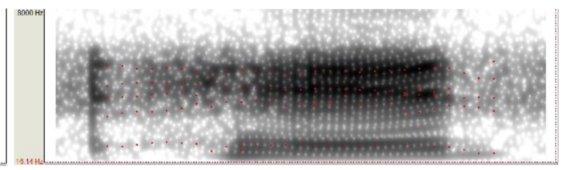
BA) SR2 off



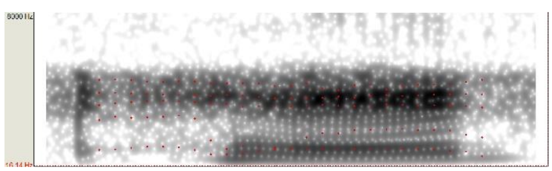
AB) SR1 Step1 (CF: 6.0 kHz, CR: 1.5)



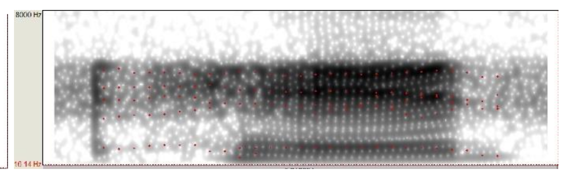
BB) SR2 Step1 (CT1: 4.4 kHz, CT2: 5.8 kHz, CR: 1.3)



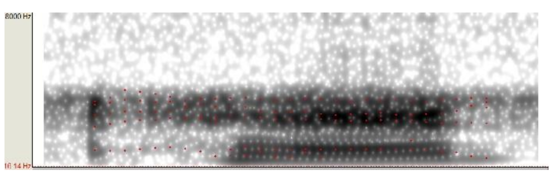
AC) SR1 Step2 (CF: 3.0 kHz, CR: 2.4)



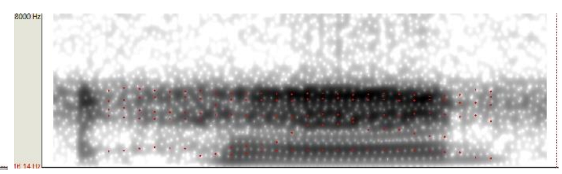
BC) SR2 Step2 (CT1: 3.4 kHz, CT2: 5.1 kHz, CR: 1.1)



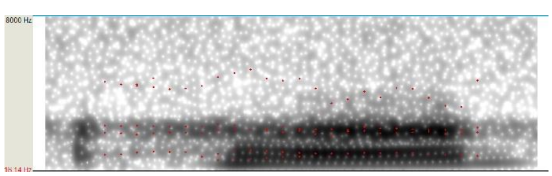
AD) SR1 Step3 (CF: 1.5 kHz, CR: 2.0)



BD) SR2 Step3 (CT1: 2.2 kHz, CT2: 3.9 kHz, CR: 3.9)



AE) SR1 Step4 (CF: 1.5 kHz, CR: 4.0)



BE) SR2 Step4: (CT1: 1.1 kHz, CT2: 2.5 kHz, CR: 1.4)

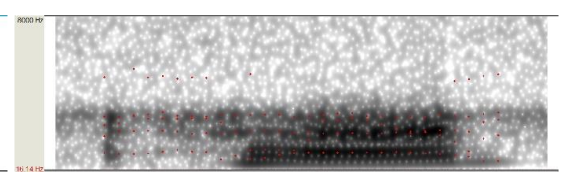


Figure 34. Spectrograms of /ɛ/ across SoundRecover1 (left hand column) and SoundRecover2 (right hand column) settings. CF indicates cut-off frequency for SoundRecover1. CR indicates compression ratio. CT1 and CT2 indicate cut-off frequency 1 and 2 for SoundRecover2, respectively. Red dots within spectrograms are formant tracks from Praat (Boersma & Weenink, 2017).

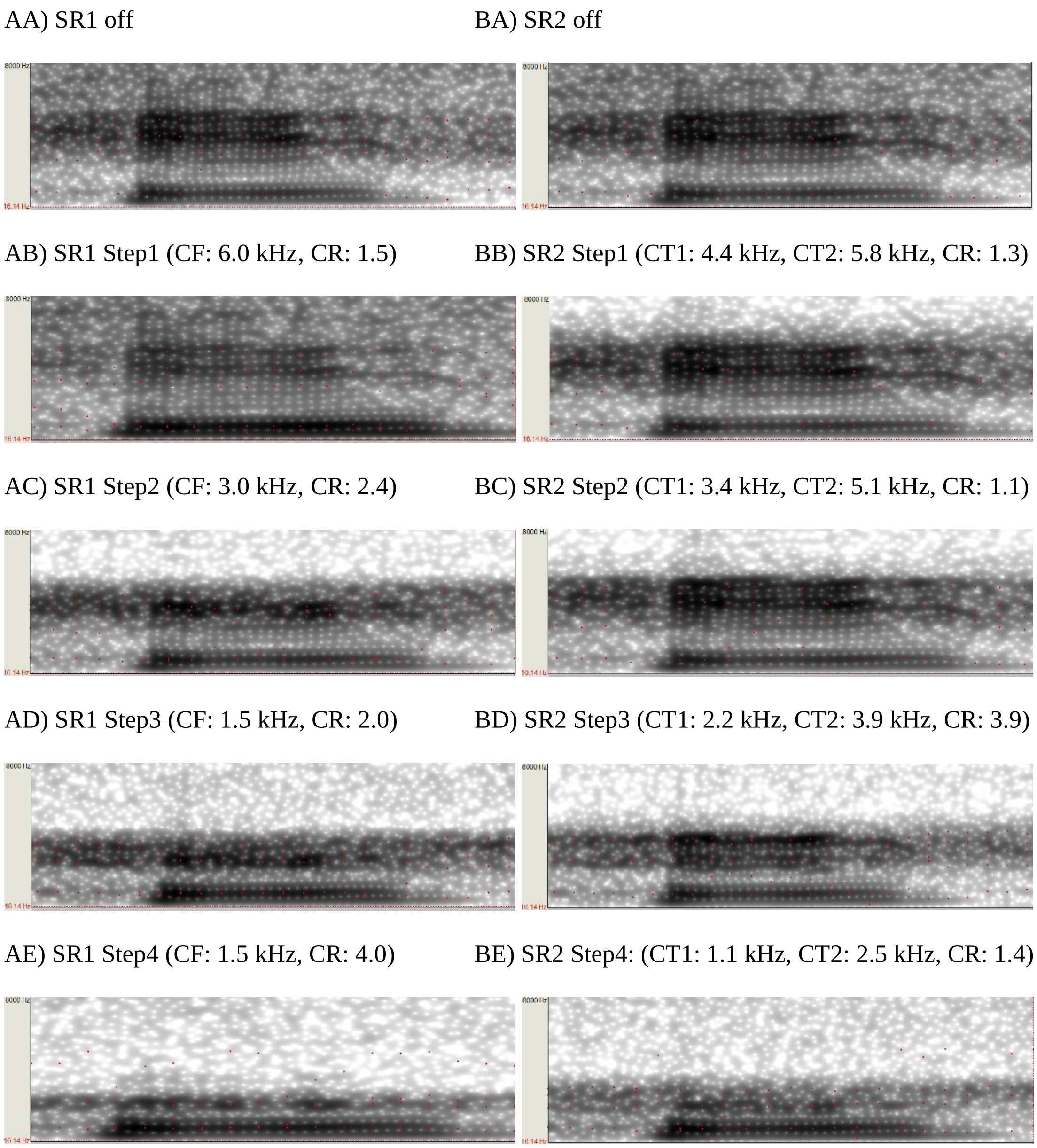


Figure 35. Spectrograms of /ɪ/ across SoundRecover1 (left hand column) and SoundRecover2 (right hand column) settings. CF indicates cut-off frequency for SoundRecover1. CR indicates compression ratio. CT1 and CT2 indicate cut-off frequency 1 and 2 for SoundRecover2, respectively. Red dots within spectrograms are formant tracks from Praat (Boersma & Weenink, 2017).

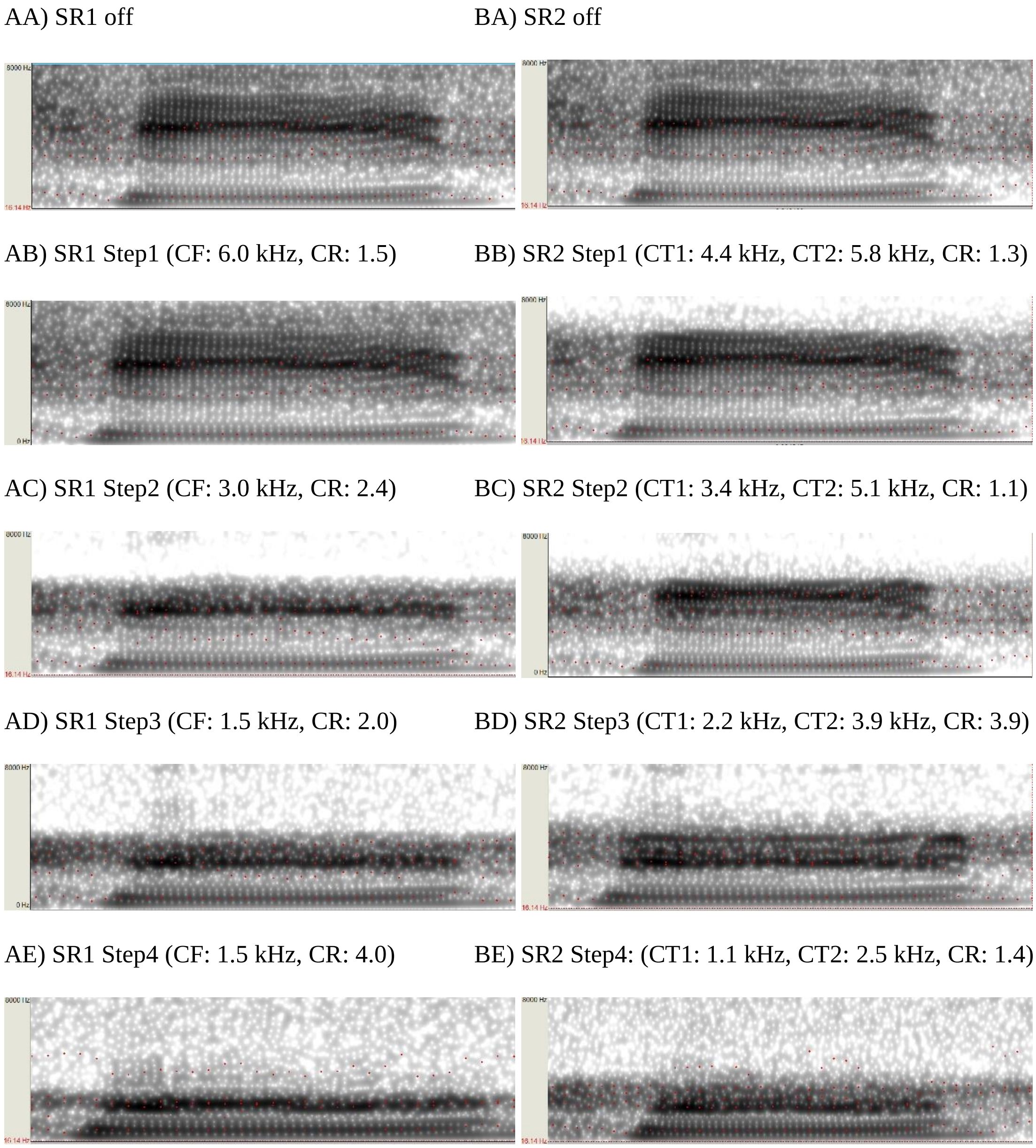


Figure 36. Spectrograms of /i/ across SoundRecover1 (left hand column) and SoundRecover2 (right hand column) settings. CF indicates cut-off frequency for SoundRecover1. CR indicates compression ratio. CT1 and CT2 indicate cut-off frequency 1 and 2 for SoundRecover2, respectively. Red dots within spectrograms are formant tracks from Praat (Boersma & Weenink, 2017).

5.4.1 Preserving low frequency information

One of the goals of NLFC is to maintain low frequency information below the cut-off frequency so that natural formant ratios of vowels, F0 and formants with low values (i.e. F1) are maintained (Wolfe *et al.*, 2010). In Table 6, F0 and F1 of /ɛ/, /ɪ/ and /i/ were similar across all SR1 and SR2 settings. The spectrograms in Figures 34, 35 and 36 also showed that F0 and F1 did not change across SR1 and SR2 settings. This was also consistent with the spectrograms presented in Wolfe *et al.* (2017). Wolfe *et al.* (2017) presented the sentence “my name is asa” in a series of spectrograms with non-adaptive (CT: 1500 Hz, CR 2.1:1) and adaptive (CT1: 1479 Hz, CT2: 3600 Hz, CR 1.4:1) NLFC settings. The spectrograms showed that formant structures below 1500 Hz were preserved for both types of NLFC. Adaptive and non-adaptive NLFC has minimal effects on vowel formants with low frequencies (i.e. F1) and F0.

The results from the present study showed that the participants had minimal changes to F1 productions of /ɛ/, /ɪ/ and /i/ across all SR1 and SR2 settings. This is consistent with the spectral information presented in Table 6 and the spectrograms in Figures 34, 35 and 36. As SR1 and SR2 increased in strength with lower cut-off frequencies and higher compression ratios, F1 did not change, as a result, F1 productions by the participants remained consistent. Even when other formants were changing with different SoundRecover settings, F1 remained consistent. This shows that the speech motor control system can independently regulate and control F1 and F2. This is consistent with the studies by MacDonald *et al.* (2011) and Villacorta *et al.* (2007). Their results found that the manipulations in F1 caused speech changes to F1 but no speech changes to F2. As well, MacDonald *et al.* (2011) manipulated F2 and found significant changes to F2 and minimal changes to F1. Thus, NLFC can preserve low frequency information and cause minimal changes to speech productions that has low frequency information.

5.4.2 Changes to high frequency information

The higher frequency bands of speech were affected by the SoundRecover processor. Table 6 shows the changes in F2 while F1 and F0 remained constant for vowels /ɛ/, /ɪ/ and /i/ across the different parameters for SR1 and SR2. As well, the spectrograms in

Figures 34, 35, and 36 show the changes in the higher frequency areas while the lower frequency regions remained consistent across the different parameters for SR1 and SR2. Similarly, the spectrograms presented in Wolfe *et al.* (2017) showed spectral information above 1500 Hz was distorted when non-adaptive and adaptive NLFC were enabled compared to when NLFC was off. This is consistent with SoundRecover processing, which is aimed mainly at the higher frequency bands of speech and would be expected to affect the high frequencies more than the lower frequencies.

The additional cut-off frequency used by adaptive NLFC, SR2, may be able to preserve higher formants, ratios and other speech cues compared to non-adaptive NLFC, SR1. Figure 26 shows the changes in spectral peak of /s/ for SR1 and SR2. The bandwidth of /s/ at each of the SR1 settings was smaller than SR2. As well, Figures 34, 35, and 36 show how higher formants and formant ratios change with different SR1 and SR2 parameters. As the NLFC processor becomes stronger (lower cut-off frequencies and higher compression ratios), the high frequency information moves to a lower frequency region. The spectrograms for SR1 (left columns in Figures 34, 35, and 36) show the high frequency information were more compressed and formant ratios were not as preserved compared to the spectrograms of SR2 (right columns in Figures 34, 35, and 36). This was also seen by Wolfe *et al.* (2017), where the formant ratios and structures were more preserved in the sentence “my name is asa” for adaptive NLFC than non-adaptive NLFC. As well, Glista *et al.* (2016c) illustrated on the Speechmap of Audioscan® Verifit2 that the peak of the upper formants of /i/ remained in the same location when SR2 was enabled as when SoundRecover was off. However, the bandwidth of the peak was smaller when SR2 was enabled. In comparison, when SR1 was enabled, the peak of the upper formants of /i/ decreased by approximately 1000 Hz compared to SoundRecover off. These illustrative diagrams from the present study, Wolfe *et al.* (2017) and Glista *et al.* (2016c) depict how SR2 preserved formant structures and high frequency speech information better than SR1 by using an adaptive NLFC paradigm with two cut-off frequencies.

The effects of the additional cut-off frequency with adaptive NLFC can be seen at stronger settings of the processor. In Figure 32, SR1 and SR2 did not significantly differ from each other for F2 speech changes at weaker settings (Step1 and Step2). However, at stronger settings (Step3 and Step4), SR1 and SR2 were significantly different. The magnitude of the F2 production change for SR1 decreased with stronger settings, whereas, the magnitude of the F2 production change for SR2 continued to increase with stronger settings. A possible reason for the differences in speech production at the stronger settings could be due to the sound distortions by the compression ratios. SR1 used higher compression ratios than SR2 at the stronger settings. With a higher compression ratio, more compression occurs in the higher frequency region. SR1 may sound more distorted than SR2 at the stronger settings. The speech motor control system may disregard the auditory feedback in SR1 at the stronger settings because it may sound unnatural. As a result, speech production went back to baseline at the stronger settings for SR1. Whereas, with SR2, the speech motor control system continued to respond across all settings. Comparison to the range of perturbations commonly studied in the literature may help shed light on why our listeners responded to SR2 changes but not SR1 at stronger settings. The study by MacDonald *et al.* (2010) manipulated F1 and F2 of /ε/ by gradually increasing F1 up to 360 Hz and decreasing F2 up to 420 Hz. The results found that the magnitude of compensation would asymptote at a certain magnitude of perturbation (~200 Hz for F1 and ~250 Hz for F2) and then decreased in magnitude of compensation with increasing magnitudes of perturbations. MacDonald and colleagues suggested that the speech motor control system uses auditory and proprioceptive feedback to monitor for speech errors. When the formants are auditorily perturbed to a large enough extent, the proprioceptive feedback is incongruent with the perturbed auditory feedback, such that the locations of the articulators may not be possible for the auditory sounds. If the perturbed auditory feedback is too large, the speech motor control system may not rely on the auditory feedback for speech errors. This may be consistent with our findings: it is possible that the perturbation of SR1 produced auditory feedback that cannot be produced within the range of the motor system and was thereby disregarded. In contrast, the SR2 feedback was not disregarded, which may indicate that the SR2 processor maintained vowels within the normal motor production range. Further research is needed to determine this relationship between sound distortions in auditory feedback and speech production, and to determine whether SR2 provides better support for auditory feedback and speech production outside of laboratory conditions.

5.4.3 Group differences: an effect of hearing loss

The results in the present study showed differences in speech production changes between hearing aid users and the control groups with normal hearing individuals. In Figure 29, hearing aid users responded similarly to SR1 and SR2 across the different vowels. However, older adults and younger adults with normal hearing had different responses to SR1 and SR2, especially for /i/ where SR2 had more speech changes than SR1. As well, hearing aid users had around 250 Hz of /s/ production changes across SR1 and SR2 settings, whereas, the control groups with normal hearing individuals had around 100 Hz of /s/ production changes (see Figure 33). This suggests that the perception of the sounds in hearing aid users are different than the control groups with normal hearing to elicit the differences in speech production.

The auditory system is a complex system such that an impairment within the system cannot be easily fixed with amplification devices. Hearing aids cannot restore the auditory system of an individual with hearing loss to be similar to that of an individual with normal hearing. For example, outer hair cells in the cochlea are usually damaged in individuals with hearing loss. This causes the auditory filters in the cochlea to be broader and flatter, which results in a reduction in frequency selectivity (Dubno & Dirks, 1989, Peters & Moore, 1992; Glasberg & Moore, 1986). As well, cochlea damage can also affect loudness and pitch perception, frequency discrimination and/or temporal processing (see Moore (1996) for review). Hearing aids cannot restore outer hair cells or other cochlea damages and hearing aid users will still receive degraded speech input from their hearing aids. This may have caused the hearing aid users to have different speech changes than the control groups with normal hearing individuals. It would be interesting for future studies to use the current paradigm to include unaided conditions to compare the differences in compensation between aided and unaided.

Previous studies have also shown that hearing aid users give different sound quality ratings compared to individuals with normal hearing. A study by Parsa *et al.* (2013) examined the effects of different NLFC parameters on sound quality ratings of music and speech with adults and children with normal hearing and sensorineural hearing losses. Their results demonstrated that individuals with hearing loss rated frequency compressed speech about 5-25% higher sound quality ratings than normal hearing listeners. As well, Glista *et al.* (2016a) had individuals with normal hearing and hearing loss rate sentences that were filtered with different adaptive NLFC settings from “very bad” to “very good”. Their results showed a general trend for individuals with hearing loss to give higher ratings than individuals with normal hearing. These studies show that individuals with hearing loss and individuals with normal hearing perceive NLFC sounds differently which may result in differences in speech production. For instance, in the current study, the normal hearing listeners may perceive the frequency compressed sound as poorer in sound quality than hearing aid users, resulting in less /s/ production changes than hearing aid users. Future studies may want to include a sound quality rating test such that relationships between perceived sound quality and changes in speech production can be determined.

5.4.4 Group differences: an effect of aging

The F1 and F2 production changes across the three groups of talkers suggest that there are aging effects. Figure 30 of the results section shows that all groups had minimal changes to F1 production. However, changes in F2 production between older adults (hearing aid users and normal-hearing groups) and younger adults were different. With younger adults, as SoundRecover increased in strength, the magnitude of F2 change increased. In contrast, in older adults, at stronger SoundRecover settings (Step3 and Step4), the magnitude of F2 changes decreased, such that F1 and F2 production changes were not significantly different. This suggests that the speech motor control system was not responding to larger perturbations in older adults. Further investigations are required to examine whether these aging effects are due to processing changes, changes in the motor speech system, cognitive effects, or other factors.

5.4.5 Changes in speech production follows frequency lowering

In NLFC, the perturbations that are created in auditory feedback are decreasing in frequency, such that high frequency information is lowered to a lower frequency area. It was hypothesized that changes in speech productions would occur in the opposite direction of the frequency lowering. Studies that have manipulated auditory feedback by changing vowel formants (Mitsuya *et al.*, 2015; Purcell & Munhall, 2006; Villacorta *et al.*, 2007), F0 (Burnett *et al.*, 1998; Jones & Munhall, 2000), intensity (Heinks-Maldonado & Houde, 2005; Larson, Sun, & Hain, 2007), and spectral noise of fricatives (Casserly, 2011; Shiller *et al.*, 2009) have shown that compensations to perturbations occurs in the opposite direction of the manipulation. The current results did not show this pattern. All groups of talkers followed the frequency lowering, such that F2 was decreasing by increasing NLFC strength, the F2 productions in response to the frequency lowering also decreased in value. This also occurred with changes in /s/ productions, such that as the spectral peaks of /s/ were lowered by increasing NLFC strength, the spectral mean of /s/ productions also followed the frequency lowering.

Experimental designs may have resulted in differences in compensation patterns. The quality of the perturbation may have influenced speech production changes. In this experimental setup, the participants wore hearing aids and the hearing aids manipulated the auditory feedback by using NLFC. Hearing aids also have other features that may affect the sound quality, such as wide dynamic range compression and attack/release times. In comparison to other perturbation studies, the manipulations were conducted using a computer algorithm and participants wore headphones or insert earphones (Jones & Munhall, 2000; Mitsuya & Purcell, 2016; Shiller *et al.*, 2009). As well, the frequency lowering caused by NLFC has larger perturbations than a typical formant or fricative perturbation study. The NLFC at the strongest setting lowered F2 by approximately 1000 Hz. Whereas, in vowel formant perturbation studies, F1 is usually changed by 200 Hz and F2 is usually changed by 250 Hz (MacDonald *et al.*, 2011). These differences in compensation patterns between NLFC and other perturbation studies suggest that the speech motor control system is sensitive to differences in auditory feedback. Further studies are needed to examine speech production differences using hearing aid and computer algorithms.

5.4.6 Implications, limitations and future studies

The current study demonstrated that non-adaptive and adaptive NLFC maintained low frequency information but differed on how it processed high frequency information. The adaptive NLFC maintained higher formant structures and the high frequency region was not as compressed compared to non-adaptive NLFC, especially at stronger settings. The amount of changes in speech production varied with the strength of the NLFC and hearing aid users responded to each NLFC setting differently. It is possible that the adaptive NLFC processor maintained the auditory path for speech feedback more effectively than the non-adaptive version. Overall, these findings indicate that the NLFC setting and type that clinicians choose may have an effect on the patient’s speech. Future research is needed to determine clinical settings and processor types that would have minimal detrimental effects on speech production, and to extend these findings outside of a laboratory paradigm.

The wide range of participants in the current study imposed limitations. The hearing aid users had different amplification histories. Some of the users used a different manufacturer brand other than Phonak and some were not candidates for frequency lowering. Future studies should include a period of acclimatization so that the participants have experience with the processor and hearing aid sound. Studies by Glista *et al.* (2012), Wolfe *et al.* (2011, 2017) have shown that a minimum of four weeks is needed for acclimatization to the processor. Another limitation is that the hearing aid users varied in degree and configuration of hearing loss. Glista *et al.* (2009) reported greater benefits for NLFC with greater degree of hearing loss. Future studies may want to categorize the group of participants based on hearing loss and have participants that are candidates for NLFC technology.

The current study mainly examined changes to lower formants, F1 and F2, in vowels. F1 and F2 are higher in intensities and can be easily extracted (Peterson & Barney, 1952;

Hillenbrand *et al.*, 1995). As well, F1 and F2 can adequately distinguish different vowels and are the most important for vowel quality (Peterson & Barney, 1952; Potter & Steinberg, 1950). However, higher vowel formants (i.e. third and fourth formants) and formant ratios can be affected by NLFC. Higher formants values are lower in intensity, as a result, are more difficult to accurately determine. The current study also only examined changes to /s/. There are other fricatives that have lower spectral peaks than /s/ and may be affected by NLFC, such as /z/ and /ʃ/ (Stelmachowicz *et al.*, 2002, 2004). Future studies may want to extract higher formants and use other high-frequency phonemes to determine how NLFC changes their spectral information and speech productions.

Overall, NLFC changes auditory feedback by moving high frequency information to a lower frequency range. These changes in auditory feedback resulted in changes in speech production, specifically to speech cues that are higher in frequency, such as F2 in vowels and spectral means of /s/. As the strength of the NLFC processor increased with lower cut-off frequencies and higher compression ratios, more changes in speech production occurred. The NLFC did not affect speech cues that are lower in frequencies, such as F0 and F1 in vowels. There are differences between non-adaptive and adaptive NLFC, such that adaptive NLFC has less compression in the higher frequency region compared to non-adaptive NLFC. These differences resulted in differences in speech production. Further studies are needed to examine the differences between non-adaptive and adaptive NLFC and how speech production changes with the use of NLFC.

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Chapter 6

6 Preliminary analysis of changes in vowel and sibilant /s/ productions after acclimatization to adaptive and non-linear frequency compression

6.1 Introduction

The ability to perceive sounds and hear one's own voice are important to speech production and perception. Speech production is determined by the individual's ability to perceive speech by hearing and auditory speech perception is determined by the individual's ability to produce speech. A hearing loss may affect the person's ability to produce accurate speech and perceive speech from their own voice or within the auditory environment. Other auditory factors that may affect speech production are degree of hearing loss, age when the hearing loss was acquired, the type of amplification and digital signal processes the individual uses (Glista, Scollie, & Sulkers, 2012; Kosky & Boothroyd, 2001; Ozbic & Kogovsek, 2010; Stelmachowicz *et al.*, 2004). Differences in speech abilities are expected between individuals with hearing loss and normal-hearing, however, the magnitude and types of differences may vary. In the literature summarized below, studies have examined the impact of hearing loss and device use on vowels and consonants.

6.1.1 Vowels

Vowels are important for prosodic and segmental features of speech. The accuracy of vowel production is mainly determined by auditory feedback (Ozbic & Kogovsek, 2010). A hearing loss may reduce the amount of auditory feedback and may result in changes in vowel productions. Common reported errors are less differentiation between vowels (Ozbic & Kogovsek, 2010), neutralization towards a central "schwa" vowel (Cowie & Douglas-Cowie, 1983; Ozbic & Kogovsek, 2010; Plant, 1984; Smith, 1975), substitutions of target vowels with neighbouring vowels (Coughlin, Kewley-Port & Humes, 1998; Dorman *et al.*, 1985; Owens, Talbott & Schubert, 1968; Richie, Kewley-Port, & Coughlin, 2003), increased vowel duration (Cowie & Douglas-Cowie, 1983; Plant, 1984), substitution of diphthongs for vowels and nasalization of vowels (Richardson *et* *al.*, 1993). These errors may be due to sensorineural hearing losses being more likely to affect audibility in the high frequencies. As a result, a person with a hearing loss may be able to perceive low, more audible formants, such as fundamental frequency (F0) and first formant (F1), much better than higher and less audible vowel formants like second formants (F2) and third formants (F3). Thus, less errors have been reported with low and back vowels than high, middle and front vowels (Ozbic & Kogovsek, 2010; Smith, 1975).

The vowel production of children with hearing loss is different from that of children with normal hearing. Ozbic and Kogovsek (2010) compared vowel formant values between children with normal hearing and children with severe and profound hearing losses between the ages of 5 to 11 years. Their results revealed that children with hearing loss had less differentiation of vowels and a more centralized vowel space. The F1 and F2 formant ranges were reduced and the standard deviations were larger in children with hearing loss compared to children with normal hearing. The results also found that vowels that had clear visual information, such as the jaw opening of /a/ and /i/, had smaller standard deviations than vowels with minimal visual information for children with hearing loss. The children with hearing loss mostly differed from the children with normal hearing in the F2 ranges and in the degree of overlap between vowel categories. There were more errors in the high, middle and front vowels than low and back vowels. In addition, the vowel spaces of children varied by degree of hearing loss. Specifically, the vowel space of the group with severe hearing loss was more differentiated than the profound HL group but less than children with normal hearing.

Case studies have also revealed the impact of hearing loss on speech production. For example, Plant (1984) found that a 17 years old male deafened at age 11 produced vowels in isolation with normal ranges. However, during spontaneous speech, the male subject produced /ə/ more frequently with a tendency towards centering most vowels. Similarly, Smith (1975) found that children with severe to profound hearing loss produced low, central vowels more correctly and there was tendency for all vowels to drop to a more neutral position. These vowel production errors in children are similar to adults with hearing losses. Studies that examined vowel production in postlingually deafened speakers found increased vowel durations, vowel reductions, decreased spectral contrast distances among vowels and increased vowel dispersion in the formant space (Cowie & Douglas-Cowie, 1983; Lane & Webster, 1991; Plant, 1984).

6.1.2 Fricatives

In comparison to vowels which have low and high frequency formants, fricatives such as /s/ and /ʃ/ contain high frequency content, and as such have been studied extensively in the literature related to hearing loss and speech production. Sibilants are differentiated by differences in their distribution of energy in their spectra, such as spectral mean, skewness and kurtosis (Ghosh *et al.*, 2010). For example, the centroid of the spectrum of /s/ is higher than that of /ʃ/ (Ghosh *et al.*, 2010). To differentiate between sibilants, an individual may have to rely on their ability to hear and use cues of frequency distribution and level in the high frequencies, yet listeners may have the greatest hearing loss in the frequency regions that carry the primary cues (Stelmachowicz *et al.*, 2004). The perception and production of fricatives are susceptible to deterioration following a hearing loss (Lane & Webster, 1991).

The perception of /s/ is important in semantic and syntactic language development, such as plural markers (e.g. cow vs. cows) and verb tense (e.g. jump vs. jumps). Children with hearing loss may have difficulty perceiving high frequency fricatives, which may delay their language development. A set of studies, summarized by Stelmachowicz *et al.* (2004), explored the role of auditory experience in early phonological, linguistic and morphological development. They had an early identification (EI) group that was aided by 12 months of age, a late identification (LI) group that was aided after 12 months of age and a group with children with normal hearing. The results found that the EI group had marked delayed in the acquisition of all phonemes, with the shortest delay for vowels and longest delays for fricatives. The acquisition for the LI group was substantially longer than the early EI group. Similar results were found by Moeller and colleagues (2007) in which children who received amplification before 6 months of age had significantly delayed phonological development for fricatives, despite acquiring other classes of speech sounds later than, but at a rate similar to, children with normal hearing. The delay in fricative production is consistent with the notion that these children may have insufficient access to the high frequency components of speech due to the limited bandwidth of conventional hearing aids, sensorineural hearing loss in the high frequency region or reduced audibility in contexts of noise and reverberation.

The misarticulation of fricatives can be seen across a variety of hearing losses in children. Elfenbein, Hardin-Jones and Davis (1994) found that misarticulation of fricatives and deletion of /s/ as an inflectional morpheme were common, particularly for children with three pure tone average thresholds greater than 45 dB Hearing Level (HL). They also found that children with mild hearing losses exhibited misarticulation of fricatives and demonstrated corresponding semantic and syntactic errors in morphology such as plural markers and verb tense. The similarities among fricative errors for children with mild to severe hearing loss indicate that production of fricatives is affected by any degree of hearing loss in the frequencies associated with the acoustic spectrum of fricative sounds.

The production of fricatives in adults with hearing loss can also be impacted. Lane and Webster (1991) had three post-lingually deafened adults from 1.5-6 years read the Rainbow Passage and the Phonetic Inventory Sentences. The results found that deafened adults were articulating palatovelar fricatives with a more front place of articulation and there was less differentiation between fricatives in comparison to normal hearing listeners. Plant (1984) had a 17 years old male who was deafened at age 11 read the Rainbow passage. He tended to omit word final /s/. Overall, a hearing loss can increase perception and production errors in fricatives and vowels. This may impair spoken language development in children with hearing loss and deterioration of production and perception in adults with hearing loss.

6.1.3 Hearing aids, high frequency audibility, and speech

As summarized above, a hearing loss can increase production errors in vowels and fricatives. This may impair spoken language development in children with hearing loss and degradation of production in adults with hearing loss. Individuals with hearing loss may use amplification devices to provide audibility. Effects of these devices can be seen through changes in speech production and perception abilities. However, some of these individuals may need additional digital signal processing features in the hearing instruments to provide audibility. Some of the additional features that are recommended are the use of extended bandwidths or frequency lowering technology (Stelmachowicz *et al.*, 2002, 2004). The extension of bandwidth into the high frequencies may constrain low frequency amplification, output power of the hearing aids, increase distortion, create hearing aid feedback or create subject discomfort (McCreery *et al.*, 2012; Turner & Cummings, 1999). As a result, frequency lowering technology is a solution to avoid the effects of limited bandwidths. The goal of frequency lowering technology is to provide audibility to high frequency information regions through moving high frequency sounds to a lower frequency range where audibility is more likely (Kuk *et al.*, 2009; Wolfe *et al.*, 2010).

Non-linear frequency compression (NLFC) is one of the types of frequency lowering technology available in current commercial hearing aids. With NLFC, inputs above a cut-off frequency are compressed by a specified ratio so that high frequency inputs are shifted to a lower frequency range where audibility is more likely to be achieved. Inputs below the cut-off frequency are not compressed and do not overlap with the compressed region to preserve formants and formant ratios (Wolfe *et al.*, 2010). Reviews of older NLFC technology are found in Auriemmo *et al.* (2009), Simpson (2009) and McCreery *et al.* (2012).

Phonak, a hearing aid manufacturer, uses NLFC in their SoundRecover program. In their older version of SoundRecover, SoundRecover1 (SR1), their NLFC is non-adaptive. The cut-off frequency and compression ratio is constant for all incoming signals (Glista *et al.*, 2016b). In their newest version of SoundRecover, SoundRecover2 (SR2), is adaptive as the cut-off frequency changes based on the energy distribution of the incoming signal (Glista *et al.*, 2016b). In SR2, there are two cut-off frequencies, CT1 and CT2, in which CT1 has a lower cut-off frequency and CT2 has a higher cut-offcut-off frequency. SR2 will rapidly analyze the incoming signal and will switch between CT1 and CT2 depending on the energy distribution. If the spectrum of the incoming signal consists of mostly high frequency content, CT1 is used. In contrast, if the spectrum of the incoming signal is mostly low frequency dominant, CT2 is used. With a higher cut-off frequency for low frequency stimuli, it reduces the effects of NLFC where NLFC may not be needed to improve audibility. Both versions of SoundRecover have the common goal to preserve low frequency information (i.e., vowels) and increase audibility in high frequency through NLFC. However, SR2 may be better able to reduce distortions caused by NLFC due to its adaptive behaviour. The current literature on NLFC mainly studies the effect of SR1 on speech perception and production.

The benefits from NLFC technology may vary across adult and children who use hearing aids. Glista and colleagues (2012) evaluated changes in speech perception abilities, such as speech detection of /s/ and /ʃ/, /s-ʃ/ discrimination and plurals and consonant recognitions in children between the ages of 11 to 18 years during 16 weeks of acclimatization to NLFC. The findings showed that acclimatization to NLFC varied across children. Some children showed benefit in the beginning and speech perception scores remained consistent across the acclimatization period. Other children showed a gradual improvement in speech perception abilities. There were also children who received little benefit from the NLFC. Similarly, studies by Wolfe and colleague (2010, 2011) evaluated the use of NLFC in children with moderate to moderately-severe hearing loss. The results showed that audibility and recognition of high frequency speech sounds increased with NLFC use. After six months of acclimatization to NLFC, recognition of speech sounds in quiet improved. Whereas, after several weeks to several months of acclimatization, recognition of speech sounds in noise improved. The studies by Simpson, Hersbach and McDermott (2005) and Glista *et al.* (2009) showed improvements in speech perception abilities with NLFC in adults with hearing loss.

There are studies that found NLFC provided limited benefits but was not detrimental in the hearing aid user’s speech perception abilities. Simpson, Hersbach and McDermott (2006) found that their adult hearing aid users subjectively perceived benefit from NLFC in quiet and noisy situations, however, no significant differences in speech recognition scores were found between NLFC disabled or enabled. They also found that if the acclimatization to NLFC was incomplete, it may cause confusion among fricatives. Picou, Marcrum and Ricketts (2015) measured consonant recognition in quiet, consonant discrimination threshold in quiet, sentence recognition in noise and sound quality for speech and music in adults with mild to moderate hearing loss after 3-4 weeks of NLFC acclimatization. They found no differences between NLFC enabled and conventional hearing aids. However, they did find one benefit with NFLC enabled: thresholds were better for /s/ discrimination. Likewise, Hillock-Dunn and colleagues (2014) found no differences between NLFC enabled and disabled in phoneme and spondee identification in quiet and noise in children between the ages of 9 to 17 years. However, children with greater difference in audible bandwidth between NLFC enabled and disabled were more likely to demonstrate improvements in high-frequency consonant identification in quiet and spondee identification in noise. Other studies by McDermott and Henshall (2010) and Perreau, Bentler, and Tyler (2013) showed no significant differences between NLFC enabled and disabled on speech perception measures.

NLFC should have minimal effects on vowel production and perception as frequency bands below the cut-off frequency are not affected by the compression ratio. Glista *et al.* (2009) and Simpson *et al.* (2005) found no significant differences in vowel identification and vowel phoneme scores between NLFC enabled or disabled. Perreau *et al.* (2013) found that vowel perception abilities in quiet using NLFC were better than conventional amplification. However, Perreau *et al.* (2013) suggests that if the cut-off frequency in NLFC is set too low (e.g. 1500 Hz) and a high compression ratio is used, higher formants such as F2 and F3 might be too severely compressed or reduced and spectral smearing of the input signal might occur. Further, Alexander (2016) found that vowel recognition decreased at lower cut-off frequencies (less than 2200 Hz) because F2 shifted to another frequency area. Thus, Perreau *et al.* (2013) and Alexander (2016) suggest using a higher cut-off frequency and a lower compression ratio to prevent negative impacts on vowel production and perception.

Preliminary results have shown there are differences between non-adaptive and adaptive NLFC. Wolfe *et al.* (2017) evaluated audibility and speech recognition abilities in children with high frequency hearing loss using non-adaptive and adaptive NLFC in Phonak hearing aids. Their results found after four to six weeks of acclimatization to adaptive NLFC, the children had better plural detection and word recognition scores than with non-adaptive NLFC. However, there were no differences between the two types of NLFC with phoneme recognition scores and detection thresholds. Glista *et al.* (2016a) presented two case studies that acclimatized to adaptive and non-adaptive NLFC. The results found that there was a benefit to using NLFC compared to when NLFC was turned off. As well, when there was a difference between adaptive and non-adaptive NLFC, the adaptive NLFC had better scores. Based on the two studies that compared adaptive and non-adaptive NLFC, when there was a difference between the two types of NLFC, adaptive NLFC showed greater benefit.

In summary, there are some improvements in speech perception abilities with NLFC. However, there are limited studies that measured changes in vowel and fricative production associated with NLFC use. The purpose of the current study was to measure changes in vowel and fricative production after acclimatization to SR1 and SR2 in hearing aid users. We hypothesized that differences in vowel production after acclimatization to SR1 and SR2 may occur. Further, we hypothesized that SR2 may be able to protect and conserve lower frequency regions better, and that this could lead to better vowel production with SR2. Finally, we expected an improvement in fricative production after acclimatization for both types of SoundRecover.

6.2 Method

6.2.1 Participants

Recruitment of four participants ages 9-24 years took place at the H.A. Leeper and Speech and Hearing Clinic in London, Ontario, Canada and within a participant database. To be included in the study, participants were required to have a bilateral sensorineural hearing loss, sloping to at least a moderately severe high-frequency pure-tone-average (HF-PTA) hearing level averaged across 2000, 3000 and 4000 Hz. Participants were required to be at least six years of age and full-time users of digital behind-the-ear (BTE) hearing aids prior to entering the study. All participants were assessed as full-time hearing aid users (i.e. achieving continuous hearing aid usage during school or waking hours) prior to beginning data collection. Participants had to have English as a first language, be in good, general health and could not be enrolled in speech therapy during the duration of the study.

Hearing threshold testing was measured with a Grason-Stadler 61 audiometer in a doubled walled sound treated booth. Pure-tone air conduction thresholds were measured bilaterally at all octave and interoctave frequencies between 250 and 8000 Hz using Etymotic Research ER-3A insert earphones coupled to the participant's personal earmolds. The threshold equalizing noise (TEN-HL) test and interpretation of the results was used to determine cochlear dead regions (Malicka, Munro, & Baker, 2010; Moore, 2004; Moore, Glasberg, & Stone, 2004). Table 7 is a summary of participant's case history information, audiometric assessment and TEN test results. Participants are listed from least to greatest HF-PTA in the left-ear.

Table 7. Summary of case histories and audiometric assessments

Case	**Age**	**Sex**	**Previous HA make**	**Right**			**Left**		
				PTA (dB HL)	**HF- PTA (dB HL)**	**DR (kHz)**	**PTA (dB HL)**	**HF-PTA (dB HL)**	**DR (kHz)**
1	9	M	Phonak	65	82	None	67	83	None
2	11	M	Phonak	70	78	None	70	78	None
3	10	F	Oticon	45	70	4.0+	75	97	2.0+
4	24	M	Phonak	97	118	2.0+	100	120	1.5+

Notes: M = male; F = female; HA = hearing aid; PTA = pure-tone-average (.5, 1, & 2 kHz); HF-PTA = high frequency-PTA (2, 3, & 4 kHz); DR = dead regions

6.2.2 Hearing aid fitting

Each participant was provided with bilateral study hearing aids: Phonak Naida Q90-SP behind-the-ear hearing aids coupled to their own personal earmolds. Hearing aid fitting was conducted within the Audioscan® Verifit1 (Audioscan, Dorchester, ON, Canada) and followed protocols from the Desired Sensation Level (DSL) method v5.0 for pediatrics (Bagatto *et al.*, 2005; Scollie et al, 2005). The gain and features of the hearing aids were held constant throughout the study. The microphone setting of the hearing aids were set to Phonak's RealEar Sound configuration. The digital noise reduction, volume control, and automatic program features were disabled. A data-logging feature that tracked hearing aid usage over the duration of the study was enabled.

A coupler-based verification strategy was used to reduce the impact of acoustic feedback and room noise/reverberation effects during real-ear verification procedures. This also allowed for replicable measures across repeated fitting appointments. The output of the hearing aids was matched to prescriptive targets that incorporated the participant's real-ear-to-coupler difference values at input levels of 55, 65, 75 for digitized speech passages and a 90 dB SPL pure tone signal found in the Audioscan® Verifit1. Hearing aid gain was adjusted using a research version of Phonak Target v4.1 programming software to best possible match to DSL targets. Minor adjustments to the gain of hearing aid were also completed when there were specific concerns by the participant and were only done at the beginning of the study.

Following the initial fitting, frequency compression parameters for SR1 and SR2 were verified and individually adjusted according to an established protocol by Scollie and colleagues (2016). The amplitude compression and gain from the initial fitting were held constant for the SoundRecover fittings. Frequency compression settings were determined for each ear. The maximum audible output frequency (MAOF) region was determined such that it was between the peak-defined and the long term average speech spectrum-defined limits of audibility. The stimulus /s/, available in Audioscan® Verifit1, was fitted within the upper shoulder of the MAOF region. This maximized /s/ audibility was programmed to the weakest possible strength for SR1 or SR2. The location of /s/ within the SR1 and SR2 settings were matched so that the two processors were comparable. Tables 8 and 9 show a summary of the SR1 and SR2 settings, respectively.

Table 8. Summary of SoundRevover1 settings

Case	Right		Left	
	Cut-off Frequency (Hz)	**Compression Ratio**	**Cut-off Frequency (Hz)**	**Compression Ratio**
1	4160	2.1	4160	2.1
2	4320	2.5	4320	2.5
3	4000	2.8	2880	3.0
4	1440	4.0	1400	4.0

Table 9. Summary of SoundRecover2 settings

Case	Right			Left		
	CT1 (Hz)	**CT2 (Hz)**	**CR**	**CT1 (Hz)**	**CT2 (Hz)**	**CR**
1	1120	4640	1.6	1120	4640	1.26
2	1120	4480	1.3	1120	4480	1.3
3	800	4160	1.43	800	4800	1.28
4	320	1920	1.67	160	1920	1.88

Notes: CT1 = cut-off frequency1; CT2 = cut-off frequency2; CR = compression ratio

6.2.3 Speech production measures

Participants completed two speech production tasks: vowels and plurals. In the vowel task, three tokens of each vowel /i, ɪ, e, ɛ, æ, ɔ, u/ were collected in /hVd/ context (“heed”, “hid”, “hayed”, “head”, “had”, “hawed”, and “who’d”, respectively). In the plurals task, two tokens of each word (“crabs”, “pigs”, “ants”, “books”, and “cups”) were collected. These words were taken from the UWO Plurals Test (Glista & Scollie, 2012). Words were presented live by the examiner within a carrier phrase “Say the word _____”. Two different lists of vowels and plurals were created such that the words were randomly presented. The researcher randomly chose a vowel and plurals list for each testing session. The participant was unable to see the list or the examiner’s face during the task. Participants were asked to repeat the target word in the carrier phrase, if he/she said it incorrectly.

6.2.4 Equipment

Participants were seated in a double walled sound attenuated booth and spoke into a studio grade AKG condenser microphone (C 4000 B). The microphone was connected to a preamplifier, analog-to-digital converter and desktop computer. The microphone recordings were digitized at a sampling rate of 44100 Hz sampling rate, using 24-bit resolution. The SpectraPLUS FFT Spectral Analysis System was used to store the recorded sound files.

6.2.5 Offline formant analysis

The method for offline formant analysis is the same method reported in Munhall *et al.* (2009). The harmonicity of the power spectrum was used to estimate the vowel boundaries. The boundaries were inspected and corrected if necessary. Vowel formants (F1 and F2) were estimated from the middle 40-80% of the vowel's duration, with a 25 ms window that was shifted in 1 ms increments until the end of the middle portion of the vowel segment. A single average value for each of the formants was calculated from these sliding window estimates. Formant estimates were examined and were relabeled if incorrect (e.g. F2 being labelled as F1) or removed if the formant under examination was well beyond the distribution of other tokens.

6.2.6 Offline spectral mean analysis for /s/

Praat (v6.0.28; Boersma & Weenink, 2017) was used to segment the sibilant /s/ out of the plural words. For each /s/ file, the sound was processed through a first order high pass filter to attenuate low frequencies. Then, Welch's averaged modified periodogram was used for spectral estimation (Welch, 1967) with a window size of 1024 points and a 50% overlap multiplied by the Hamming window. The spectrum was normalized and the spectral mean was calculated according to Forrest *et al.* (1988).

6.2.7 Study design and data collection sequence

A single-subject design, similar to Glista, Scollie and Sulkers (2012) was used to evaluate a significant change at the level of the individual. In single-subject design, significant changes in performance at the individual level can be determined because each participant serves as their own control (Gast, 2010). This design was used because the participants in this area of clinical practice are highly unique: (1) they vary in age and onset of hearing loss; (2) they vary in degree and configuration of hearing loss and hearing aid settings. As well, past studies that measured performance with frequency lowering hearing aids (Simpson, 2009) have reported large between-subject variability. For these reasons, a single-subject design was considered appropriate, because group mean trends would likely not reflect individual outcomes.

The experiment was sectioned into two phases: SR1 and SR2. The two phases were counterbalanced across the participants and participants were blind to the conditions. Each phase consisted of two parts: pre- and post-testing. Between testing sessions was an acclimatization period of around six weeks (see Table 10 for acclimatization times). Six weeks of acclimatization time were chosen because Glista and colleagues (2012) found frequency compression benefit after six weeks. Acclimatization period varied slightly between and within participants because of scheduling factors and/or illness. Participants were compensated with a $10 gift card for every hour of testing and allowed to keep the study-worn aids for their participation in the study.

Table 10. Summary of SoundRecover1 and SoundRecover2 testing orders and acclimatization times

Case	Condition 1	Acclimatization Time (weeks)	Condition 2	Acclimatization Time (weeks)
1	SR2	9.3	SR1	6.0
2	SR1	6.0	SR2	6.4
3	SR1	7.0	SR2	5.9
4	SR2	7.9	SR1	9.0

Notes: SR1 = SoundRecover1; SR2 = SoundRecover2

6.3 Results

6.3.1 Analysis Strategies

Indices of vowel production change between hearing aid conditions included (a) change in F1 for each vowel, (b) change in F2 for each vowel, (c) change in vowel space area, and (d) change in vowel space shape.

Change in formant production. For each participant, confidence intervals (CIs) were calculated around F1 and F2 values for each vowel within the reference conditions (SR1pre, SR2pre, and SR1post). This facilitated an evaluation of change across hearing aid conditions by creating a criterion range of values around the baseline. Outside of this range, changes were deemed significant at the 10% significance level, two-tailed. For each comparison, CIs were computed from the mean F1 and F2 production for each vowel of the reference condition, which is $\pm$ 1.645 times the standard deviations of each vowel production (Portney & Watkins, 2000). The CIs were used as a criterion to demonstrate if (a) the mean score of the comparison condition fell within the CI or (b) the mean score of the comparison condition fell above or below the CI. A significant change between hearing aid conditions was observed when the comparison condition fell outside the CIs of the reference condition.

Change in vowel space area. A permutation test was employed to evaluate whether there were statistically significant changes in participants' vowel space areas between different hearing aid conditions. Vowel space area was measured as the area enclosed by the irregular polygon formed by the F1 and F2 values of the vowels /i, ɪ, e, ɛ, æ, ɔ, u/. There were generally three tokens of each vowel spoken by participants, although there were instances with as few as two and as many as five. By participant, these tokens were used to calculate pooled variances for production in F1 and F2. It was deemed there were too few tokens to calculate meaningful variance by vowel for each participant. The pooled standard deviations for F1 and F2 were then used to simulate 100 000 vowel spaces for each participant using normally distributed variation in F1 and F2. These simulations created probability distributions for vowel space area for each reference condition. When comparing pre versus post hearing aid conditions, the simulation and probability distributions were created using tokens from the pre condition. The position of the post condition's actual vowel space area in the pre condition's area distribution was determined. The post area was deemed statistically different from the pre area distribution if it fell below the lower 5th percentile or above the upper 95th percentile. When comparing SR1post to SR2post conditions, the simulations were conducted using SR1post vowel space tokens.

Change in vowel space shape. Changes in the configuration of vowels in the F1 by F2 vowel space were quantified using a Procrustes analysis (Gower, 1975) similar to Mitsuya *et al.* (2015). A dissimilarity measure, d-index, quantified the changes in vowel space shape between hearing aid conditions. The d-index represents the sum of squared errors between two vowel spaces after a best fit linear transformation has been applied to align corresponding vowels in the two vowel spaces. The simulations from the vowel space area analysis also created probability distributions for d-index for each of the reference conditions. The post condition was deemed statistically different from the pre condition if the d-index calculated using the actual pre and post vowel spaces was above the 90th percentile of the distribution of d-index obtained using the pre condition vowel spaces generated randomly. When comparing SR1post to SR2post conditions, the simulations were conducted using SR1post vowel space tokens.

6.3.1.1 Normative vowel space values

Case Studies 1-3 were children between the ages of 9-12 years from the London, Ontario area. References for typical vowel productions in 7 children with normal-hearing between the ages of 9-12 years were provided by the Child Amplification Laboratory at Western University. The averages and standard deviations for F1 and F2 of /i/, /u/ and /ɔ/ were provided. The upper limits of the 90% CIs were calculated for /i/ and /ɔ/ and the lower limit of the 90% CI for /u/ was calculated from the averages and standard deviations. These criteria were used to determine the limits of normal vowel productions for adolescents with normal hearing.

Case Study 4 was an adult male. References for typical vowel productions in 8 young adult males with normal hearing were provided by Chapter Three of the thesis. The averages and standard deviations for F1 and F2 of /i, ɪ, e, ɛ, æ, ɔ, u/ were used to calculate 90% CI. The upper limits of the 90% CIs for /i, ɪ, e, ɛ, æ, ɔ/ and the lower limit of the 90% CI for /u/ were used to determine the limits of normal vowel productions for young adult males with normal hearing.

6.3.1.2 Sibilant /s/ productions

References for typical /s/ productions in children with normal-hearing between the ages of 9-15 years were provided by Flipsen *et al.* (1999). This reference value was used to calculate the criterion of a normal production of /s/. The frequency mean (6480 kHz) and standard deviation (1190 kHz) provided by Flipsen *et al.* (1999) were used to calculate the criterion to determine if the participant's productions of /s/ were normal. The criterion was determined to be the lower limit of the 90% CI calculated from the frequency mean and standard deviation. If the participant's /s/ production was above 4522 kHz, it was determined to be a normal production of /s/.

Reference values for typical /s/ productions in young adult males were provided by Haley *et al.* (2010). The frequency mean was 6200 Hz and standard deviation was 600 Hz. These values were used to calculate the criterion for a normal production of /s/. If the /s/ production from Case Study 4 was above 4026 Hz, it was determined to be a normal production of /s/.

6.3.2 Case Study 1

Figure 37 shows the vowel productions for Case Study 1 at pre and post hearing aid conditions for SR1 and SR2. Compared to the normative vowel space, the vowel spaces for Case Study 1 are larger. The F1 dimension of Case Study 1 has a higher F1 limit than the normative vowel space. The F2 dimension of Case Study 1 has higher and lower F2 limits than the normative vowel space. Figure 38 shows the number of normal /s/ productions out of 10 utterances at pre and post conditions for SR1 and SR2.

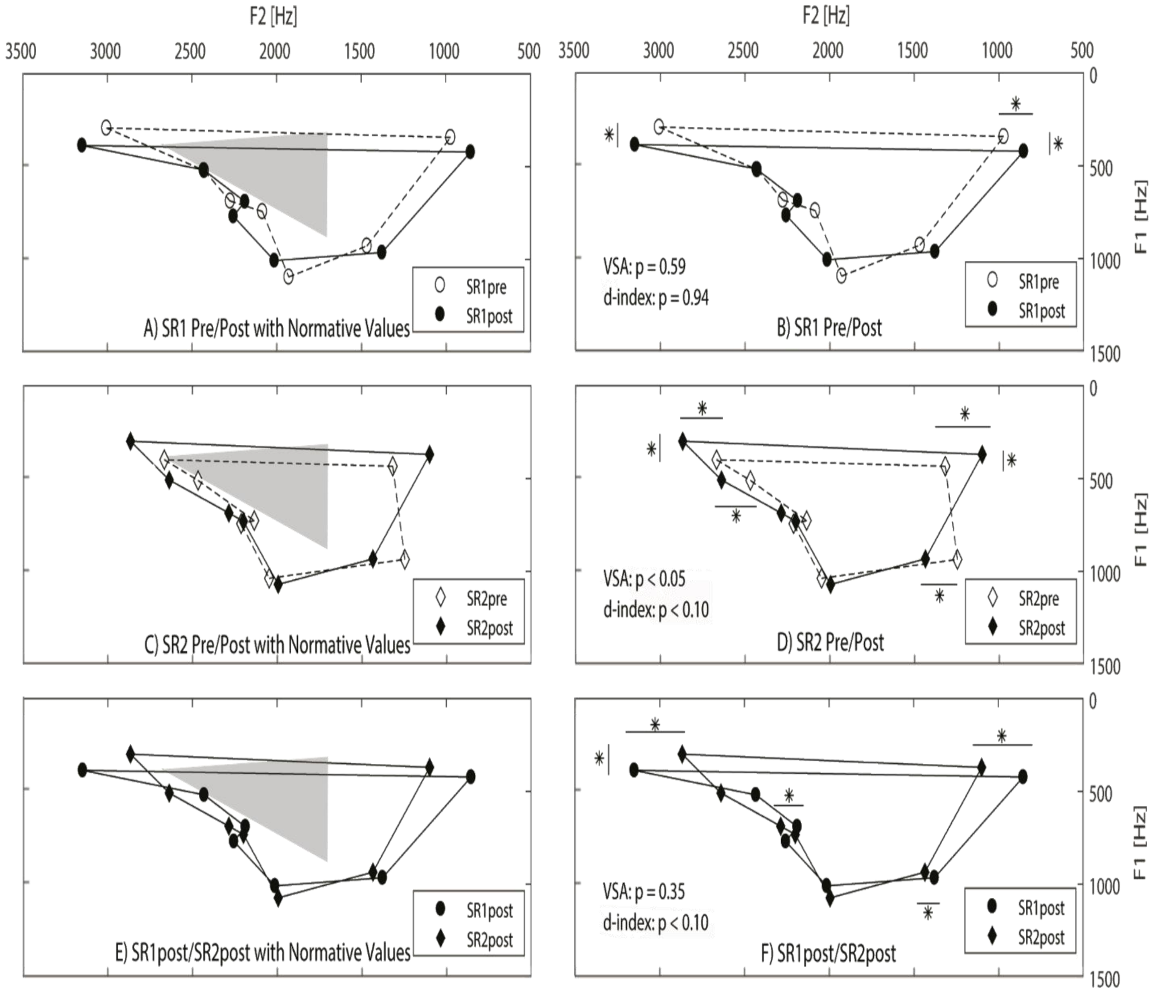

Figure 37. F1/F2 vowel plots for SoundRecover1 and SoundRecover2 at pre and post conditions for Case Study 1. The shaded regions are normative vowel productions for /i/, /u/ and /ɔ/ collected by the Child Amplification Laboratory at Western University. VSA indicates statistical difference between vowel space areas for the two conditions. Dissimilar (d)-index indicates statistical difference between vowel space shapes for the two conditions. * indicates significant difference (*p* ≤ 0.05).

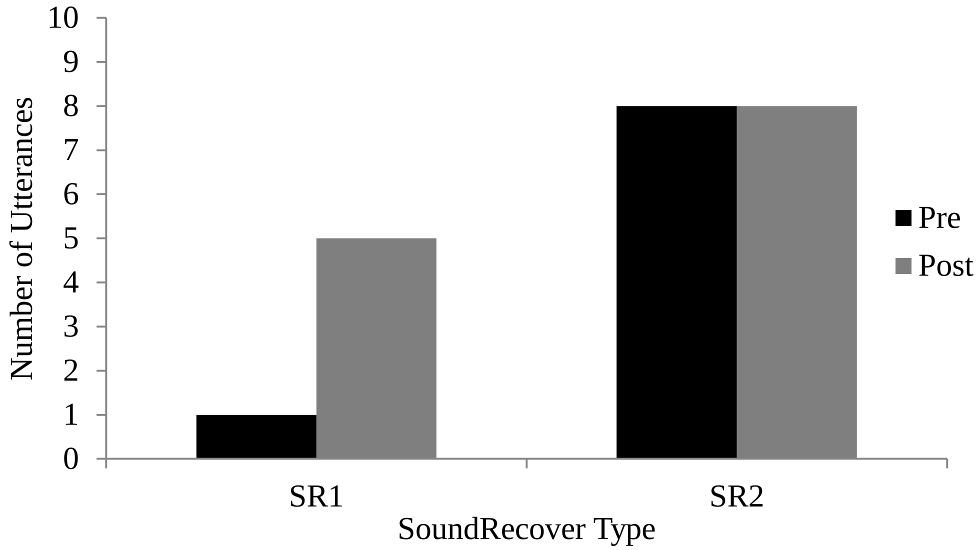

Figure 38. Number of normal /s/ productions at pre and post SoundRecover1 and SoundRecover2 conditions for Case Study 1.

Comparison between SR1pre and SR1post conditions. The vowel spaces have similar vowel space areas and vowel space shape. The vowel spaces had significantly different corner vowel productions for /i/ and /u/. The F1 dimension decreased with acclimatization to SR1 as F1 values increased in values for /i/ and /u/. There was an increase of the F2 dimension at SR1post as /u/ had a lower F2 value than SR1pre. There was an increase in the number of /s/ productions that fell within the normal range after acclimatization to SR1.

Comparison between SR2pre and SR2post conditions. SR2post was significantly larger in vowel space area than SR2pre. The shape of the vowel space of SR2post was significantly different than at the SR2pre timepoint. The two vowel spaces differed in the corner vowels /i, ɔ, u/ and /e/. The F1 dimension increased after acclimatization with SR2 as /i/ and /u/ lowered in F1 value compared to SR2pre. The F2 dimension also increased at SR2post as /i/ increased in F2 value and /u/ decreased in F2 value compared to SR2pre. The number of /s/ productions in the normal range did not differ between SR2pre and SR2post.

Comparison between SR1post and SR2post conditions. The vowels spaces had similar vowel space areas, however, the vowel space shape of SR2post was signifcantly different than SR1post. The two vowel spaces differed in the corner vowels /i, ɔ, u/ and /ɛ/ for F2 values. The back vowels /ɔ/ and /u/ had higher F2 values at SR2post than SR1post. Whereas, /i/ had a lower F2 value at SR2post than SR1post. This shows that the F2 dimension was smaller with SR2post than SR1post.

6.3.3 Case Study 2

Figure 39 shows the vowel productions for Case Study 2 at pre and post hearing aid conditions for SR1 and SR2. Compared to the normative vowel space, the vowel spaces for Case Study 2 were larger for all hearing aid conditions. The F1 dimension for Case Study 2 is approximiately similar to the F1 diemension of the normative vowel space. The F2 dimension of Case Study 2 has higher and lower F2 limits than the normative vowel space. Figure 40 shows the number of normal /s/ productions out of 10 utterances at pre and post conditions for SR1 and SR2.

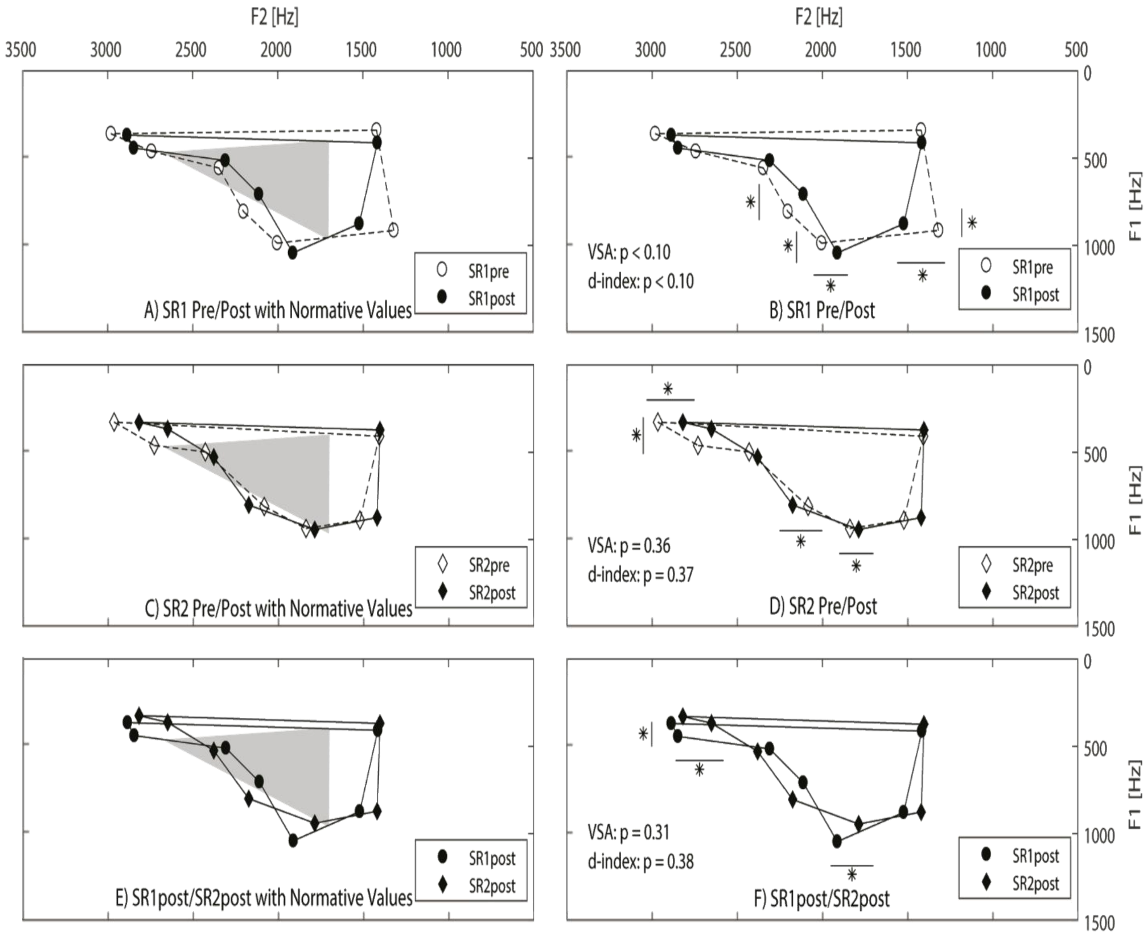

Figure 39. F1/F2 vowel plots for SoundRecover1 and SoundRecover2 at pre and post conditions for Case Study 2. The shaded regions are normative vowel productions for /i/, /u/ and /ɔ/ collected by the Child Amplification Laboratory at Western University. VSA indicates statistical difference between vowel space areas for the two conditions. Dissimilar (d)-index indicates statistical difference between vowel space shapes for the two conditions. * indicates significant difference ($p \leq 0.05$).

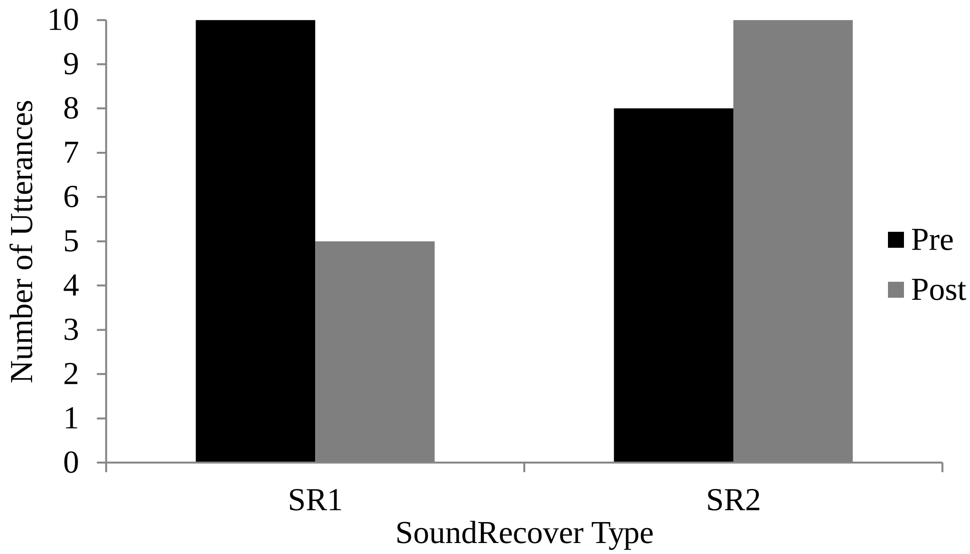

Figure 40. Number of normal /s/ productions at pre and post SoundRecover1 and SoundRecover2 conditions for Case Study 2.

Comparison between SR1pre and SR1post conditions. After acclimatization with SR1, the vowel space area significantly decreased in size and the shape of the vowel space significantly changed. The F2 dimension decreased in size after acclimatization to SR1 as /ɔ/ decreased in F2 value compared to SR1pre. Other vowels that were significantly different between SR1pre and SR1post were /ɛ/ and /æ/. The F1 values of /ɛ/ and /ɔ/ decreased after acclimatization to SR1. In contrast, /æ/ increased in F1 value at SR1post. There was a decrease in the number of /s/ that fell within the normal range after acclimatization to SR1.

Comparison between SR2pre and SR2post conditions. The vowel spaces have similar vowel space areas and vowel space shape. Vowels that were significantly different between SR2pre and SR2post were /i, e, ɛ and æ/. The F2 dimension decreased in size after acclimatization to SR2 as the F2 value of /i/ decreased. The F2 value of /æ/ also decreased in value after the acclimatization period, whereas, /ɛ/ increased in F2 value. The F1 value of /e/ was lower at SR2post than at SR2pre. There was an increase the number of /s/ productions that fell within the normal range after acclimatization to SR2.

Comparison between SR1post and SR2post conditions. The vowel spaces have similar vowel space areas and vowel space shape. The vowels that were significantly different between SR1post and SR2post were /e/ and /æ/. The F1 and F2 values of /e/ were lower at SR2post than SR1post. The F2 value of /æ/ was lower at SR2post than SR1post.

6.3.4 Case Study 3

Figure 41 shows the vowel productions for Case Study 2 at pre and post hearing aid conditions for SR1 and SR2. Compared to the normative vowel space, the vowel spaces for Case Study 3 were larger, except for the SR2pre condition. The SR2pre condition had similar F1 and F2 dimensions as the normative vowel space. The upper limit of the F1 dimension for the SR2post condition is similar to the upper limit of the normative vowel space, however, the lower limit of the F1 dimension is lower than the lower limit of the normative vowel space. Case study 3 had poor /s/ productions in all testing conditions.

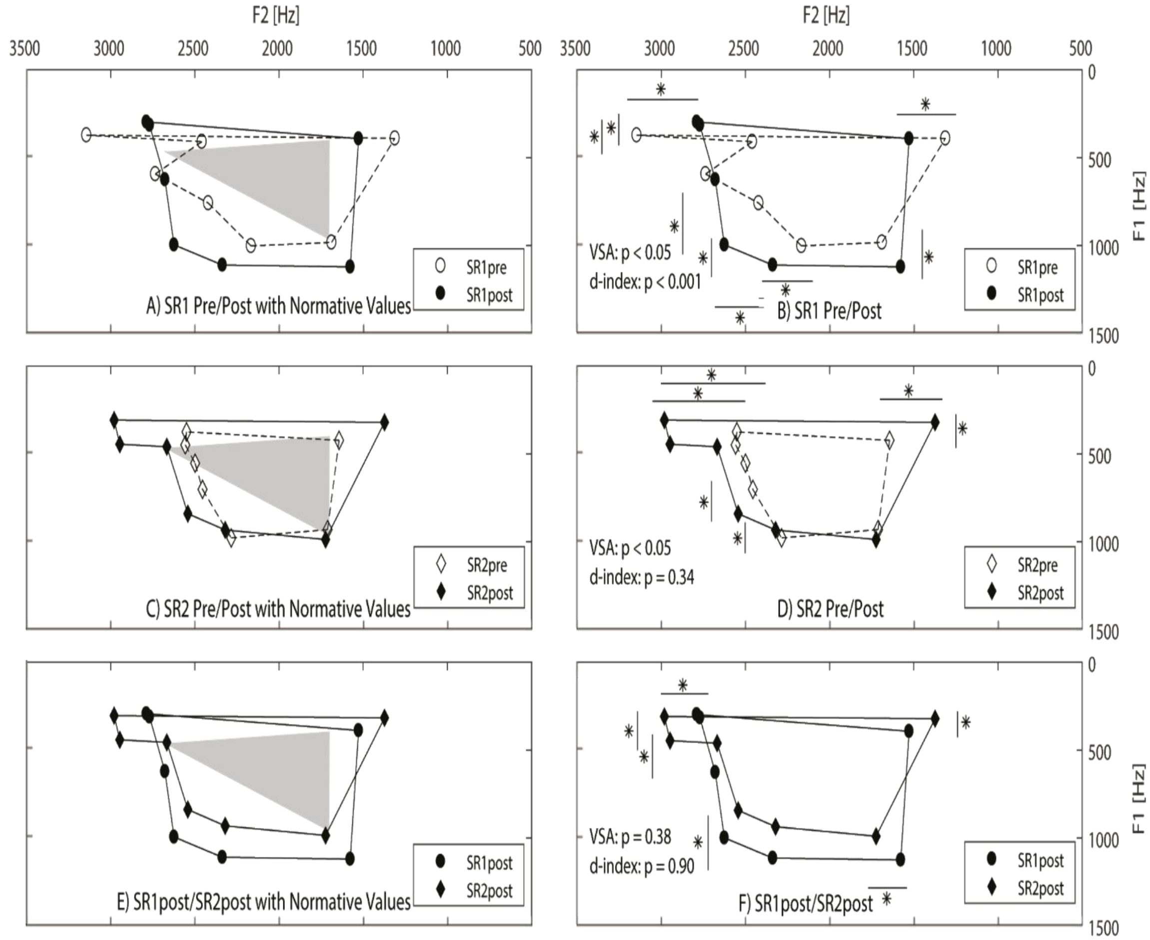

Figure 41. F1/F2 vowel plots for SoundRecover1 and SoundRecover2 at pre and post conditions for Case Study 3. The shaded regions are normative vowel productions for /i/, /u/ and /ɔ/ collected by the Child Amplification Laboratory at Western University. VSA indicates statistical difference between vowel space areas for the two conditions. Dissimilar (d)-index indicates statistical difference between vowel space shapes for the two conditions. * indicates significant difference ($p \leq 0.05$).

Comparison between SR1pre and SR1post conditions. The vowel space area increased in size after acclimatization to SR1. The vowel space shapes between SR1pre and SR1post were significantly different from each other. The F2 dimenson decreased in size after acclimatization to SR1 as the F2 value for /i/ decreased and the F2 value for /u/ increased. The F1 dimension increased in size at SR1post as the F1 values for /æ/ and /ɔ/ increased and the F1 value for /i/ decreased. Other vowels that changed in productions after acclimatization to SR1 were /e/ and /ɛ/. The F1 value of /e/ was lower at SR1post. The F1 and F2 values of /ɛ/ increased after acclimatizaton to SR1.

Comparison between SR2pre and SR2post conditions. The vowel space area significantly increased in size after acclimatization to SR2. The vowel space spaces between SR2pre and SR2post were similar to each other. The F2 dimension decreased after acclimatizaton to SR2 as the F2 value of /i/ decreased and F2 value of /u/ increased. The F1 values of /u/ and /æ/ significantly decreased at SR2post compared to SR2pre. Other values that significantly changed in productions after acclimatization to SR2 were /e/ and /ɛ/. The F1 value of /ɛ/ and F2 value of /e/ were higher at SR2post.

Comparison between SR1post and SR2post conditions. The vowel space areas were similar between SR1post and SR2post. The vowel space shapes were signifincantly different between SR1post and SR2post. The F2 dimension was larger in SR2post than in SR1post as the F2 value of /i/ was signficantly higher in SR2post. The F1 dimension was smaller in SR2post than in SR1post as the F1 value of /æ/ was smaller in SR2post. Other vowels that were significantly different between SR1post and SR2post were /ɪ, e, and ɔ/. The F1 value of /ɪ/ was lower at SR2post than at SR1post. The F1 and F2 values of /e/ were higher at SR2post. The F2 value of /ɔ/ was higher at SR2post.

6.3.5 Case Study 4

Figure 42 shows the vowel productions for Case Study 4 at pre and post hearing aid conditions for SR1 and SR2. Compared to the normative vowel space of young adult males collected from Chapter Three, the vowel spaces for Case Study 4 were within the F2 dimension of the normative data. The vowel spaces for SR2 were similar to the normative vowel space, except that front, close vowels (/i/ and /e/) have lower F1s than the normative vowel space. The vowel space for SR1post has a higher F1 limit than the normative vowel space. The front, close vowels (/i/ and /e/) for SR1pre and SR1post also have lower F1s than the normative vowel space. Figure 43 shows the number of normal /s/ productions out of 10 utterances at pre and post conditions for SR1 and SR2.

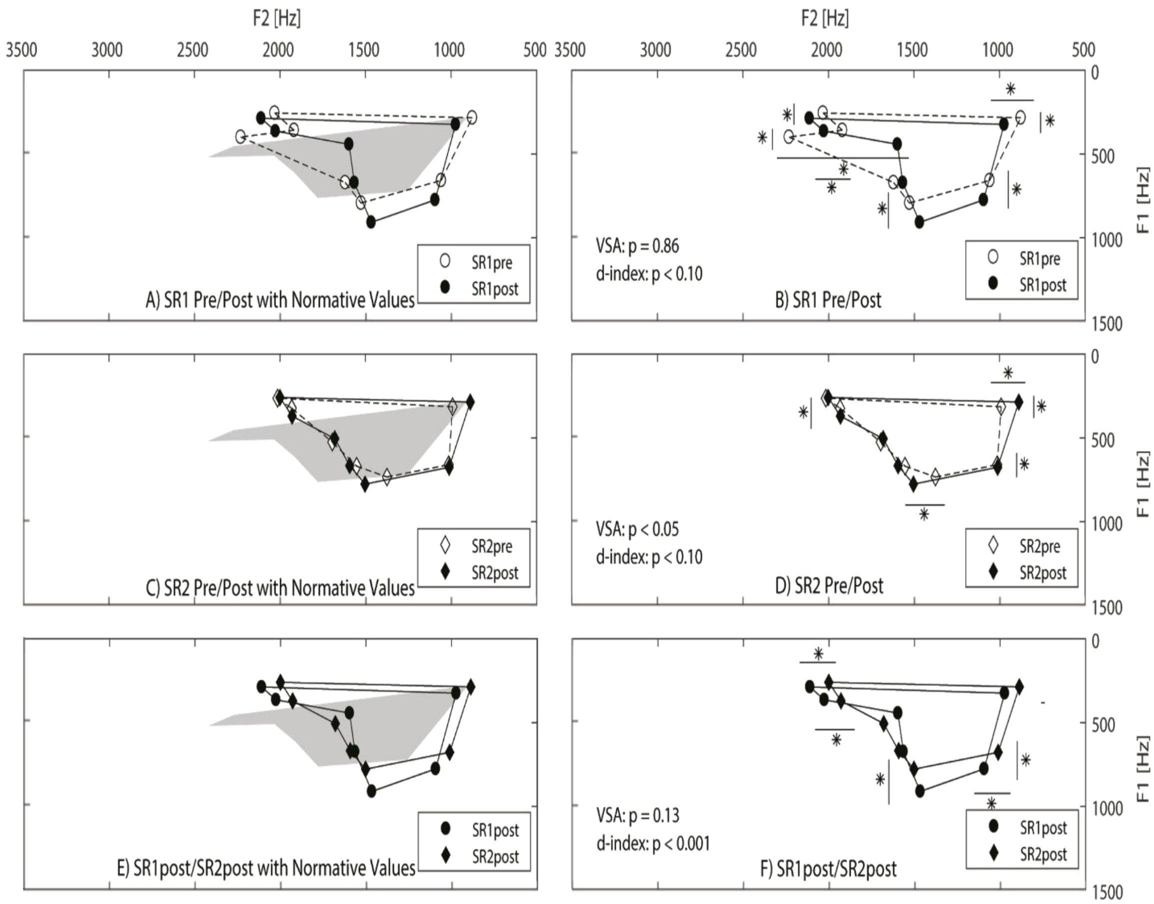

Figure 42. F1/F2 vowel plots for SoundRecover1 and SoundRecover2 at pre and post conditions for Case Study 4. The shaded regions are from vowel productions from young adult males collected from Chapter Three. VSA indicates statistical difference between vowel space areas for the two conditions. Dissimilar (d)-index indicates statistical difference between vowel space shapes for the two conditions. * indicates significant difference ($p \leq 0.05$).

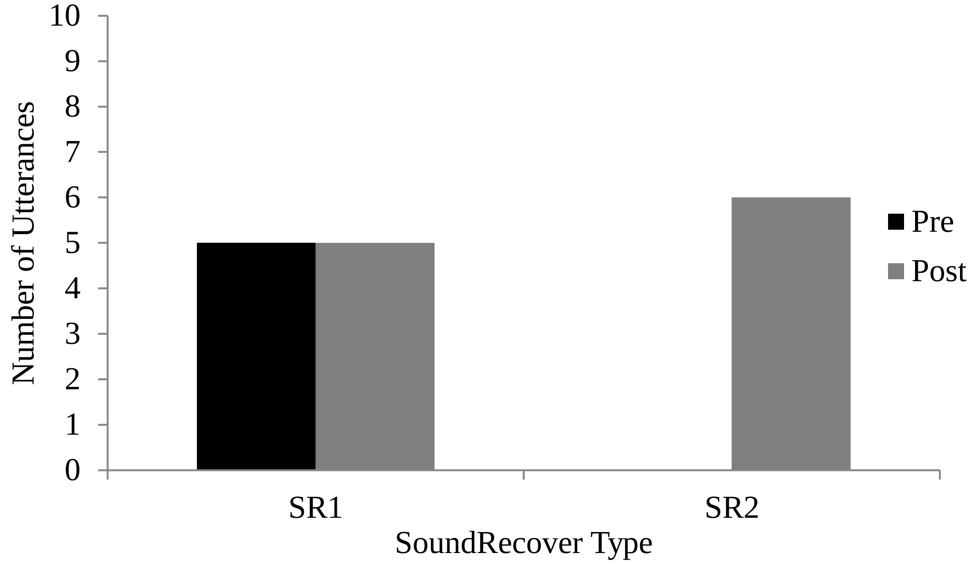

Figure 43. Number of normal /s/ productions at pre and post SoundRecover1 and SoundRecover2 conditions for Case Study 4.

Comparisons between SR1pre and SR1post conditions. The vowel space areas were similar between SR1pre and SR1post. The shape of the vowel space significanty changed after acclimatization to SR1. The F2 dimensions of SR1post is simaller than SR1pre as the F2 value of /u/ is higher in SR1post. The F1 dimension increased in size after acclimatization to SR1 as the F1 values of /æ/ and /ɔ/ were higher at SR1post. The F2 values of /e/ was higher and the F2 value of /ɪ/ was lower at SR1post. The F1 values of /i, ɪ, and u/ were higher at SR1post than at SR1pre. The number of /s/ productions within the normal range remained the same between SR1pre and SR1post.

Comparisons between SR2pre and SR2post conditions. After acclimatization to SR2, the vowel space area significantly increased in area and changed in shape. The F2 dimension increased in size after acclimatization to SR2, as the F2 value of /u/ decreased. The F2 value of /æ/ was higher at SR2post than at SR2pre. The F1 values of /e/ and /ɔ/ were higher after acclimatization to SR2. The F1 of /u/ was lower at SR2post than at SR2pre. The number of /s/ productions within the normal range increased after acclimatization to SR2.

Comparisons between SR1post and SR2post conditions. The vowel space areas of SR1post and SR2post were similar to each other, however, they differ significantly in shape. The F2 dimension of SR2post was smaller than SR1post as the F2 values of /i/ and /e/ were lower at SR2post. The F2 value of /ɔ/ was lower and the F2 value of /ɛ/ was higher at SR2post than at SR1post. The F1 dimension of SR2post is smaller than SR1post as the F1 values of /æ/ and /ɔ/ were lower at SR2post. The F1 and F2 values of /ɪ/ were higher at SR2post than at SR1post.

6.4 Discussion

A series of case studies were presented in the current study that evaluated the changes in vowel and sibilant /s/ productions after acclimatization to non-adaptive (SR1) and adaptive (SR2) NLFC. Changes in speech production with SR1 and SR2 varied across participants. In general, each participant had significant vowel changes in their speech after acclimatization to the processor and the two types of NLFC resulted in different vowel changes. All but one case study (Case Study 3) showed changes in sibilant /s/ productions after acclimatization to the processors.

6.4.1 Acclimatization effects

All case studies had approximately six weeks of acclimatization to SR1 and SR2. The number of normal /s/ productions changed before and after acclimatization for most of the case studies. As well, significant changes in vowel space area, vowel space shape, and/or formant values between pre- and post-conditions were observed for participants. The vowels that were mainly affected by acclimatization for SR1 were /i/, /æ/, /ɔ/ and /u/ and for SR2 were /i/, /e/, /æ/ and /u/. These vowels are the corner vowels of the vowel space and changes in their formant values would result in changes in vowel space area and vowel space shape. When compared to the normative vowel spaces, even after acclimatization, the vowel spaces of the case studies were still different from normative values. This is consistent with results by Ozbic and Kogovsek (2010) where vowel productions of children with severe and profound hearing losses were different from children with normal hearing. As well, more changes were seen with corner vowels in the case studies as children with severe to profound hearing losses usually produces central vowels more correctly (Smith, 1975).

Acclimatization effects have also been found in other studies that have evaluated the effectiveness of NLFC. Wolfe *et al.* (2017) evaluated audibility and speech recognition abilities in children with high frequency hearing loss using non-adaptive and adaptive NLFC in Phonak hearing aids. Their results found after four to six weeks of acclimatization to adaptive NLFC, the children had better plural detection and word recognition scores than with non-adaptive NLFC. Similar acclimatization effects have been in Glista *et al.* (2009, 2012) and Simpson *et al.* (2005). However, Wolfe *et al.* (2017) have recommended that more changes in speech perception measures may occur with a longer acclimatization time of more than 4-6 weeks. For example, Wolfe *et al.* (2010, 2011) have shown that audibility and recognition of high frequency speech sounds increased with NLFC use after 4-6 weeks. When they examined after six months of acclimatization with NLFC, recognition of speech sounds in quiet improved.

6.4.2 Comparisons to normative vowel space

The case studies presented had different vowel spaces compared to the normative vowel space. This was expected as vowel spaces for individuals with severe to profound hearing losses are different individuals with normal hearing (Ozbic & Kogovsek, 2010; Plant, 1984; Smith, 1975). The vowel spaces of the case studies were mostly larger in shape compared to the normative vowel spaces. This was unexpected as previous studies have reported vowel spaces of individuals with severe to profound hearing losses were more centralized and F1 and F2 formant ranges were reduced compared to individuals with normal hearing (Ozbic & Kogovsek, 2010; Smith 1975). As well, other studies have reported smaller vowel space shapes and/or areas due to decreased spectral contrast distances among vowels compared to individuals with normal hearing (Cowie & Douglas-Cowie, 1983; Lane & Webster, 1991; Plant, 1984). The current study collected the vowels in /hVd/ context, whereas, other articles collected vowels differently. Plant (1984) collected vowels during spontaneous speech and Ozbic and Kogovsek (2010) collected vowels at the initial, medial, and final positions of target words. These differences in vowel collection may have resulted in differences in vowel data.

In further detail, the F1 ranges for the case studies were closer to the normative F1 range. The lower limit of the F1 dimensions for most of the case studies were approximately within the normative lower limit range. However, the upper limits of the F1 dimensions for the case studies were usually higher in frequency than the normative F1 upper limit. Case Studies 2 and 4 had F1 dimensions that were approximately within the normative F1 dimensions. In contrast, the F2 ranges for the case studies were larger than the F2 dimension in the normative vowel space. This was expected that individuals with hearing loss would have more F2 differences than F1 differences compared to normative vowel space. Ozbic and Kogovsek (2010) reported that children with hearing loss mostly differed from the children with normal hearing in the F2 ranges.

The vowel spaces for Case Study 4 were closer to the normative vowel space than the other case studies. This could be due to the age difference between Case Study 4 and the other case studies. Case Study 4 was a young adult male and Case Studies 1-3 were children between 9-12 years of age. Huber *et al.* (1999) reported larger standard deviations for formant productions in children compared to adults. As well, Peterson and Barney (1952) and Hillenbrand *et al.* (1995) had listeners judge vowel productions of children and adults. The listeners found that the children's vowel productions were more variable compared to adult's vowel productions.

6.4.3 Differences between SoundRecover1 and SoundRecover2

6.4.3.1 Vowel productions

The comparison between SR1 and SR2 found that the two processors mainly differed in vowel space shape. The differences in vowel space shape could have been due to the significant differences in F1 and F2 values between SR1 and SR2. Three out of the four cases (Case Studies 1, 2 and 4) had more F2 differences than F1. As well, most of the significant differences between SR1 and SR2 occurred for the front vowels /i/, /e/ and /æ/ that are in the corners of the vowel space. These front vowels tend to have higher F2s compared to other vowels. The results suggest that SR1 and SR2 differs in their processing of sounds, specifically in the high frequency regions where F2 may be located.

These significant differences in vowels between SR1 and SR2 are consistent with the differences in the processing of auditory stimuli by the two processors. SR1 has one cut-off frequency for all auditory stimuli (Glista *et al.*, 2016c). In contrast, SR2 has two cut-off frequencies: upper and lower cut-off frequencies (Glista *et al.*, 2016c). The upper cut-off frequency is used when the auditory stimuli is low frequency dominant and the lower cut-off frequency is used when the auditory stimuli is high frequency dominant. SR2 should be able to preserve the low frequency regions better than SR1. Wolfe *et al.* (2017) presented spectrograms for “my name is asa” that were processed by non-adaptive and adaptive NLFC. The results showed that the adaptive NLFC preserved formant structures and formant ratios better than SR1. There was less compression used in the higher frequency bands with adaptive NLFC than non-adaptive NLFC. As well, Glista *et al.* (2016c) showed the changes in formant peaks of /i/ when SoundRecover was disabled and when SR1 and SR2 were enabled. The upper peaks of /i/ remained in the same location when SR2 was enabled as SoundRecover off, whereas, SR1 lowered the peak by 1000 Hz. These differences in spectral information of vowels between SR1 and SR2 may have resulted in the vowel production differences.

6.4.3.2 Sibilant /s/ productions

The comparison between SR1 and SR2 found that the two processors differed in /s/ productions. Three out of the four case studies (Case Studies 1, 2, and 4) after acclimatization to the SoundRecover processors had more normal /s/ production with SR2 than SR1. A possible reason that this may have occurred is differences in compression ratio used by the two processors. SR2 used a lower compression ratio than SR1. A lower compression ratio causes less compression to the high frequency bands of speech. This reduces sound distortions and creates sounds that may have a more natural bandwidth. This is demonstrated by the spectrograms presented in Wolf *et al.* (2017), where SR2 has less compression in the higher frequencies than SR1. Studies by Ellis and Munro (2013) and Souza *et al.* (2013) have showed decrease in sentence recognition as compression ratio increased with non-adaptive NLFC. Similarly, Alexander (2016) showed reduced vowel and consonant recognition as compression ratio increased with non-adaptive NLFC. The results from the current study suggests that SR2 may result in more normal productions of /s/ as it uses lower compression ratios than SR1.

6.4.4 Limitations and Future Research

The present study is a preliminary analysis of changes in speech productions after acclimatization to SR1 and SR2 in four case studies. A limitation of this current study is the small sample size. Future research should continue to examine speech production changes but with a larger sample size. As well, performance without the use of NLFC was not evaluated in the current study. Thus, conclusions regarding benefits of using SR1 or SR2 over no NLFC use could not be determined. Studies by Glista *et al.* (2009), Brennan *et al.* (2014), and McCreery *et al.* (2014) have shown individuals with hearing losses perform similarly or better with non-adaptive NLFC than without NLFC.

Another limitation in the present study is the limited normative data that was used. The normative data that was used for Case Studies 1-3 were from 7 children that were similar in ages as the case studies and from the same region of Southwestern Ontario. However, the normative data were limited to three vowels and were not collected in similar contexts as the current data. As well, the normative data used for Case Study 4 was from a sample of 8 young adult males from Southwestern Ontario. The normative data for /s/ productions were provided by other articles and were not collected within the research lab. Thus, appropriate normative data is needed for better comparisons between individuals with normal hearing and hearing loss.

In conclusion, the present study evaluated a new NLFC strategy that is adaptive, called SR2 in Phonak hearing aids. SR2 uses two cut-off frequencies that allows for different processing of high and low frequency stimuli and uses lower compression ratios compared to SR1. The present study showed there are differences in speech productions between SR1 and SR2 that may be due to the differences in processing strategies. Specifically, production changes in the high frequency bands of speech such as F2 in vowels and spectral means of /s/ are affected. The results suggest there is a benefit to using SR2 as there are more normal productions of /s/ after 6 weeks of acclimatization compared to SR1. There are limitations in the current study, such as small sample size and lack of normative data, that may limit the interpretations of the results. Further research is needed to examine the differences between non-adaptive and adaptive NLFC and the benefits it has on speech perception and production.

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Chapter 7

7 Discussion

7.1 Research Aims

This dissertation involved the manipulation of auditory feedback to evaluate how individuals with normal hearing and hearing loss change their speech production. Chapter 2 examined the sound pressure level needed to elicit maximum magnitudes of compensations for altered auditory feedback studies. Chapter 3 examined formant compensation differences in individuals with normal hearing and hearing aid users. In Chapter 4, the same groups of participants also participated in a study that manipulated intensity using a paradigm that was similar to the formant manipulation study in Chapter 3. Chapters 5 and 6 manipulated auditory feedback using non-adaptive and adaptive non-linear frequency compression (NLFC) in hearing aids. Chapter 5 examined the differences in vowel and fricative productions for various non-adaptive and adaptive NLFC settings in individuals with normal hearing and hearing aid users. In Chapter 6, case study analyses were used to measure changes in vowel and fricative production after six weeks of acclimatization to non-adaptive and adaptive NFLC.

7.2 Summary of findings

Auditory feedback includes auditory information from air and bone conduction pathways. When manipulating auditory feedback by perturbing acoustic information, processed signals transmitted via the air conduction pathway need to be sufficiently higher level than the unprocessed bone conduction signal for the error to be perceived. Chapter 2 determined that the sound pressure level presented through the headphone transducers needed to be at 80 dBA to elicit the maximum magnitudes of compensation compared to lower sound pressure levels of 50, 60 and 70 dBA.

One of the ways to perturb auditory feedback is to manipulate vowel formants (Chapter 3). Older and younger adults with normal hearing detected first (F1) and second (F2) formant perturbations and corrected for these perturbations by compensating in the opposite direction. This suggested the speech motor control system was not affected by aging. In contrast, hearing aid users had less formant compensation than talkers with normal hearing. These results suggested that auditory feedback may not play as important a role in speech error detection for hearing aid users and the hearing aid users may be using a different feedback system to detect (actual) formant errors.

Another way to perturb auditory feedback is to manipulate intensity (Chapter 4). All groups of talkers in the study: older and younger adults with normal hearing and hearing aid users, had similar patterns of compensation. This suggested that the speech motor control system, when controlling for intensity perturbations, may not be affected by hearing loss (mediated by the amplification of hearing aids) and aging effects.

NLFC in hearing aids induces changes in auditory feedback as it moves high frequency information to a lower frequency region. These changes in auditory feedback resulted in changes in speech production, specifically to speech cues that are higher in frequency, such as F2 in vowels and spectral means of /s/ (Chapter 5). As the strength of the NLFC processor increased with lower cut-off frequencies and higher compression ratios, more changes in speech production occurred. The NLFC did not affect speech cues that are lower in frequency, such as fundamental frequency (F0) and F1 in vowels. This is consistent with the design of NLFC processing, which is aimed mainly at the higher frequency bands of speech and would be expected to affect high frequencies more than lower frequencies.

Both types of NLFC, adaptive and non-adaptive NLFC, preserved low frequency information (Chapter 5). The changes in F1 for vowels were smaller compared to F2 changes for both types of NLFC. However, there are differences between non-adaptive and adaptive NLFC. Adaptive NLFC uses less compression in the higher frequency region and uses a higher cut-off frequency for low frequency stimuli compared to non-adaptive NLFC. These variations in NLFC resulted in differences in speech production, where adaptive NLFC elicited more changes in speech production than non-adaptive NLFC for high frequency stimuli (Chapter 5). Changes in vowel and /s/ productions were also found after acclimatization to the non-adaptive and adaptive NLFC in a series of case studies (Chapter 6). There are significant differences in vowel space shape, vowel space area, and formant values between use of the two types of NLFC. Most of the vowel differences occurred with the corner vowels: /i/, /æ/, /ɔ/ and /u/. As well, the two types of NLFC mostly differed in F2 values compared to F1 values. The results also showed that the hearing aid users had more normal productions of /s/ after acclimatization to adaptive NLFC than non-adaptive NLFC. This is consistent with the differences between non-adaptive and adaptive NLFC processing, in which adaptive NLFC should be able to preserve low frequency stimuli and improve high frequency audibility better than non-adaptive NLFC.

7.3 Implications

7.3.1 Speech motor control system

The speech motor control system is a complex system that regulates various aspects of speech. Chapter 3 determined that a hearing loss may impair the speech motor control system to detect formant errors in auditory feedback. In contrast, Chapter 4 determined that a hearing loss did not affect the speech motor control system's ability to correct for intensity perturbations. These results suggest that the regulation mechanisms for formant and intensity production may differ in the speech motor control system. There have been studies that showed a change in F0 may result in changes in formants (Eckey & MacDonald, 2015; MacDonald & Munhall, 2012). As well, Larson, Sun and Hain (2007) have shown that the regulation of F0 and intensity may be relatively independent from each other, however, they do interact in certain conditions. Further research is needed to understand how the mechanisms interact with each other and regulate different speech cues.

7.3.2 Speech compensation

Studies that have manipulated auditory feedback by changing vowel formants (Mitsuya *et al.*, 2015; Purcell & Munhall, 2006; Villacorta *et al.*, 2007), F0 (Burnett *et al.*, 1998; Jones & Munhall, 2000), intensity (Heinks-Maldonado & Houde, 2005; Larson *et al.*, 2007), and spectral noise of fricatives (Casserly, 2011; Shiller *et al.*, 2009) have shown that compensation to perturbations occurs in the opposite direction of the manipulation. Chapters 3 and 4 demonstrated that the speech motor control system in hearing aid users, and older and younger adults with normal hearing compensated in the opposite direction to the formant and intensity perturbations, as expected. However, when the same groups of talkers participated in Chapter 5, where changes in speech production were measured with different parameters of NLFC, and when listening through hearing aids rather than through insert earphones, the compensation pattern was different. In NLFC, perturbations of auditory feedback decrease signal frequency, such that high frequency information is lowered to a lower frequency area. It was hypothesized that changes in speech production would occur in the opposite direction of the frequency lowering. The results showed all groups of participants followed the frequency lowering response and did not compensate in the opposite direction. This difference in compensation patterns between Chapters 3, 4 and 5 showed that the auditory feedback system treated the auditory feedback from the hearing aid with NLFC and the formant/intensity manipulations differently. Further research is needed to determine why the speech motor control system followed the responses for NLFC, and some speculated reasons are discussed below.

Experimental designs may have resulted in differences in compensation patterns. The auditory feedback provided by NLFC may have interacted with other digital signal processing within the hearing aids. Non-linear signal processing in the aids may have changed the auditory cues compared to how listeners receive speech under headphones or insert earphones. For example, processing such as wide dynamic range compression includes level-dependent change of gain, in speeds that vary with attack/release times across the frequency range of processing. As well, NLFC is manipulating other speech cues due to the compression of high frequency information, such as upper formants and formant ratios. In contrast, participants wore insert earphones during the formant and intensity manipulations and the perturbations were created by computer algorithms that focused specifically on the target manipulation of interest. Furthermore, the frequency lowering caused by NLFC had larger manipulations than the formant perturbation study. At the strongest setting, NLFC lowered F2 by approximately 1000 Hz in Chapter 5. Whereas, in Chapter 3, F1 was perturbed by 200 Hz and F2 was changed at most by 700 Hz. These differences in processing may have resulted in differences in compensation. Further research is needed to examine speech production changes as a result of hearing aid processing and computer algorithms, and in real-world versus under laboratory conditions.

7.3.3 Non-linear frequency compression

Changes in speech production occurred with the use of NLFC in hearing aids. Different parameters of NLFC resulted in differences in speech production (Chapter 5). Weaker settings of NLFC with higher cut-off frequencies and lower compression ratios had smaller changes in speech production. In contrast, stronger settings of NLFC with lower cut-off frequencies and higher compression ratios had larger changes in speech production. This is consistent with differences in speech perception scores with different parameters of NLFC. Alexander (2016) examined the impact of frequency compression parameters, studying the effect of six combinations of cut-off frequencies and input bandwidth (by varying compression ratios) on vowel and consonant recognition in noise. They found that a low cut-off frequency, 1600 Hz, had reduced vowel and consonant recognition, especially as compression ratio increased. In comparison, at higher cut-off frequencies (2800 Hz and 4000 Hz), phoneme recognition was unaffected. Comparable results have also been found by Ellis and Munro (2013) and Souza *et al.* (2013). It is important when fitting NLFC on hearing aid users to minimize distortion of sounds by using higher cut-off frequencies and lower compression ratios for speech perception, and the current study adds new information that it may be important for speech production as well.

Non-adaptive and adaptive NLFC elicited different speech production. The differences in speech production between the two types of NLFC were greater at stronger settings (Chapter 5). The adaptive NLFC had greater changes in speech production at stronger settings than non-adaptive NLFC. The results also showed that the hearing aid users had more normal productions of /s/ after acclimatization to adaptive NLFC than non-adaptive NLFC (Chapter 6). This was consistent with Wolfe *et al.* (2017), who found that after 4-6 weeks of acclimatization to adaptive NLFC, children with hearing loss had better plural detection and word recognition scores than with non-adaptive NLFC. There is a general trend for adaptive NLFC to have greater benefit than non-adaptive NLFC, however, there are limited studies that have compared the two types of processors. Further research is needed to determine the differences between non-adaptive and adaptive NLFC.

7.4 Limitations and future research

7.4.1 Hearing loss, hearing aids, and acclimatization

There was large variability in the hearing aid users that participated in the studies. For example, the degree and configuration of hearing loss were varied, where some hearing aid users had mild to moderate hearing losses and others had moderate to severe hearing losses. As well, the hearing aid users were not acclimatized to the study hearing aids worn. Some of the hearing aid users had minimal experience with the hearing aid manufacturer. Studies such as Ellis and Munro (2015), Gatehouse (1993), Glista *et al.* (2012) and Wolfe *et al.* (2011, 2017) have shown that speech and perceptual tests may change over 4 to16 weeks of acclimatization to hearing aids. As well, Glista *et al.* (2009) reported greater benefits for NLFC with a greater degree of hearing loss. These differences within the hearing aid group may have masked some effects of hearing loss and hearing aids. Future studies may best increase the sample size to separate the hearing loss into various categories or have the hearing aid users acclimatize to the hearing aids before experimental testing.

7.4.2 Somatosensory feedback and feedforward systems

The only modality that was manipulated in the thesis was auditory feedback. The compensation results across all the chapters showed partial compensation, in which the magnitude of speech compensation was smaller than the magnitude of perturbation (Chapter 2 - 5). The results also showed that salience of auditory feedback varied across different vowels. Compared to /ɛ/, the vowel /i/ had smaller proportions of compensation relative to the perturbation (Chapter 3). This reflects that regulation of speech production is a complex system that uses other feedback and feedforward systems. There have been other perturbation studies that manipulated somatosensory feedback. A study by Tremblay, Shiller and Ostry (2003) showed that talkers changed the position of their jaw when their jaw was pulled forward during talking. Other somatosensory perturbation studies have shown that changes to the position of articulators when speaking will result in compensatory positional change of the articulators (Folkins & Abbs, 1975; Folkins & Zimmermann, 1982; Shaiman, 1989). Future manipulations of feedback may best incorporate auditory and somatosensory feedback to determine the relationship and interactions between the two systems.

The current body of work also showed speech compensation and production differences between individuals with hearing loss and normal hearing. This suggested that the individuals with hearing loss may rely on other feedback systems more than individuals with normal hearing. A study by Nasir and Ostry (2008) studied speech learning in cochlear implant recipients with their implants turned off by altering somatosensory feedback. They used a robotic device to change the position of the jaw while the participant said /s/-initial words. The cochlear implant users showed compensation to the sensorimotor perturbation similar to individuals with normal hearing. Further, the study by Laugesen *et al.* (2009) suggested that some hearing aid users use their sensorimotor feedback to monitor and change their speech intensity. These results suggest that individuals with hearing loss use feedforward commands or sensorimotor feedback to regulate speech production, possibly in addition to or instead of the auditory feedback system. To understand how speech production changes with a hearing loss and the impact of amplification devices, other perturbation studies with auditory feedback, somatosensory feedback and feedforward systems are needed.

7.4.3 Speech perception and sound quality measures

Another limitation of this work is the lack of sound quality and speech perception measures. Speech production is affected by speech perception. If a person has poor speech perception, such as having a hearing loss, changes in speech production may occur (Lane & Webster, 1991; Langereis *et al.*, 1997; Menard *et al.*, 2007) or there may be a delay in speech development (Moeller *et al.*, 2007; Ozbic & Kogovsek, 2010; Stelmachowicz *et al.*, 2004). Hearing aid fitting protocols were developed to minimize sound distortion so that audibility and patients' acceptance of hearing aids could be achieved (Scollie *et al.*, 2016). As well, studies have shown that individuals with normal hearing and hearing loss differ in sound quality ratings for different speech stimuli (Glista *et al.*, 2016a; Parsa *et al.*, 2013). Thus, differences in speech production between hearing aid users and individuals with normal hearing may have been affected by differences in speech perception or perceived sound quality. Future work may want to include perceptual measures to understand how speech production is regulated.

7.4.4 Comparisons with other adaptive frequency lowering technology

NLFC is one of three categories of frequency lowering technology that is available in hearing aids. Unitron, GN Resound and Siemens hearing aids also use NLFC with their frequency lowering program (Rahbar, 2017; Scollie, 2013). The other category is frequency translation that can be found in Speech Rescue in Oticon hearing aids (Angelo *et al.*, 2015), Frequency Composition™ in Bernafon hearing aids (Kuriger & Lesimple, 2012) and Starkey IQ from Starkey hearing aids (Galster *et al.*, 2011). Frequency transposition is another category of frequency lowering technology that can be found in Enhanced Audibility Extender in Widex hearing aids (Rahbar, 2017; Scollie, 2013). Some of these frequency lowering processors are also adaptive, such as Speech Rescue, Spectral IQ and Enhanced Audibility Extender. Future work may compare the different adaptive frequency lowering technologies to determine if there are differences in speech perception and production measures.

7.4.5 Inclusion of other vowel formants and phonemes

Perturbation studies in the literature manipulate select phonemes. For single utterance intensity perturbations, it has mainly been /u/ (Bauer, Mittal & Hain, 2006; Hafke, 2009, Larson *et al.*, 2007) or /ɑ/ (Heinks-Maldonado & Houde, 2005; Liu *et al.*, 2012). For fricative perturbation studies, the focus has been on /s/ (Shiller *et al.*, 2009) or /ʃ/ (Casserly, 2011). For formant perturbation studies, /ε/ has been used. The vowel /ε/ has mainly been used because it is a front mid-open vowel, in which the articulators have freedom to adjust and thus compensate for auditory feedback perturbations. In contrast, a perturbation of vowel formants for /æ/ may be more limited because the tongue is already at the bottom of the mouth. Additionally, F1 and F2 are sufficiently separated in /ε/ so that formants can be correctly estimated and manipulations of one formant would have minimal effects on the other formant. However, the study by Mitsuya *et al.* (2015) has shown that Fl compensation occurred across F1 manipulations for /i/, /ɪ/, /e/, /u/, /æ/, and /ɔ/. As well, Chapter 3 has shown that F2 manipulations of /ɪ/ and /i/ also result in speech compensation. Mitsuya *et al.* (2015) demonstrated that each vowel is regulated differently as each vowel received the same auditory feedback manipulation (a 200 Hz manipulation of F1) and each vowel had a different magnitude of compensation. Further, the current study only examined changes in production for /s/ using NLFC. However, other fricatives with high frequency information, such as /ʃ/ and /z/ can also change with NLFC. Thus, the generalization of one phoneme sound to other phonemes may have limitations (Pile *et al.*, 2007). Future work may want to examine other phonemes and speech stimuli.

The examination of other formants, such as third (F3) and fourth (F4) formants, is needed to understand how NLFC affects vowel perception and production. The current studies in the thesis were limited to F1 and F2. Fant (1960) and Stevens (1998) have shown that there is covariance between F2 and F3 for some vowels. As well, Wolfe *et al.* (2017) and Glista *et al.* (2016b) have shown that upper formants change with different NLFC parameters or type of NLFC. Future work may want to include examinations of upper formants.

7.5 Concluding statements

Manipulations in auditory feedback resulted in changes in speech production. However, these changes in speech production were proportional to the manipulation, such that the magnitude of speech changes were smaller than the magnitude of the perturbation. This suggests that other feedback and feedforward systems are also regulating speech production. A hearing loss may have an effect on the detection of speech errors, and as a result, individuals with hearing loss may rely on other feedback and feedforward systems to regulate their speech production. The impact of hearing aid use on speech production is not a well-studied area, with few investigations currently informing our understanding of the complex interplay between hearing impairment, acclimatization to impairment and/or intervention, and the multiple paths of feedback during speech production. This work provides initial insight into how hearing aids may change how sounds are perceived and changes in speech production may occur with age, hearing impairment, and the use of hearing aids and digital signal processes, such as NLFC.

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Appendix

Appendix A: Western University Health Science Ethics

Western Research

Research Ethics

Western University Health Science Research Ethics Board
HSREB Delegated Initial Approval Notice

Principal Investigator: Dr. David Purcell
Department & Institution: Health Sciences\Communication Sciences & Disorders, Western University

HSREB File Number: 106517
Study Title: Evaluation of speech behaviours in normal hearing and hearing impaired individuals
Sponsor: Natural Sciences and Engineering Research Council

HSREB Initial Approval Date: April 15, 2015
HSREB Expiry Date: April 15, 2016

Documents Approved and/or Received for Information:

Document Name	Comments	Version Date
Recruitment Items	webpage recruitment for younger adults	2015/03/06
Recruitment Items	flyers for recruitment	2015/03/06
Recruitment Items	webpage recruitment for older adults	2015/03/06
Recruitment Items	Poster: Normal hearing adults	2015/03/06
Recruitment Items	Poster: hearing impaired adults	2015/03/06
Data Collection Form/Case Report Form	Hearing Test Form	2015/03/06
Data Collection Form/Case Report Form	Speech production form	2015/03/06
Data Collection Form/Case Report Form	Case History: Normal hearing adults	2015/03/06
Data Collection Form/Case Report Form	Information for follow up sessions	2015/03/06
Data Collection Form/Case Report Form	Music History Questionnaire	2015/03/06
Data Collection Form/Case Report Form	Case History: Hearing Impaired Adults	2015/03/06
Data Collection Form/Case Report Form	Language History and Experience Questionnaire	2015/03/06
Other	Debriefing Script	2015/03/06
Letter of Information & Consent	Hearing adults	2015/03/31
Letter of Information & Consent	Controls	2015/03/31
Western University Protocol		

The Western University Health Science Research Ethics Board (HSREB) has reviewed and approved the above named study, as of the HSREB Initial Approval Date noted above.

HSREB approval for this study remains valid until the HSREB Expiry Date noted above, conditional to timely submission and acceptance of HSREB Continuing Ethics Review.

The Western University HSREB operates in compliance with the Tri-Council Policy Statement Ethical Conduct for Research Involving Humans (TCPS2), the International Conference on Harmonization of Technical Requirements for Registration of Pharmaceuticals for Human Use Guideline for Good Clinical Practice Practices (ICH E6 R1), the Ontario Personal Health Information Protection Act (PHIPA, 2004), Part 4 of the Natural Health Product Regulations, Health Canada Medical Device Regulations and Part C, Division 5, of the Food and Drug Regulations of Health Canada.

Members of the HSREB who are named as Investigators in research studies do not participate in discussions related to, nor vote on such studies when they are presented to the REB.

The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

Ethics Officer, on behalf of Dr. Marcelo Kremenchutzky, HSREB Vice Chair

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RELEVANT SHORT PAPER AND ABSTRACT PUBLICATIONS

Nguyen-Vaccarello, L.L.T.*, Mitsuya, T., Scollie, S., Purcell, D.W. (2015). *Interaction of air and bone conduction in speech production during altered auditory* *feedback*. *Canadian Audiologist*, 2(6). Retrieved from http://canadianaudiologist.ca/issue/volume-2-issue-6-2015/2016-caa-conference-podium-and-poster-abstracts/

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RELEVANT PRESENTATIONS

Nguyen-Vaccarello, L.L.T. (2016, Mar). Alternative approach in measuring hearing aid outcome: altered auditory feedback. Oral presentation at Western Graduate Research, London, ON.

Nguyen-Vaccarello, L.L.T. (2016, Feb). Probing the auditory feedback system in individuals with hearing loss. Oral presentation at Health and Rehabilitation Sciences Graduate Research Forum, London, ON.

Nguyen, L.L.T. (2015, Nov). Evaluation of speech compensation behaviours to speech perturbations in normal hearing and hearing impaired listeners. Oral presentation at the Research in Hearing Science Seminar Series, London, ON.

Nguyen-Vaccarello, L.L.T.*, Mitsuya, T., Scollie, S., Purcell, D.W. (2015, Oct). *Interaction of air and bone conduction in speech production during altered auditory feedback*. Poster presentation at the 18th Annual Canadian Academy of Audiology Conference, Niagara Falls, ON.

Nguyen-Vaccarello, L.L.T*., Purcell, D.W., Adams, S.G., & Scollie, S. (2015, Oct). *Speech regulation through altered auditory feedback in normal and hearing impaired adults.* Poster presentation at the 18th Annual Canadian Academy of Audiology Conference, Niagara Falls, ON.

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