

**The Role of Aided Signal-to-Noise Ratio  
in Aided Speech Perception in Noise**

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**Abstract**

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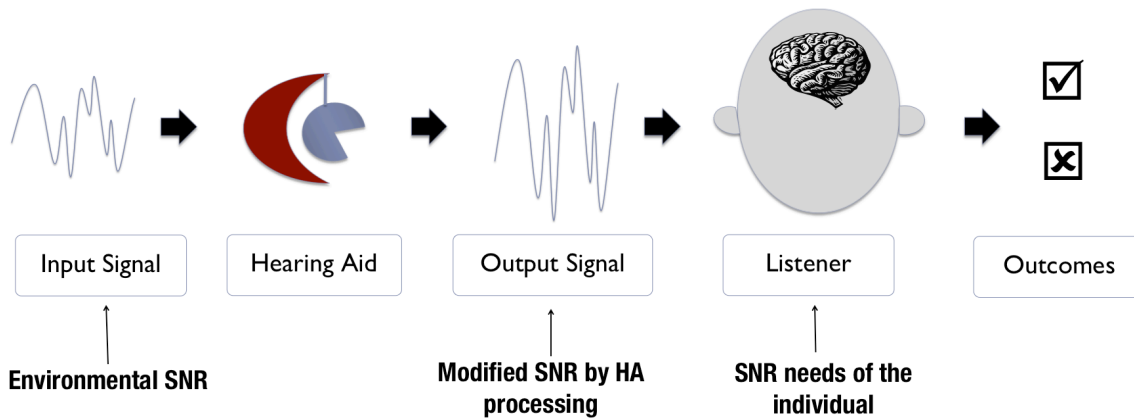
Despite advances in hearing aid technology and careful fitting processes, outcomes vary widely among hearing aid users, particularly in background noise. Predicting benefit with hearing aids is not easy to do, as many variables contributing to outcomes are not known. The general aim of this research is to explore a previously undefined variable in outcomes, the signal-to-noise ratio. The level of speech compared to the level of background noise (i.e., signal-to-noise ratio; SNR) has been linked to speech perception and quality of brain responses at the input stage of processing. However, hearing aid processing is likely to modify the SNR, therefore delivering an altered SNR version to the listener. This study first aims to quantify the amount of change in SNR made by hearing aid processing, and second, to use the SNR changes at the output of the hearing aid to predict changes in speech percetion. Two groups were studied, normal and impaired hearing listeners. Our hypothesis is that speech perception is dependent in

part on the SNR at the output of the hearing aid, since this is the signal entering the listener's auditory system. To test this hypothesis, we quantified the SNR at the output of the hearing aid using acoustic measures of the hearing aid output. We then correlated the changes in output SNR to changes in speech perception scores, measured in a background of noise. The results showed that hearing aids for both groups of listeners were changing the SNR from input to output by a small, but statistically significant, amount (mean change: -0.25 dB). However, the change in SNR was not predictive of changes in speech perception for either group. We discuss the limitations and generalizations of these results, as well as future directions for research.

# Chapter 1: Introduction

Despite advances in hearing aid signal processing over the last few decades, and careful verification using recommended clinical practices, successful use of amplification continues to vary widely. This is particularly true in background noise, where approximately 60% of hearing aid users are satisfied with their performance in noisy environments (Kochkin, 2005). Dissatisfaction can lead to undesirable consequences, such as discontinued hearing aid use, cognitive decline, and poor quality of life (Chia et al., 2007; Chisholm et al., 2007).

Many factors may contribute to aided speech understanding in noisy environments, including hearing aid centered variables (e.g., directional microphones, audibility) and patient centered variables (e.g., age, motivation, biology). Although many predictors of hearing aid outcomes are known (e.g., audibility, unaided hearing handicap, age), a large portion of the variance in outcomes remains unexplained. Several years ago, Souza and Tremblay (2006) put forth a framework for thinking about the possible sources in hearing aid variability, which was modified for the purposes of this research (Figure 1). One variable that may be of importance is the ratio between the level of speech and the level of background noise, called the signal-to-noise ratio (SNR). We will consider the SNR at various stages of the speech understanding process.



**Figure 1. Stages of potential variables involved in aided processing of speech in noise (modified from Souza and Tremblay, 2006): 1) Input or Environmental Signal, 2) HA Processing, 3) Modified Output Signal, 4) Listener Processing, and 5) HA Outcomes.**

## **2.1 Rationale for Considering the Signal-to-Noise Ratio**

There are four primary reasons supporting the notion that the SNR may be an important variable in hearing aid outcomes. The first is that the environmental, or input SNR, likely influences hearing aid outcomes. The second is that there is a strong relationship between input SNR and speech perception, which can be influenced by a number of factors. The third is that individual's have unique SNR preferences for listening to speech in noise. And finally, there is biological evidence to support behavioral data showing the importance of SNR to cortical neural responses. These reasons will be discussed in more detail in the following sections.

### **2.1.1. Environmental SNR**

The environmental, or input, SNR is likely to play a large role in aided speech perception. For example, understanding speech in a background of airplane noise is heard at an average of -2 dB SNR, an unfavorable listening environment (Pearsons et al., 1977). On the other hand, speech is often heard inside urban homes at +9 dB SNR, a much more favorable listening environment (Pearsons et al., 1977). Support for the idea that environmental SNR may influence outcomes comes from research showing that listeners are more dissatisfied and receive less benefit with their aids in noise than in quiet environments (e.g., Cox & Alexander, 1992; Kochkin, 2005; Humes, 2009). From a large-scale survey, two of the top three reasons for non-adoption of aids were that aids did not perform well in noise (48%) and/or that they picked up background sounds (45%; Kochkin, 2007). And of the people who did try aids, nearly half of them returned their aids due to lack of perceived benefit in noise or amplification of background noise. Therefore, not only is the listener's opinion about their aid important in general, the issue of how the client performs in noise with their hearing aid is of particular concern.

Humes (2009) confirmed the results by Kochkin in a study comparing hearing aid technology. He showed that the magnitude of hearing aid benefit, regardless of technology, depended on the environment the listener was asked about. On a questionnaire, the listener was asked about how helpful the aid was in different situations, grouped into four sub-scales: speech in quiet, speech in noise, speech with reduced cues (e.g. no visual cues), and a miscellaneous category. Over all technologies, the aids were least helpful in speech in noise environments, while the aids were most helpful in speech in quiet situations.

Cox and Alexander (1992) also showed differences in benefit depending on environment, including measures of speech intelligibility. Self-reported benefit was assessed using a questionnaire and objective benefit was measured using four conditions of speech intelligibility. The conditions were designed based on Walden et al. (1984) and Pearsons et al. (1977): 1) speech communication at normal conversational levels with visual cues and low noise/reverberation, 2) low external noise, but reduced speech cues due to reverberation, low intensity, or limited visual cues, and 3) high external noise, raised speech levels, and visual cues available. Speech perception tests were designed to mimic these situations, and results measured near the time of the fitting. The greatest benefit occurred with aids for condition 2 and the least benefit for condition 3. However, ten weeks post-fitting, speech testing was done again and showed the greatest benefit for condition 1 and the least for condition 3. These results suggest objective benefit may be different depending on environment, with high-level noise environments consistently showing the least benefit. At 10-weeks post-hearing aid fitting, the questionnaire results showed that the least benefit was perceived for subscales with items asking about background noise and reverberation situations, and the most benefit was perceived for familiar talkers and quiet situations. These results suggest that aids are more beneficial in “easier” listening environments, with less distorted speech cues from noise or reverberation.

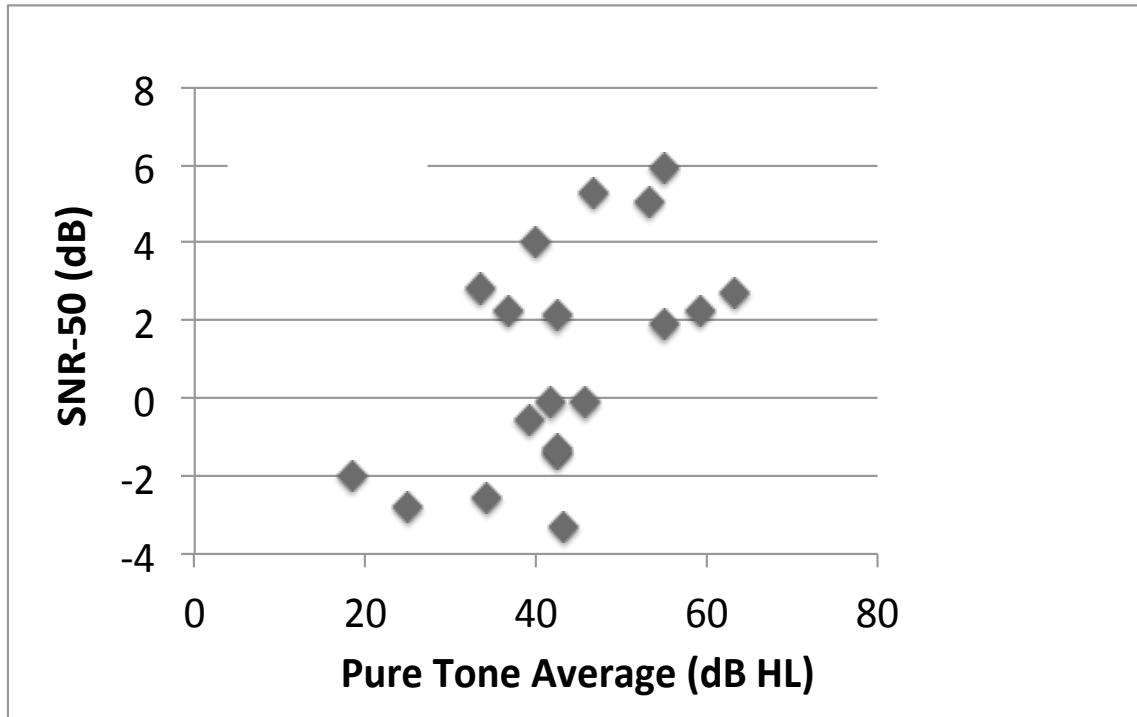
Collectively, these studies suggest that self-reported benefit and satisfaction, as well as speech perception, likely differ between listening environments, and SNR may play a role in these perceptions. In aided listening, however, the environmental SNR is likely modified by hearing aid processing. The signal received by the listener is an

altered version of the input signal, which may or may not change hearing aid outcomes. This topic is discussed in detail in later sections (2.2 and 2.3).

### **2.1.2. Environmental SNR and Speech Perception**

The **second** reason supporting the theory that SNR may be an important variable in hearing aid outcomes is that there is a strong relationship between the environmental SNR (as measured in the laboratory) and speech perception. The environmental or input SNR is usually measured as the long-term average root-mean-squared levels of speech minus those for noise. The environmental SNR is adjusted by increasing or decreasing the level of speech or noise independently, prior to presentation to the listener. Speech perception is measured by presenting a speech sound to the listener, either under headphones or in the free field, and the listener is required to repeat what he or she heard/understood. Sometimes noise is played in the background, which may be presented at a pre-determined, fixed SNR, or it can be adaptively varied until a certain criterion is met. Alternatively, the speech level can adaptively vary, while the noise level remains fixed. The goal of adaptive procedures is to find the SNR where the listener scores at a particular performance level, often at the speech reception threshold (50%), called the SNR-50. Regardless of measurement procedure, both listeners with normal hearing and hearing loss benefit from improvements in SNR (e.g., Carhart & Tillman, 1970). Predicting a listener's SNR-50 from hearing thresholds can be difficult. In Figure 2, high variability among listeners with similar hearing thresholds can be seen (unpublished data from Bentler, 2012). For example, people with an average hearing threshold (pure tone average of 500, 1000, and 2000 Hz) of 40 dB HL may only need an SNR of -3 dB to perceive 50% of speech, while

others need +4 dB SNR. This range suggests that some people have greater or worse abilities to cope with competing noise, even with similar hearing threshold levels.



**Figure 2. The relationship between pure tone average hearing thresholds and a listener’s SNR-50 (unpublished data, Bentler, 2012).**

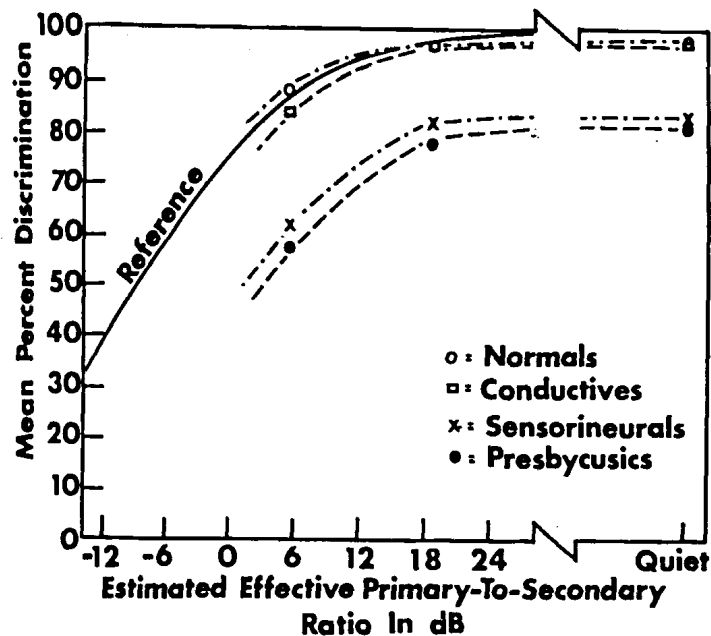
**Predicting how a person performs in background noise can be difficult, due to the variability observed between people with the same pure tone average.**

A person’s speech perception score will depend on many factors, such as the type of test materials used, location of sound sources, and presentation level of speech. The effectiveness of masking a speech sound partially depends on three acoustic characteristics of the noise, or masker: 1) the spectral shape, 2) the temporal characteristics, and 3) the intensity relative to the intensity of the speech, or the SNR (Miller, 1947). As the noise spectrum becomes more like speech, the more effective it is

at masking the target speech signal (Feston and Plomp, 1990). Temporal characteristics, such as how severely the amplitude is modulated over time, also influence the effectiveness of the masker. An un-modulated masker is one that does not change in envelope amplitude over time. Real life examples of this would be airplane or traffic noise (louder intensities), or a refrigerator or computer hum (softer intensities). If the amplitude remains constant, or un-modulated, the masker is very effective at reducing speech intelligibility. However, if the masker amplitude is modulated (e.g., competing speech, music, or restaurant noise), then normal hearing listeners are able to take advantage of the moments where the intensity is low (and the SNR is high), to capture more information about the target speech signal (Carhart, Tillman, & Greetis, 1969; Duquesnoy, 1983; Miller, 1947; Stuart, Phillips, & Green, 1995). Depending on the test stimuli, normal hearing listeners can receive a 4-10 dB improvement in speech intelligibility in a modulated background noise compared to an un-modulated noise (Miller, 1947; Carhart et al., 1969; Plomp and Mimpen, 1979; Wilson et al., 2010). On the other hand, people with hearing loss do not benefit from the amplitude modulations in a competing masker as much as normal hearing listeners (Dirks, Morgan, and Dubno, 1982; Festen & Plomp, 1990; Summers and Molis, 2004; Wilson et al., 2010). Hearing impaired listeners receive only a 0-2 dB release from masking in modulated noise (e.g., Festen & Plomp, 1990; Bronkhorst, 2000).

Finally, the SNR drives the effectiveness of the masker. Plomp & Mimpen (1979) showed that the level of the background noise explains 39% of the variance observed in 210 subjects on a speech perception task (keeping noise level fixed and varying the speech level). The speech reception threshold increases as noise level increases (across a range of 22.5 to 67.5 dBA), and the rate of change between these two variables changes

depending on the age and hearing thresholds of the subject. Carhart & Tillman (1970) found significant improvements in speech perception as SNR improved from -6 to +12 dB SNR for listeners with normal, conductive, and sensorineural loss. However, the function for listeners with sensorineural loss was shifted to the right by 11-14 dB (Figure 3). This indicates that listeners with sensorineural hearing loss may benefit from SNR improvements as much as normal or conductive loss listeners, but need overall better audibility. The needs for greater audibility in listeners with sensorineural loss has largely been attributed to elevated hearing thresholds (from the lack of cochlear gain for soft sounds) and poor frequency selectivity (from broader tuning curves; for a review, see Oxenham and Bacon, 2003).



**Figure 3. Relationship between environmental SNR (measured in a laboratory) and speech perception from Carhart & Tillman (1970). Listeners with sensorineural hearing loss need improved audibility compared to those with normal or conductive hearing loss to achieve the same speech perception score.**

Factors other than audibility that are likely to influence the relationship between SNR and speech perception are age and cognitive abilities. To determine whether aided speech intelligibility could be predicted, Humes (2002) measured 33 pre-fit variables in a group of 171 older hearing aid users with symmetrical, sensorineural hearing loss, and used a multiple regression analysis to determine which factors could predict four conditions of speech intelligibility. The pre-fit variables included 1) a measure of intelligibility, divided into 11 sub-scales, 2) a series of auditory processing tasks presented at 90 dB SPL (i.e., dichotic consonant vowel identification, pitch pattern identification, syllable sequence identification, duration discrimination, and temporal order for tones), 3) the latency of Wave V of the auditory brainstem response, 4) distortion product otoacoustic emissions in the left and right ears at low, mid, and high frequencies, 5) pure-tone averages for the left and right ears, 6) aided pure-tone average of the better ear, and 7) age of the subject. Unaided and aided speech perception was measured in four conditions: 1) a nonsense syllable identification task presented at 65 dB SPL in babble with +8 dB SNR, 2) sentence identification presented at 50 dB SPL in quiet, 3) sentence identification presented at 65 dB SPL in babble at +8 dB SNR, and 4) sentence identification presented at 80 dB SPL in babble at 0 dB SNR. To determine the predicting power of the pre-fit variables on the composite speech score, a regression analysis was performed. The regression resulted in the following contributing factors to speech intelligibility, and percentage of variance explained: hearing loss (52.3%), non-verbal IQ and aging (7.1%), verbal IQ (5.4%), and otoacoustic emissions and miscellaneous (1%). Hearing loss resulted in a negative beta coefficient, indicating that as hearing loss declined (i.e., better hearing thresholds), speech intelligibility measured at a fixed SNR improved. All other coefficients were positive, so an improvement in

those areas also meant an improvement in speech intelligibility. In summary, of the total variance in speech intelligibility scores explained by the pre-fit variables in the regression model (67.7%), Humes found that hearing loss explained almost 80% of it, with cognitive function and age as secondary factors. These results confirm those found by Gatehouse (1994) showing hearing threshold levels as the primary predictor of speech perception (up to 24.8% of the variance), followed by temporal resolution (up to 14.2% of variance), age (4.7%), personality factors (11.1%), and frequency resolution (8.2%). The results from Humes (2002) and Gatehouse (1994) suggest that once audibility has been achieved with amplification, there are still likely central contributions (such as age and cognition) to speech perception. Central factors may be moderating the relationship between SNR (either environmental or the modified SNR at the hearing aid output) and hearing aid outcomes, and we attempt to control for those factors in this study.

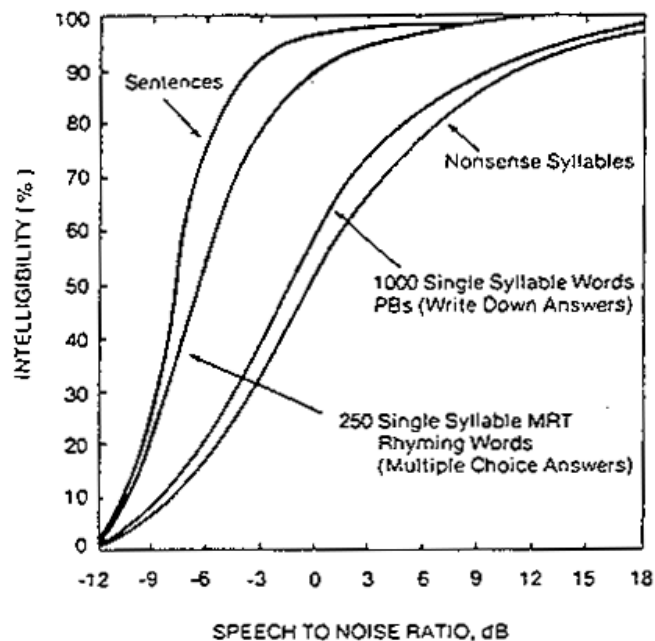
Speech perception evidence suggests that older adults have more trouble understanding speech than younger adults (for reviews, see Pichora-Fuller & Souza, 2003; Gordon-Salant, 2005). Age-related deficits in speech recognition seem to be most pronounced in adverse listening environments, such as understanding speech in competing speech situations (Miller et al., 2009). Older listeners perform worse than younger listeners on speech recognition tasks in background babble (e.g. Dubno, Dirks, & Morgan, 1984), with a single interfering talker (e.g. Duquesnoy, 1983; Frisina & Frisina, 1997; Tun, O’Kane, & Wingfield, 2002), in reverberant environments (e.g. Helfer & Wilber, 1990; Gordon-Salant & Fitzgibbons, 1999), and when speech is presented at fast rates (e.g. Fitzgibbons & Gordon-Salant, 1994; 1996; Gordon-Salant & Fitzgibbons, 1999). Some researchers have suggested that the decline in speech

intelligibility with age is from poor audibility of the speech signal due to threshold elevation (Zurek & Delhorne, 1987; Lee & Humes, 1993). However, other studies have shown speech recognition deficits, even when audibility is taken into account statistically, or through the use of customized amplification based on individual hearing thresholds (e.g. Helfer & Wilber, 1990; Pichora-Fuller, Schneider, & Daneman, 1995; Dubno et al., 2008; Rossi-Katz & Arehart, 2009). Thus, it seems that audibility alone cannot explain the age effects observed in speech recognition.

In addition to age and decreased audibility, declines in speech understanding may also be attributed to declines in temporal processing abilities (e.g., Fitzgibbons & Gordon-Salant, 1994; 1996; Gordon-Salant & Fitzgibbons, 1999). Temporal processing is the ability to extract useful information from a signal over time. When a signal is temporally degraded (e.g., rapid speech), older listeners in particular have more trouble understanding speech than younger listeners (Fitzgibbons & Gordon-Salant, 1994; 1996). A reduction in temporal processing abilities can also lead to a deficit by older adults to process fundamental frequency information (Vongpaisal & Pichora-Fuller, 2007; Arehart, Souza, Muralimanohar, & Miller, 2011; Souza, Arehart, Miller, & Muralimanohar, 2011), which is partially perceived by temporal coding of the signal. Fundamental frequency information (i.e. voice pitch) contributes to the listener's ability to separate competing talkers (e.g. Assmann & Summerfield, 1990; Arehart, King, & McLean-Mudgett, 1997). Older listeners do not benefit from fundamental frequency differences as much as younger adults do when identifying competing speech (Summers & Leek, 1998; Rossi-Katz & Arehart, 2009). Therefore, for older listeners in particular, a more favorable SNR may alleviate some of the reliance on temporal cues and cognition to fill in the missing information. More specifically, an improved SNR means that

greater audibility may be achieved, which is the primary predictor of speech perception (Humes, 2002; Gatehouse, 1994).

Finally, context can influence the relationship between SNR and speech perception. As illustrated in Figure 4 (Killion, 1985), context within the speech stimuli plays a large role in how much benefit from SNR improvements is gained, with steeper slopes for sentence material and shallower slopes for nonsense syllables. Furthermore, older listeners, with or without hearing loss, rely more heavily on contextual cues than young, normal hearing listeners (Pichora-Fuller et al., 1995; Wilson & McArdle, 2012). This could be attributed to older listeners reduced ability to benefit from temporal cues or draw from cognitive resources, as suggested previously.



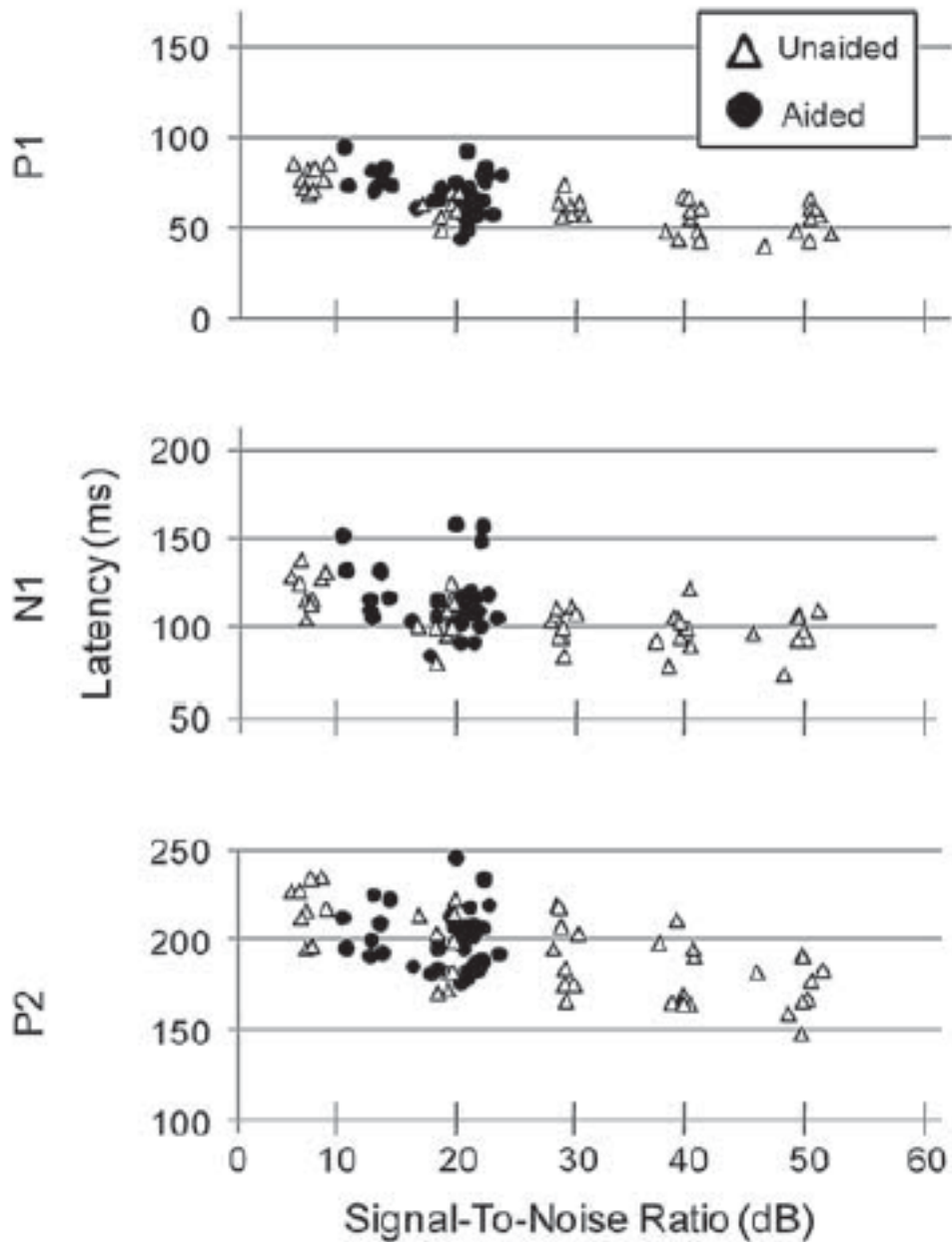
**Figure 4. The relationship between SNR and speech perception as a function of context, originally published in an abstract by Webster and reprinted by Killion (1985). The slope of the performance intensity function is steeper for high context speech material (i.e., sentences) than for low context stimuli (i.e., non-sense syllables).**

### **2.1.3. Individual Noise Tolerance Levels**

A **third** reason supporting the theory that SNR may be an important variable in hearing aid outcomes is that each listener has unique willingness to tolerate noise (e.g., Nabelek et al., 1991). Recently noise tolerance has been measured through a test procedure called the Acceptable Noise Level Test, where speech is presented to the listener at their most comfortable level, then noise is introduced and varied until the listener reports a level they “are willing to tolerate for a long period of time”. The difference between the speech and noise is called the acceptable noise level (ANL), or their preferred listening SNR. On average, people prefer a 10 dB SNR for listening, regardless of age or hearing status (Nabelek et al., 2006). Of primary interest is that a listener’s preferred SNR is not correlated to their speech perception in noise results, as measured at a +8dB SNR (Nabelek et al., 2004). The relationship between a listener’s ANL and SNR-50 is unknown. This suggests that individuals have unique SNR needs (e.g., SNR-50) and desires (e.g., ANL) required to perform in a given listening environment, which may not be correlated. These SNR requirements have largely been unacknowledged in clinical management of patients fit with hearing aids. Furthermore, the results from some studies showed that an individual’s noise tolerance levels predict hearing aid use to a high degree of accuracy (Nabelek et al., 2006; Freyaldehoven et al., 2008). More specifically, the more noise the listener is willing to tolerate, the more likely they are to succeed with hearing aids. However, a subsequent study has found no relationship between noise tolerance and a more comprehensive battery of hearing aid outcomes (Schwartz and Cox, 2012). Nevertheless, the SNR needs of the individual will likely mediate the relationship between HA output SNR and HA outcomes.

#### **2.1.4. Biological Evidence**

**Finally**, there is biological evidence to support the behavioral data showing that SNR may be an important variable in hearing aid outcomes. A possible explanation for the variability in behavioral results comes from neuroscience, where researchers demonstrated that neural codes in the cortex are disrupted by the presence of noise (Tremblay et al., 2006; Souza and Tremblay, 2006.). It may be that some listeners have neural responses that are more sensitive to noise inputs, causing a reduction in speech perception. More specific to aided listening, potential audibility benefits from changes in hearing aid gain may not be realized, depending on the SNR changes made by hearing aid processing and delivered to the listener (Billings, Tremblay et al., 2007, 2009, 2011). For example, Figure 5 illustrates the latency, or timing, of cortical brainwaves (P1, N1, P2) in response to the onset of a 1000 Hz tone (Billings, Tremblay, and Miller, 2011). Open symbols are for an unaided listening environment, where latency decreases as the SNR (measured at the output of the hearing aid) improves, as expected. The closed symbols represent the aided condition, where the SNRs measured at the output of the hearing aid do not change as expected with input SNR (illustrated by the clustering of data points around 10-20 dB SNR). Furthermore, the latency does not decrease with increases in environmental SNR as they do in the unaided condition, due to the fact that the hearing aid output SNR is not improving with improvements in environmental SNR. In other words, the hearing aid gain, or amount of amplification, is making the level of the tone louder, but also the level of the noise. Improvements in environmental SNR are not observed at the output of the aid in this study, and are reflected in these cortical measures.



**Figure 5. The latency of cortical brainwave measures (P1, N1, P2) are shown as a function of SNR measured at the output of the hearing aid. The unaided condition shows latency decreasing as SNR increases, as expected. However, the aided condition shows that all output SNRs cluster around 10-20 dB SNR, suggesting that the hearing aid is amplifying the tone level as well as the background noise level (therefore, the SNR remains similar across conditions).**

### **2.1.5. Summary**

Collectively, SNR could be an influential variable at multiple stages of processing (Figure 1), and could be a potential predictor for hearing aid outcomes. The four reasons discussed previously support the theory that SNR could be an influential variable in hearing aid outcomes, and specifically to speech perception. What we know is that the environmental SNR is strongly correlated to speech perception. What we don't know is whether the hearing aid is modifying the SNR being delivered to the hearing aid user, and how the SNR at the hearing aid output relates to speech perception. Further, it's unknown what role the individual's SNR needs (SNR-50) or desires (ANL) have on the speech perception, hearing aid benefit, and satisfaction in real world environments. This study will take the first step in addressing this problem.

As mentioned previously though, in aided listening the input, or environmental SNR, may be altered by hearing aid processing. The next two sections discuss the potential modification of SNR by hearing aid processing in more detail (illustrated by the hearing aid stage in Figure 1).

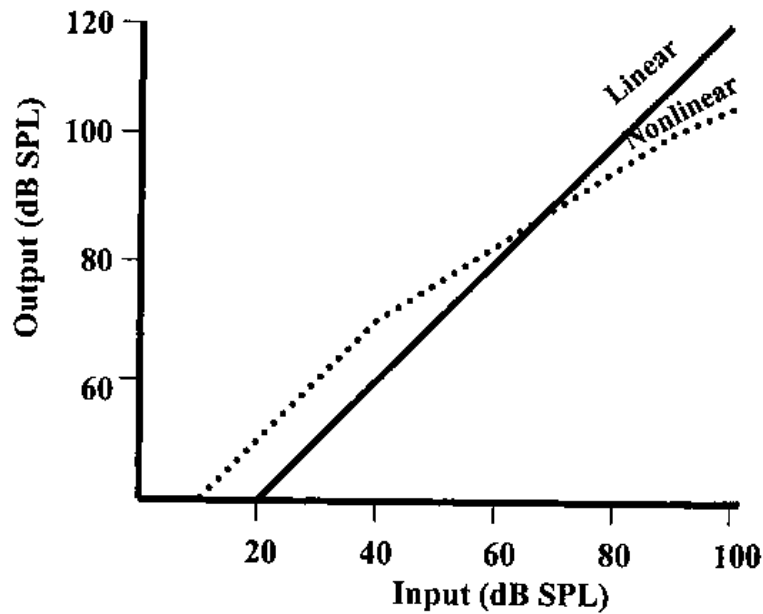
## **2.2. Effects of Hearing Aid Processing on Behavioral SNR and Speech Perception**

There have been a number of hearing aid technologies historically marketed to improve speech perception in noise, such as wide dynamic range compression (WDRC) and noise reduction algorithms. The outcomes of these technologies have traditionally been measured behaviorally, by measuring change in speech perception in noise with or without the technology of interest. This section will provide a tutorial on three types of

hearing aid processing (linear, WDRC, and noise reduction), as well as review the literature on their behavioral outcomes.

### **2.2.1. Linear Amplification**

Listeners with hearing loss were historically aided with linear amplification, where each input level receives the same amplification, or gain. For example, a 50 dB SPL input signal will receive the same amount of gain as a 70 dB SPL input signal (Figure 6). Many researchers have demonstrated benefit in understanding speech in background noise with linear amplification at set SNRs (e.g., Humes et al., 1997; Humes et al., 2001; Moore & Glasberg, 1988; Bentler & Duve, 2000; Humes, 2002), which provides the most benefit to softer speech input levels and at more favorable SNRs. Therefore, the benefit provided by linear amplification over unaided listening is due to the improvements in audibility of the speech signal, which diminishes as presentation level increases, or as SNR becomes less favorable (e.g., Humes et al., 1997; Bentler & Duve, 2000; Humes, 2002). While percent correct scores improve with linear amplification in some test situations at fixed SNRs, the listener's SNR-50 may not improve (Bentler & Duve, 2000). The improvement in SNR-50 with linear aids will depend on the fixed level of either the speech or noise, and the listener's hearing threshold levels. If the level provides good audibility in the unaided condition (e.g., 65 dB SPL), then the aided benefit will be limited.



**Figure 6. Relationship between input level and output level (dB SPL) for linear and non-linear processing (e.g., wide dynamic range compression; courtesy of Valente, 2002). Linear amplification provides the same amount of gain for all levels of input, while compression processing provides more gain for low level sounds and less gain for high level sounds.**

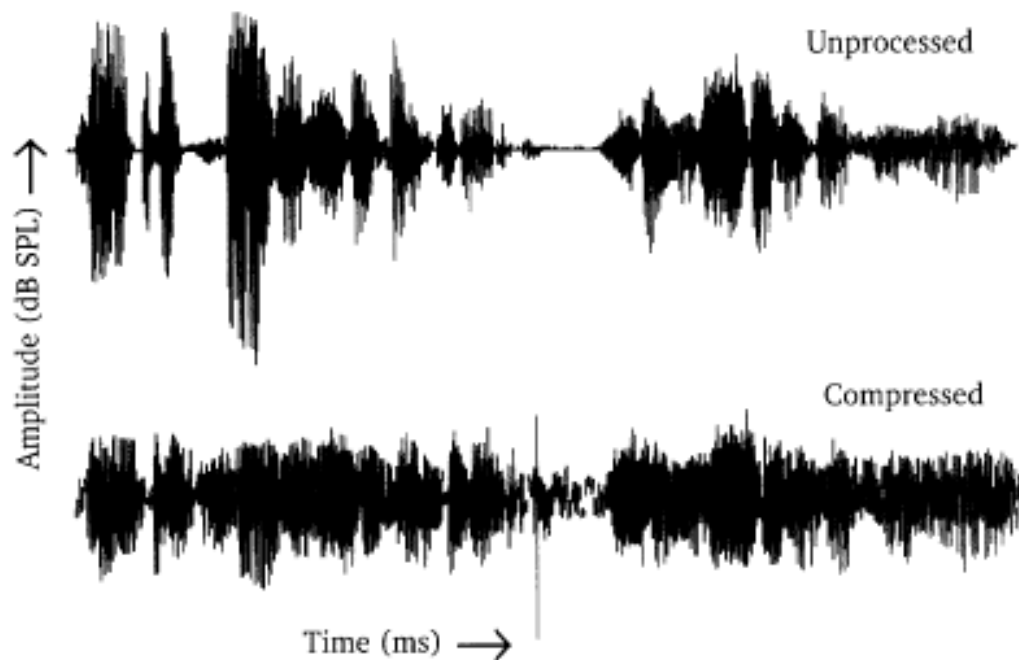
### **2.2.2. WDRC Amplification**

A disadvantage to linear processing is that the high level sounds are often uncomfortably loud given the reduced dynamic range of hearing impaired listeners. Hearing aids are more likely to use WDRC processing, which applies different amounts of gain to the input signal depending on level (Figure 6 and 7). Soft levels receive more amplification than loud levels, designed to accommodate for reduced cochlear function in listeners with sensorineural hearing loss. Figure 7 shows the effect that compression has on a waveform of clean speech, where low level portions of speech are amplified more than the high levels of speech (bottom waveform), unlike the original signal or

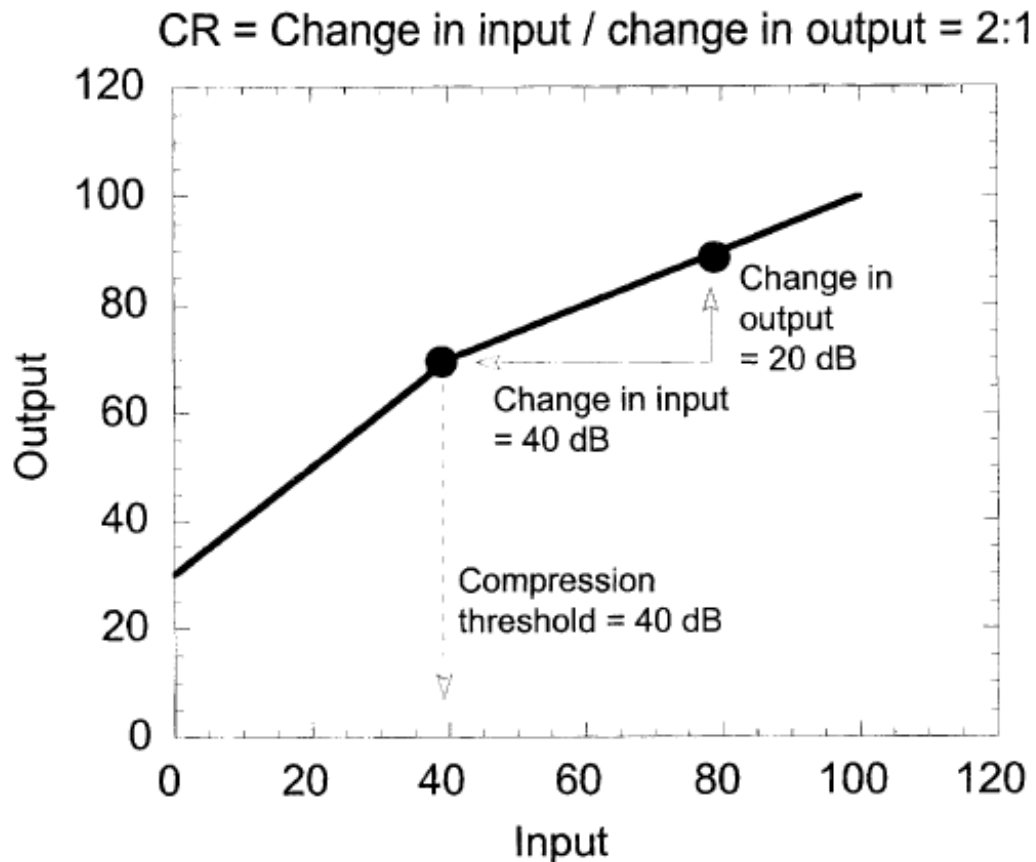
linear processing where all input level are treated equally (top waveform). The amount of difference in gain, or amplification, between soft and loud input levels is specified by the compression ratio. If the gain between a soft and loud signal is very similar, the compression ratio is likely to be close to 1:1, suggesting a 1 dB change in input equals a 1 dB change in output. However if the difference in gain between a soft and loud signal is very different, the compression ratio is likely to be quite high (e.g., 4:1). Figure 8 illustrates the compression ratio effect between a 40 dB and 80 dB input level. The strength of the compression ratio will effect how drastically the envelope of the waveform is distorted.

In quiet environments, WDRC processing may provide a greater improvement to speech perception than linear processing at soft input levels (e.g., 50 dB SPL), since more gain is applied in these situations (e.g., Souza and Turner, 1998; Kam & Wong, 1999; Humes et al., 1999). At conversational and louder input levels, generally listeners provide similar speech understanding results with WDRC and linear processing (e.g., Kam & Wong, 1999; Larson et al., 2000; Humes, 1999). In noisy environments, there are often no consistent differences in speech understanding between WDRC and linear processing regardless of input level or SNR (e.g., Bentler & Duve, 2000; Souza et al., 2007; Kam & Wong, 1999; Larson et al., 2000; Humes et al., 1997). An exception to this may be at high presentation levels (e.g., 90 dB SPL) where linear processing may cause higher distortion if peak clipping is used, and leads to decreased speech perception in noise (Bentler & Duve, 2000). A few studies have demonstrated an inconsistent difference in speech perception in noise between WDRC and linear processing as a function of SNR (Kam & Wong, 1999; Larson et al., 2000). At favorable SNRs, WDRC processing can lead to better performances for listeners than linear processing, while at less favorable

SNRs, linear processing may lead to better performance (Kam & Wong, 1999); although this effect might interact with speech presentation level (Larson et al., 2000). Wide dynamic range compression may not improve the SNR-50 compared to linear processing (e.g., Bentler & Duve, 2000).



**Figure 7. Time waveforms showing an uncompressed (or linearly processed) speech signal and a compressed speech signal (Souza, 2002). In the compression signal, low level speech sounds receive greater gain than high level speech sounds.**



**Figure 8. Relationship between input level and output level (dB SPL) for a compression hearing aid (Souza, 2002). The compression ratio specifies the amount of gain reduction to apply as input level increases.**

### 2.2.3. Noise Reduction Algorithms

A disadvantage to WDRC processing is that during soft portions of speech, the background noise may be amplified to an unwanted level. An attempt to improve speech perception in noise has been through digital noise reduction algorithms. Many modern hearing aids implement a classification scheme, designed to analyze the spectral, temporal, and/or amplitude characteristics of the incoming signal and classify it as speech, noise, speech+noise, music, or other. Once classified, decision rules are

implemented to make various changes to the hearing aid settings, which could include activation of digital noise reduction algorithms, among other things. Decisions may be made about the strength and time constants of gain reduction in these algorithms. The digital noise reduction algorithms of interest to this study were those typically used for long lasting noise, such as babble or steady-state noise (as opposed to those used for transient or wind noises). These noise reduction algorithms usually fall into two categories: modulation based and spectral subtraction.

Modulated based algorithms use information on the modulation rate and depth of the incoming to signal to estimate whether speech is present and at what SNR (Chung, 2004; Bentler & Chiou, 2006; Kates, 2008; Chung, 2010). Once these decisions are made, the algorithm decides how much gain reduction to implement. If the modulation rate is low (e.g., speech dominated) and the modulation depth is high (likely a favorable SNR), then the gain is minimally reduced (Chung, 2010). If the modulation rate is high and the modulation depth is low, then gain reduction is greater. These gain changes generally occur slowly within each channel designated by the noise reduction algorithm. The amount of gain reduction and how fast changes are applied is not straight forward, and likely depends on the modulation depth, rate, overall level, frequency weighting, and type of noise (Chung, 2004; Bentler & Chiou, 2006; Chung, 2010, Dillon, 2012). This noise reduction method doesn't improve the SNR within a channel, given that the gain reduction is applied to both the speech and noise within the channel, but it may improve the overall SNR compared to the listener's hearing thresholds (Dillon, 2012). Therefore, speech perception is generally not improved with modulation-based algorithms (e.g., Walden et al, 2000; Alcantara et al., 2003; Bentler et al., 2008; Zakis et al., 2009).

Spectral subtraction methods (e.g., Wiener filtering) use the estimated spectra of the noise and speech plus noise to implement the decisions about gain reduction. The estimated spectrum of the noise is made during portions of the signal without speech present, then the gain of the speech plus noise signal is adjusted to closely match that of more “clean” speech (Chung, 2004; Bentler & Chiou, 2006; Chung, 2010; Dillon, 2012). The amount of gain reduction depends on the estimated SNR and noise spectrum, and the gain reduction is typically faster than that for modulation-based methods (Kates, 2008). Only one study known to date has evaluated a recent version of a spectral subtraction algorithm; however it was in combination with a modulation-based algorithm (Ricketts & Hornsby, 2005). In this study, the researchers found no difference in speech perception in noise between noise reduction activated and deactivated. However, it’s difficult to attribute the lack of change in speech perception with this algorithm to one noise reduction method or the other.

Although the effects of noise reduction algorithms on speech perception are fairly conclusive, subjective outcomes are not as clear. Often listeners report increased tolerance for noise (Mueller et al., 2006; Palmer et al., 2006), decreased listening effort (Sarampalis et al., 2009), improved comfort (Bentler et al., 2008; Zakis et al., 2009), and a stronger preference (Bentler et al., 2008; Zakis et al., 2009) for noise reduction activated compared to inactivated. The conflicting reports between speech perception and subjective benefits are curious, and analyzing the acoustic changes to the hearing aid output signal could provide clues as to what changes are perceptible to listeners. For example, perhaps the manner in which the SNR is being modified is driving the conflict in perception. Some algorithms may only change the SNR when speech is not present

(e.g., during the pauses of speech), which will not likely influence speech perception but may influence listening comfort.

#### **2.2.4. Summary**

In summary, it appears that most hearing aid technologies are unable to improve speech perception in noise. One exception is the use of directional microphones in laboratory environments. However, there may be subjective benefits to WDRC and noise reduction algorithms, such as increased comfort and sound quality. The lack of consistent improvement in speech perception in noise with these hearing aid technologies, combined with our knowledge of a strong relationship between SNR and speech perception, lead us to believe that most of these hearing aid technologies are not changing the SNR at the output of the hearing aid. However, more recently, the SNR at the output of the hearing aid has been measured acoustically, and these results suggest otherwise. The next section will discuss the changes in SNR made by hearing aid processing, measured acoustically at the output of the hearing aid (illustrated by the output signal in Figure 1).

### **2.3. Role of Hearing Aid Processing on Acoustic SNR**

The output signal generated by hearing aid processing has been studied extensively throughout the years, although rarely with speech and noise signals combined. First we review some studies that have shown the multiple hearing aid features that can alter the acoustics of a speech signal, even in quiet. For example, several researchers have evaluated the effects of compression parameters on temporal envelope, or the slow fluctuations in a speech signal (Hickson & Thyer, 2003; Jenstad &

Souza, 2005; Jenstad & Souza, 2007; Stone & Moore, 2007; Souza et al., 2006), spectral contrast or consonant vowel ratio (Stelmachowitz et al., 1995; Hickson et al., 1999; Jenstad & Souza, 2005; Bor et al., 2008), bandwidth (Stelmachowitz et al., 1995), effective compression ratio (Stelmachowitz et al., 1995; Souza et al., 2006; Henning & Bentler, 2008), dynamic range (Henning & Bentler, 2008), and audibility (Stelmachowitz et al., 1995; Souza and Turner, 1999; Henning & Bentler, 2008). As the number of compression channels increases, spectral differences between vowel formants will decrease (Bor, Souza, & Wright, 2008), the level of the consonants compared to the level of the vowels is increased (Hickson & Bryne, 1995), and dynamic range decreases (Henning & Bentler, 2008). Similarly, as compression time constants get shorter, the temporal envelope will reduce/smear (Souza et al., 2006; Jenstad & Souza, 2005; Jenstad & Souza, 2007), and the effective compression ratio will increase (Henning & Bentler, 2008). A stronger compression ratio has been linked to greater temporal envelope changes (Jenstad & Souza, 2007; Souza et al., 2006). Linear amplification may also create acoustic changes, such as changes in spectral contrast if the high frequencies have much more gain than the low frequencies (e.g., Stelmachowitz et al., 1995). The acoustic changes caused by compression processing have been linked to perceptual changes in many cases (Jenstad & Souza, 2005; Jenstad & Souza, 2007; Bor, Souza, & Wright, 2008, Stelmachowitz et al., 1995; Hickson & Thyer, 2003; Stone & Moore, 2007). In general, as the compression settings increase (e.g., time constants or compression ratio), the more modifications to the acoustic signal are made, and the more detrimental to perception.

### **2.3.1. Speech in Noise**

Less evidence is available on the acoustic effects of hearing aid processing to speech plus noise signals, likely because it's technically difficult to separate the speech from noise at the output of a hearing aid. Historically, speech and noise stimuli were presented sequentially to the aid, which has shown the hearing aid output changes depending on the type of speech or noise signal, and compression settings used (Henning & Bentler, 2005). However, more advanced hearing aid processing such as noise reduction algorithms react to the combined speech and noise characteristics. Therefore, a technique to analyze the changes to speech and noise signals presented simultaneously to the hearing aid would likely better predict outcomes.

Recently a few techniques were developed to measure acoustic changes to speech and noise, when presented simultaneously to the hearing aid (Bell et al., 2010; Hagerman & Oloffson, 2004). Bell et al. (2010) describes a technique that uses two synthetic noise signals, which are interleaved, yet non-overlapping, in the frequency domain. One of the signals contains odd harmonics, while the other contains only the even harmonics, both at an identical and very low fundamental frequency. In their example, they used a fundamental frequency of 1.5 Hz. The signals are separated at the output by using a Fast Fourier Transform (FFT) analysis, to convert the output waveform into the frequency domain. The FFT uses the same frequency spacing as used in the input signal (1.5 Hz) and an integer number of periods are included in each time window to which the FFT is applied. The interleaved harmonic method is attractive since only one measurement is needed to estimate the SNR and this approach could theoretically be useful for real ear measures of SNR. However, the between-subject variability in real ear measures using the technique described by Bell et al. (2010) is

quite high, even when the same hearing aid was used. This may be due to variability in probe tube placement during repeated real ear measures. Further, distortion arising due to non-linear processing has not been evaluated with this measure. A possible disadvantage to using this method is that it assumes the “noise” and “speech” signals occur in separate frequency regions. This is not likely the case in real life situations. More details on this method can be found in Appendix A.

While there have been more complex techniques proposed for separating speech and noise from the output of a hearing aid (e.g., coherence; Kates, 1992; Kates and Arehart, 2005; Arehart et al., 2007; Ma et al., 2009), a technique developed by Hagerman & Olofsson (2004) is proving to be an effective way to analyze hearing aid processing with simultaneous speech and noise inputs.

### **2.3.2 Phase-inversion Technique (Hagerman & Olofsson, 2004)**

In Hagerman’s phase-inversion approach, simultaneous speech and noise signals are presented to a hearing aid, and the speech and noise signals are separated at the output of the hearing aid. In order to separate the speech and noise signal at the output, two measurements must be taken. In the first measure, the original-phase speech and noise signals are presented. In the second measure, the noise (or speech) is phase-inverted. The two measures have identical modulation spectra, and can be defined as:

A input = speech + noise

B input = speech – noise

Both A and B inputs are identical at the input of the hearing aid, except for one of the signals being 180-degrees phase inverted. At the output of the hearing aid, the speech and noise signals will be present as well as an error term:

$$A \text{ output} = \text{speech} + \text{noise} + \text{errorA}$$

$$B \text{ output} = \text{speech} - \text{noise} + \text{errorB}$$

The error term is from internal noise in the hearing aid, interactions from the speech and noise, and distortion from the hearing aid processing. To extract the speech and noise levels, Hagerman and Olofsson assume the hearing aid treats the combined input signals similarly and that output signals can be added (called superposition). Superposition means that the output of a system with a number of independent inputs presented simultaneously should be equal to the sum of the outputs if each input were presented alone (Harrington & Cassidy, 1999). For example, suppose we pass a signal through a system that is the sum of two other signals ( $x_1 + x_2$ ), and the output of the system is  $y_3$ . Then we pass through  $x_1$  and  $x_2$  separately to obtain outputs of  $y_1$  and  $y_2$ . Superposition is met if  $y_3 = y_1 + y_2$  and is a characteristic of linear time invariant systems, which Hagerman & Olofsson assume of hearing aids. They state, however, that the time constants must be slow enough to be quasi-linear. To obtain the speech and noise output separately, the two outputs are combined:

$$\text{Extracted Speech} = A \text{ output} + B \text{ output} = 2 * \text{speech} + \text{errorA} + \text{errorB}$$

$$\text{Extracted Noise} = A \text{ output} - B \text{ output} = 2 * \text{noise} + \text{errorA} - \text{errorB}$$

If the error signals are small enough (discussed later), the root mean square (RMS) levels can be measured on each extracted signal and the SNR calculated. This can be measured as a function of frequency. To obtain a single SNR value, the AI/SII method can be applied (discussed later), or an average SNR across frequency can be calculated.

The error terms need to be considered carefully to decide if the method is valid. This method is based on the assumption that the time constants are long enough to make the system quasi-linear. Most compression hearing aids will not meet this assumption, mainly due to short attack times in the compressor (usually under 10 ms; Dillon, 2001), but also if the noise reduction algorithm has short time constants. Quantification of error or distortion terms can be done in a few ways, as proposed by Hagerman and Olofsson (2004). First, a simple listening check on the output signal, after extracting either the speech or noise signal, can be done. Second, a correlation can be calculated between the two speech and noise output combinations. The correlation should be low, since speech and noise should be quite different. However, the internal noise of the aid can increase the correlation. And finally, the distortion can be quantified by the use of Hilbert pairs (Olofsson & Hansen, 2006; Appendix B). Rhebergen et al. (2009) proposed a new method to separate speech and noise, which supposedly alleviates the problem of using Hagerman's technique with non-linear hearing aids (discussed in Appendix C). However, this method has not been validated, nor has the reliability been assessed. Therefore, Hagerman's separation technique will be used in this study.

A few researchers have attempted to quantify the distortions or errors caused by using Hagerman's technique in non-linear hearing aids and have concluded this method

is feasible provided care is taken in measurements (Naylor and Johannesson, 2009; Zakis & Jenstad, 2011; Hartling, Wu, & Bentler, 2012). Zakis and Jenstad (2011) quantified a number of errors generated from using Hagerman's technique with nonlinear processing, including 1) the noise floor (generated by extracting silence from completely out of phase recordings), 2) residual noise signal (generated by extracting speech and measuring the dB attenuation during a silent portion of the waveform between the extracted speech file and the non-extracted file), 3) fidelity of speech (generated by comparing extracted speech to clean speech in quiet). They compared these errors measured on the stimulus files (ranged from -51 dB to -61 dB) to those measured on the recorded and extracted files. The use of the Hagerman technique in a soundfield with no hearing aids did increase the error in all cases (-23.6 dB to -20 dB), but adding hearing aid non-linear processing only further increased the error by 3-4 dB.

Hartling, Wu, and Bentler (2012) also quantified a type of error generated from using Hagerman's technique with non-linear processing. They were interested in assessing the hearing aid algorithm stability when using this method. Two of the same measurements were taken and the outputs were subtracted from one another. Any residual noise was measured and the levels were subtracted from one of the original recordings. The resulting levels were called the attenuation, with a larger attenuation being more desirable. They measured the attenuation values across 9 SNR conditions, 10 hearing aids, and 6 hearing aid processing conditions (linear, WDRC, directional microphones, digital noise reduction, feedback cancellation, and all algorithms combined). For the linear, WDRC, and directional microphones conditions, they reported attenuation values of 25-50 dB. For the digital noise reduction and feedback cancellation algorithms, the attenuation was poorer and varied across input SNR. For

digital noise reduction, the attenuation values were generally better than 25, although two conditions showed values of only 15 dB. With feedback cancellation activated, attenuation in some aids dropped to almost 0 dB. These two studies provide guidelines to evaluate the errors generated when using Hagerman's phase-inversion technique with non-linear processing. The results from both of these studies are in agreement with error values reported by Naylor and Johannesson (2009). These studies also highlight the importance of quantifying the error terms when using Hagerman's phase-inversion technique for quantifying HA signal outputs.

Several studies have used Hagerman's phase inversion method to quantify acoustic changes to speech and noise post-hearing aid processing, most of which are measuring SNR changes (Hagerman & Olofsson, 2004; Souza et al., 2006; Naylor and Johannesson, 2009; Chong & Jenstad, 2010). Hagerman and Olofsson (2004) used the phase-inversion technique to test the noise reduction algorithms of five hearing aids programmed in linear gain, with the noise reduction algorithm turned off or on. The two speech signals used were a male voice speaking an artificial language and a simulated male speaker from the International Collegium of Rehabilitative Audiology (ICRA). The three types of noise signals were un-modulated, speech-shaped noise (same spectral shape as the ICRA voice), 6-talker babble noise, and noise with a pronounced high frequency content (mimicking a printing shop). The first 16 seconds of each input was used to stabilize the response of the aid, then the last 8 seconds were used for the analysis. The speech was presented at 60, 70, and 80 dB SPL, and at SNR of -5, 0, and +5 dB. Distortion from this technique was measured using two input signals: the original signal and the Hilbert transform of the original signal. The output

from the second signal will be the Hilbert transform of the first output signal, if the hearing aid is linear and noise-free.

The results showed that the distortion was small enough between noise reduction on and noise reduction off to use this method (0 to -30 dB, depending on frequency). All noise reduction algorithm comparisons were made referencing the noise reduction deactivated condition, which had a standard deviation between measurements of 0.22 dB (a low measurement error). Further, they performed a listening check of the cancelled speech or noise output signals and found them to be satisfactory. Unfortunately, they did not include distortion estimates for the two compression conditions (with a compression ratio of 3:1) they measured to assess feasibility using this technique. The noise reduction algorithms tested showed a variety of improvements in SNR when activated, depending on type and SNR of the input signal.

In their discussion, Olsen et al. (2005) attempted to show whether a compression benefit observed objectively (using the Hagerman and Olofsson, 2004 technique) correlated to the speech intelligibility benefit measured for compression processing. The objective compression benefit was defined as the SNR improvement estimated for compression processing over linear processing using the phase-inversion technique. Speech intelligibility was measured with sentences for linear and compression processing, and the benefit was the difference in scores. Olsen et al. (2005) also speculated that at favorable SNRs, the speech signal would control the compressor and result in gain fluctuations. These fluctuations could in turn, reduce speech intelligibility, and explain one reason why only some listeners benefit from fast-acting compression compared to linear processing. Alternatively, at less favorable SNR, the noise will control the compressor, which will keep the gain relatively constant,

depending on the noise signal. Therefore they simulated an unusual circuit for compression hearing aids to control the SNR influencing the gain. To do this, they had one speech and noise signal controlling the compressor (2:1 compression ratio, 3-channels), which was separate from the speech and noise signal used for testing speech intelligibility. The authors fixed the sensation level of the noise peaks to 15 dB SPL, and used four SNR (0, -5, -10, and -15 dB) to control the compression processor. In the audio signal used for speech testing, the noise level was fixed while the speech level was adapted to find the SNR where listeners obtained 40% correct. Speech intelligibility results showed that compression processing was significantly beneficial to all six normal hearing listeners at all SNR, except for 0 dB SNR where linear processing was significantly better than compression. As SNR improved, the benefit from compression over linear processing declined. The observed and predicted benefits are quite similar, suggesting the phase-inversion method for objectively measuring SNR changes correlates well to speech intelligibility changes. However, the unusual compression circuitry simulated in this study limits the interpretability of these results.

Souza, Jenstad, and Boike (2006) used a modified version of the Hagerman & Olofsson (2004) approach to quantify how compression was changing the long-term SNR at the output of the hearing aid. They used a single-channel and a 2-channel compression hearing aid with a 2:1 compression ratio, attack time of 5 ms, release time of 10 ms, and compression threshold of 45 dB SPL in both hearing aid conditions. A linear processing mode served as the control condition. The input was CST connected discourse in quiet and with a background of speech-shaped, unmodulated noise at SNR of -2, +2, +6, and +10 dB. They defined the SNR as the RMS level between the speech signal and the noise signal. They modified the original Hagerman & Olofsson (2004)

approach by using three combinations of input signals: 1) original speech + original noise, 2) original speech + phase-inverted noise, and 3) phase-inverted noise + original noise. Hagerman & Olofsson (2004) only used the first two combinations. The reasons behind this modification are unknown. Measurement distortion was estimated by correlating the input and output signals with the hearing aid in linear processing and found a very high correlation. Distortion with non-linear processing was not assessed. Results of the change in SNR for the linear condition was as expected: the authors found that the linear processing condition did not change the SNR. However, both single- and multiple-channel compression processing degraded the SNR at the output of the aid by 1-4 dB. This was particularly true for less favorable input SNRs. For example, with an input of -2 dB SNR, the single-channel aid showed an output of -5 dB SNR and the multiple-channel aid showed an output of -3 dB SNR. To reiterate, Souza et al. (2006) found that as the input SNR became less favorable, compression processing degraded it even more than linear processing. These results are in direct contrast to those reported by Olsen et al. (2005) who found compression benefits at less favorable SNRs, and linear benefits at more favorable SNRs.

Naylor & Johannessen (2009) tried to remedy the discrepancy on how compression affects the SNR between Olsen et al. (2005) and Souza et al. (2006). They used the phase-inversion approach (Hagerman & Olofsson, 2004) to measure the output SNR of a single-channel hearing aid and systematically varied the compression parameters and input SNR. The target speech they used was short sentences embedded in three types of background noise: 1) unmodulated noise shaped like the long-term average of the target sentences, 2) modulated ICRA noise shaped like the long-term average of the target sentences, and 3) the same single-talker as used for the target

speech, but different sentences. In general, they found that at positive SNR inputs, compression reduced the output SNR to a less favorable level. At negative SNR inputs, compression increased the output SNR to a more favorable level. The effects of varying the compression ratio and time constants were such that the more aggressive the setting (higher compression ratios and faster time constants), the greater deviation the output SNR was compared to the input SNR. In other words, the output SNR varied less (or stayed closer to 0 dB SNR) as the compression was made stronger. These results are consistent with those found both by Olsen et al. (2005) and Souza et al. (2006), with the change in SNR depending on the compression settings used in the hearing aids, as well as the input SNR.

Unfortunately, Naylor and Johannessen did not fully report the error terms or distortion measured due to compression processing and using this technique. The error terms were measured by combining the outputs of the original speech + original noise and phase-inverted speech + phase-inverted noise. The authors stated the combination of these two signals should result in a perfect null. It is unknown if this was performed with linear or compression processing. The results showed that the distortion was -20 dB below the softest noise or speech level used in the study. However, if the hearing aid compressor uses fast attack times, the error terms may increase. This would lead to higher output levels than when slower attack times are used, and therefore, appear to show minimal differences in output SNR between attack times.

Hagerman's technique has also been used to quantify effects of digital noise reduction. Chong & Jenstad (2010) measured the level changes to noise and phonemes made by digital noise reduction with linear processing as a function of the number of channels, using speech in three types of noise stimuli. They found that digital noise

reduction algorithms used in 9 hearing aids from four manufacturers were effective at reducing pink and cafeteria noise, but slightly increased ICRA noise. The reduction in noise was greater as the number of channels increased. Many of the fricatives did not change in level post-hearing aid processing, but some were enhanced up to 10 dB, particularly in pink noise and with greater number of channels. Furthermore, the spectrum of fricatives did change (both positive and negative) compared to noise reduction off, which increased as the number of channels increased. Although the SNR wasn't reported in this analysis, we can speculate that digital noise reduction may change the SNR given the reductions in noise and enhancements in speech reported here. In other words, we might anticipate noise reduction algorithms to potentially improve the SNR.

Whether the changes in the speech acoustics measured post-Hagerman separation are perceptually relevant is unknown. However, in unaided listening situations, researchers have reported that very small changes in SNR can lead to large changes in speech perception (i.e., 1 dB change in SNR approximates a 10% change in speech perception; Nilsson et al., 1994; Laurence, Moore, and Glasberg, 1983). Therefore, it's possible that the acoustic SNR changes suggested here could explain some of the variability in aided speech perception.

### **2.3.3. Summary**

In summary, using Hagerman and Oloffsson's phase-inversion technique is proving to be an effective method for defining hearing aid output SNR. Using this technique, researchers have shown that hearing aid algorithms may improve or diminish the SNR, depending on the hearing aid settings (Hagerman & Oloffson, 2004; Souza & Jenstad, 2006; Naylor & Johannesson, 2010; Chong & Jenstad, 2010; Hartling

et al., 2012). Changes in SNR have been demonstrated with a limited number of WDRC algorithms in simulated environments (Souza & Jenstad, 2006; Naylor & Johannesson, 2010), and with a simulated noise reduction algorithm in combination with linear processing (Hagerman & Oloffsson, 2004). If acoustic changes to SNR measured at the hearing aid output could be linked to perception, this may be a clinically useful measure to predict hearing aid outcomes. What needs to be addressed first, however, is whether clinically applicable hearing aid settings and algorithms result in SNR changes at the hearing aid output, and whether they relate to speech perception.

## **2.4. Role of Audibility**

It is possible that the listener's hearing threshold levels will limit any perceptual consequences of changes in speech acoustics if the signal is not audible to the listener. In other words, for benefit to be obtained, the level of the signal needs to be greater than the listener's hearing thresholds. For example, if the SNR improves by 3 dB with hearing aid processing, yet the speech and noise levels are still below either the listener's pure tone hearing thresholds, then this individual will not experience an improvement. To account for changes in audibility as a way of explaining results, the Speech Intelligibility Index is often used (previously known as the Audibility Index).

The Audibility Index (AI) quantifies the role of audibility in understanding speech. It was originally developed over 60 years ago (French and Steinberg, 1947) and has evolved over the years. Hence, there are many variations depending on which version is used. Briefly, the AI divides the speech spectrum into bands, each of which contributes independently to the speech recognition performance. The SNR is quantified for each band to determine how heavily the band should be weighted towards

the prediction of speech recognition. Once the weighting is applied to each band, the importance of that frequency band to speech intelligibility is factored in, and the resulting values are summed. The AI result is a number between 0 and 1, where 0 assumes the speech intelligibility will be quite low and a 1 means the speech intelligibility will be good. Therefore, the AI can be thought of as the proportion of speech that is audible.

More specifically, the AI (A in this equation) can be calculated as:

$$A = P \sum_{i=1}^n I_i W_i$$

where n is the number of frequency bands, i is the band number, I is the importance of that frequency band to speech perception according to theoretical and empirical testing, W is the weighting being applied to that band based on the SNR (Pavlovic, 1987). The proficiency factor (P) represents how clear the speech material is and how familiar the listener is with the talker. Under ideal situations, P = 1. The number of bands (n) used has been 20 (with equal importance), 15 (1/3-octave bands), and 5 (octave bands). Humes et al. (1986) recommends using the original 20 bands with each contributing 5% to the total speech intelligibility score, as suggested by French and Steinberg (1947). A transfer function can be applied to the A value to get the predicted speech intelligibility, depending on the speech material used (French and Steinberg, 1947; Appendix D).

The distribution of the assumed input speech spectrum is averaged over 125 ms, which gives a dynamic range of 30 dB in each frequency band (Humes et al., 1986; Pavlovic, 1987). With a SNR of +18 dB, the maximum contribution will be obtained in the band. With a SNR of -12 dB, the minimum contribution will be obtained in the

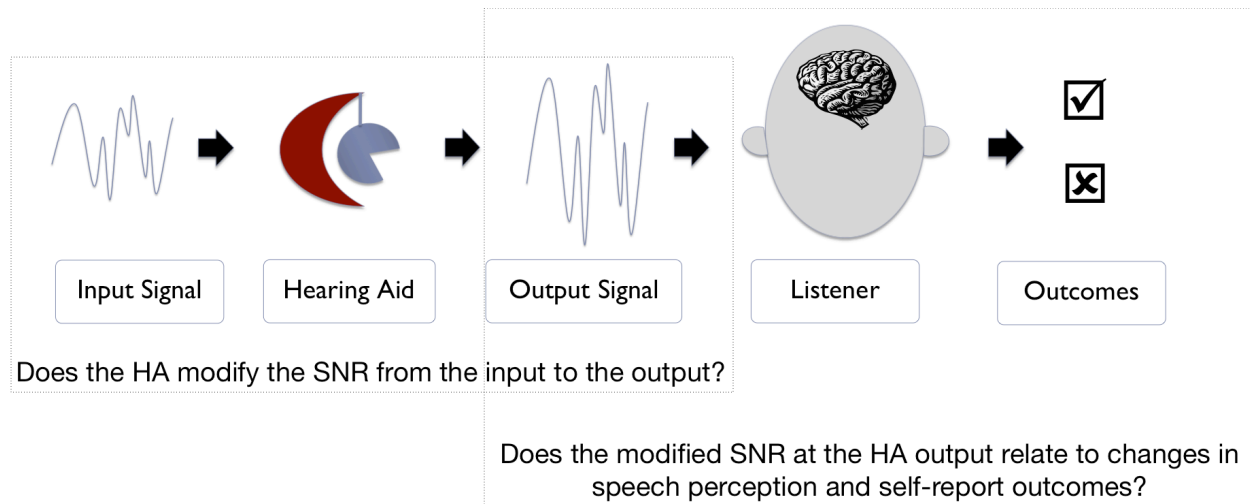
band. The original AI model assumes that the peaks of speech are important and 12 dB is added to the RMS of the speech level in each band to represent the peaks. To calculate the SNR, the RMS+12 dB of speech is compared to the RMS of the “noise.” The model assumes that the listeners hearing thresholds in quiet can be converted to equivalent internal noise levels. Therefore, the “noise” used in the model can be: 1) the higher of either the internal noise/hearing thresholds of the listener or the level of actual acoustic noise in the environment, or 2) the power sum of the two noise levels (Humes et al., 1986). To increase accuracy, the distribution of the speech material used for specific purposes could be measured and used instead of the average spectrum provided (which is similar to those discussed in the Introduction section).

In the 1997 revision of the ANSI S3.5 standard, the term Speech Intelligibility Index (SII) replaced the term AI, and is the most current version of this standard. In this standard, the number of bands (n) is flexible and can be defined by the user. Also, the peaks of speech are assumed to be 15 dB above the average speech levels. Two corrections factors have also been applied to the SII, which include a correction to account for the effects of upward spread of masking and for reduced speech intelligibility when listening at high presentation levels. This version of the SII was used in the current study to estimate the audibility in each condition, for each listener.

## **2.5. Research Questions**

The focus of this research links two stages in the processes of understanding aided speech in noise (Figures 1 and 9). Specifically, I will answer two questions: 1) to what extent does hearing aid processing change the input SNR? 2) Does a change in the hearing aid modified SNR correspond to a change in speech perception? Because

hearing impaired listeners often don't benefit from changes in SNR as much as normal hearing listeners, these questions were addressed in both normal hearing (NH) and hearing impaired (HI) groups.



**Figure 9. Framework for research questions, linking multiple stages of speech in noise processing.**

While there are several relatively independent domains of hearing aid outcomes, this project will focus on one most often used in clinical practice, speech perception (i.e., understanding) testing. Speech tests as outcome measures have been used extensively over the last century, and most researchers agree that measuring speech understanding in the clinic is an important part of the hearing aid validation process (e.g., Humes, 1999; Humes et al., 2001; Gatehouse, 1994; Cox et al, 2007). An example of a commonly used test in the clinic is the Hearing in Noise Test (HINT; Nilsson et al., 1994), where short sentences are presented to the listener in a background of unmodulated noise. The speech level is varied until the listener understands 50% of the sentences, or the

SNR where 50% correct is achieved. On the contrary, self-report measures are used to estimate performance in the listener's real world environments and are also vital to the validation process, particularly since the correlation to speech perception measures in the clinic are weak. An example of a self-report measure is the Abbreviated Profile of Hearing Aid Benefit (APHAB; Cox & Alexander, 1995). This questionnaire asks the listener to estimate the percentage of problems they have in different listening situations, both with and without their hearing aids. Normative data is available for comparison. The contributions to self-report measures largely involve non-auditory factors (e.g., personality); hence the focus on aided speech perception for this study.

Based on the research presented to so far, it's likely that hearing aid processing (both WDRC and digital noise reduction) is modifying the input SNR. The studies reported have used a limited set of compression algorithms, mostly simulated, or a limited set of hearing thresholds. It is of interest to determine whether the SNR is an important variable in hearing aid outcomes, and whether this is a clinically useful measure. Therefore, this study aims to further evaluate how hearing aid processing modifies the SNR for a given individual, and determine whether the post-Hagerman SNR calculation has any perceptual meaning. It is hypothesized that the hearing aid algorithms used in this study will modify the SNR from input to output, and that changes in SNR at the hearing aid outcome will correlate to changes in speech perception. Because hearing impaired listeners often don't benefit from changes in SNR as much as NH, it is expected that correlations for the NH group will be higher than for the HI group.

## **Chapter 3. Method**

### **3.1. Design**

A cross-sectional, within-groups design was used. All subjects completed the same protocol, regardless of hearing threshold levels. The study was approved by the University of Washington IRB.

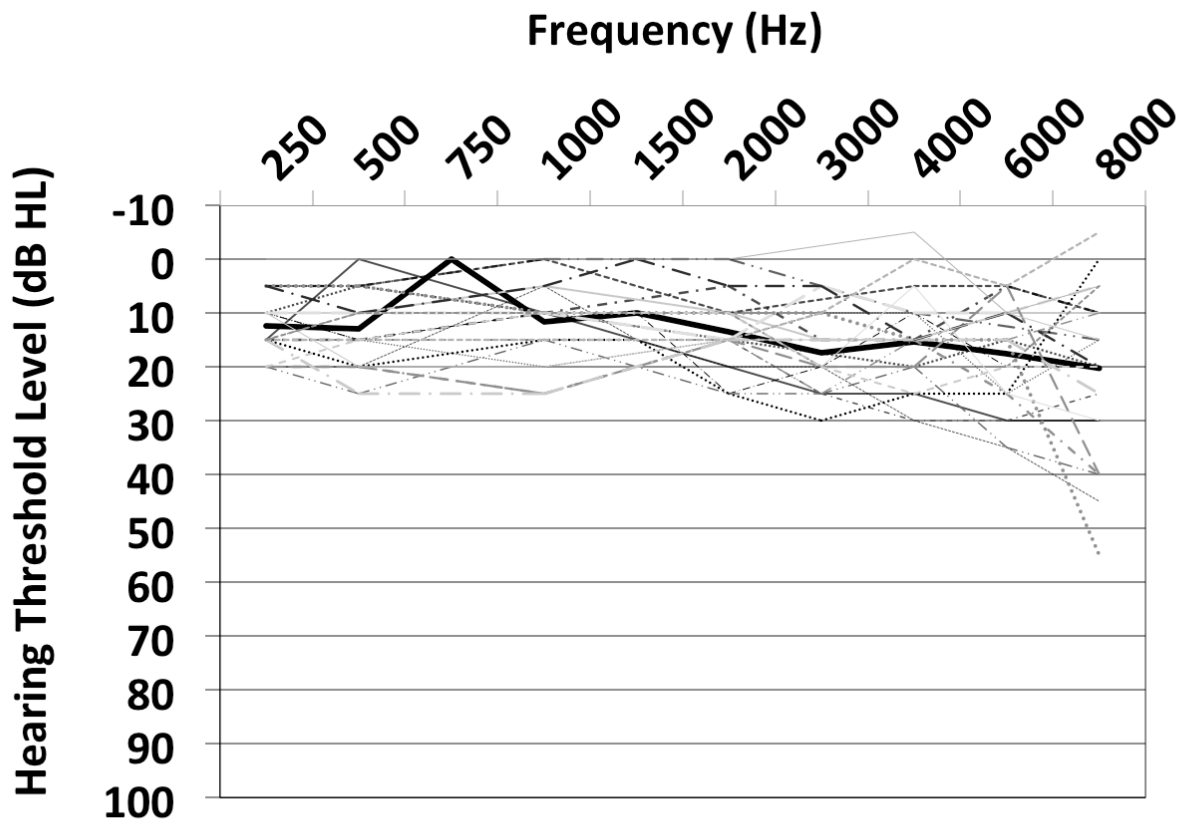
### **3.2. Subjects**

Subjects were recruited from the existing Communication Participant Pool. The database was searched for listeners with: 1) hearing thresholds better than 70 dB HL in at least one ear from 250-4000 Hz, 2) no conductive or mixed etiology in the selected test ear (described later), and 3) English speaking. People with hearing thresholds worse than 70 dB HL were excluded because it would be difficult to ensure audibility of the test signals. All identified potential subjects were called until we obtained 50 participants (25 in each hearing status group, described later), which was partially decided on with a power analysis (power = 80%).

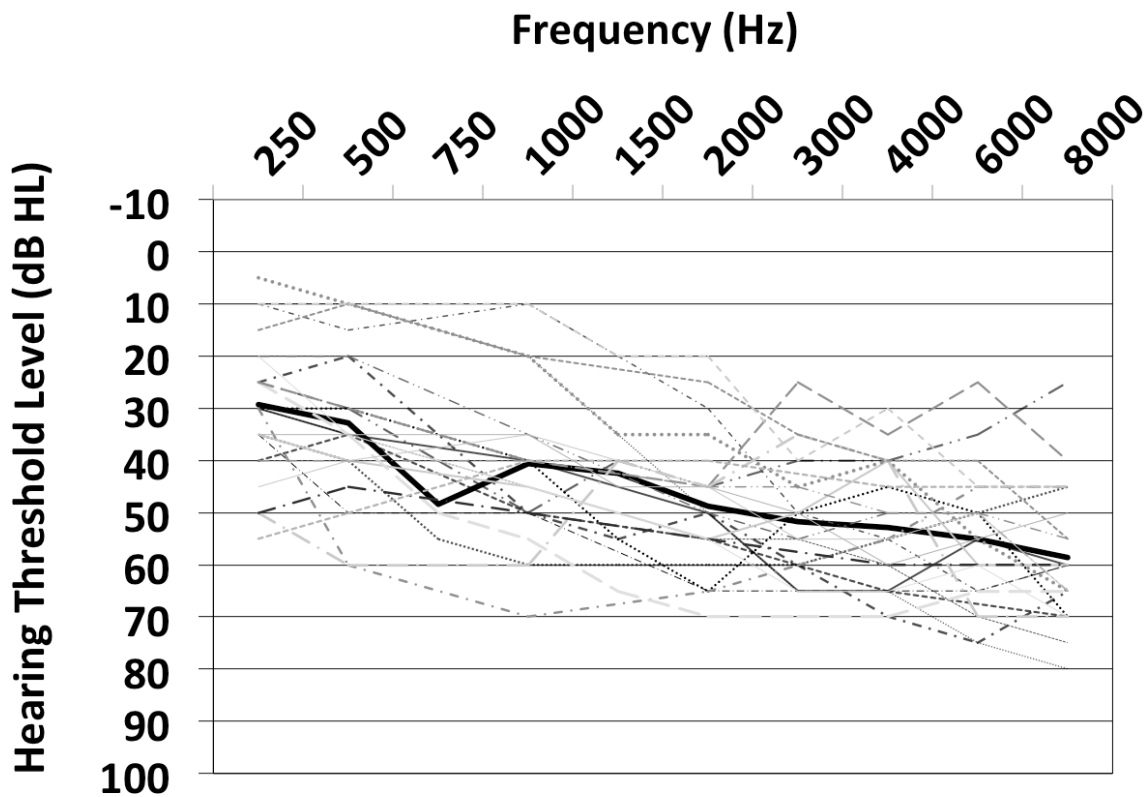
A hearing test was performed on everyone who did not have a current test within the last six months. Three participants were screened, but not included in the study, due to the task being too difficult (e.g., SNR-50 was >12 dB) or thresholds being too poor upon re-testing their hearing. Informed consent was obtained from all 50 subjects who participated in the experiment. The subjects were assigned to one of two hearing status groups: normal hearing or hearing impaired. Twenty-five subjects had normal hearing (NH group), with air and bone conduction thresholds at or above 25 dB HL from 250-4000 Hz. One subject had a threshold at 30 dB HL at 3000 Hz. The other twenty-five subjects had sensorineural hearing loss (HI group), mild to moderately-severe in degree. A conductive component was ruled out by observation of less than a

10dB air-bone gap and a detailed case history. In both groups, the better-hearing ear was selected for testing, and the opposite ear was plugged with a comply tip. Figure 10 and 11 show the mean and individual audiograms for the better-hearing ears of both groups of subjects. The pure tone averages (.5, 1, 2 kHz) for the NH and HI groups were 12.6 (SD=5.9) and 40.7 (SD=12.5) dB HL, respectively. The mean age of the NH and HI groups were 62 years (SD=12.9) and 67.5 years (SD=18.09), respectively, and were not statistically different from each other ( $t = -1.239$ ;  $p = .221$ ).

Of the 25 hearing impaired listeners, 21 (84%) were current users of hearing aids. The average years of hearing aid use based on the case history was 11.27 years (SD=9.8). Because prior experience with amplification has very little effect on speech perception, the listeners who wore hearing aids were grouped with HI listeners who did not wear aids (review, Turner et al., 1996). The table in Appendix E gives the details of each subject who participated.



**Figure 10. Mean and individual audiograms for the subjects in the normal hearing group (NH).**



**Figure 11. Mean and individual audiograms for the subjects in the hearing impaired group (HI).**

## Stimuli

The audio version of the Connected Speech Test (CST; Cox et al., 1987, 1988; Appendix F) was used for both acoustic and behavioral testing. The CST was developed to simulate real world listening situations more closely than other speech tests. The speech material is composed of passages of speech, with 9 or 10 sentences in each passage, about a familiar topic. The listener is informed of the topic prior to sentence presentation, similar to real world situations. The stimuli were transferred from the compact disc to the computer hard drive at a 44.1kHz sampling rate using Adobe Audition. The CST consists of a female speaker on one channel and 6-talker babble on the other channel. For behavioral testing, which used all practice and test passages, the

long-term average spectra of the speech and babble were equated within 7dB of each other from 200Hz – 10kHz. For acoustic testing, a 1-minute sample of both stimuli was extracted. The silent portions between sentences were removed and the long-term average spectra were matched between the speech and babble samples.

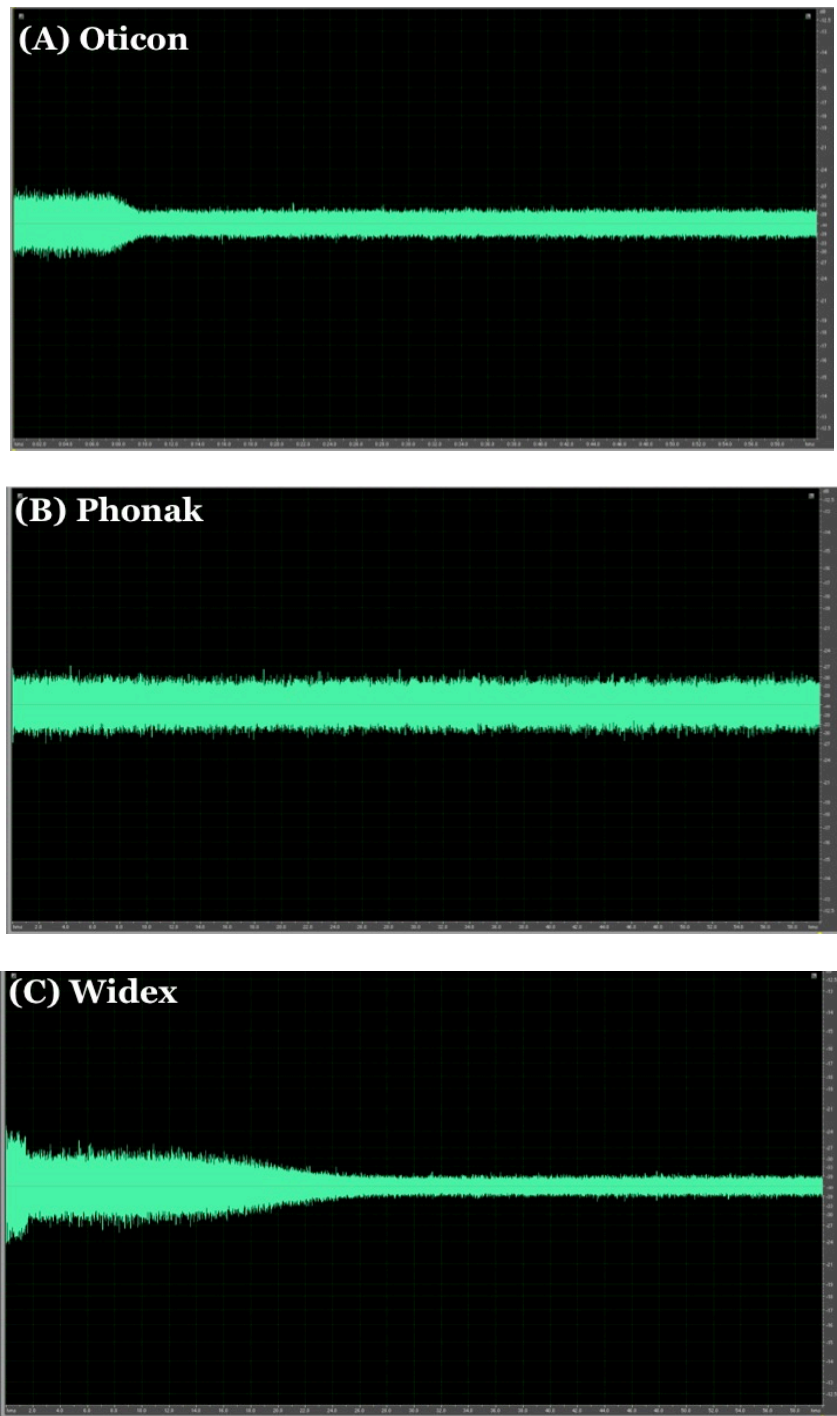
Speech and babble signals were routed from the computer to a Crown D-45 amplifier, then presented from a single loudspeaker at 0 degrees azimuth in a double-walled audiometric test booth. The speech was always presented at an average conversational level of 65 dB SPL and the noise was varied to the desired SNR (described later). The signals were calibrated using the respective calibration noises from the CST (filtered random noise with the 1/3-octave band long-term levels corresponding to test stimuli).

The CST is scored by correct or incorrect responses to 1-5 key words within each sentence. Within a topic passage, there are 25 key words. Two passages are paired together to achieve equal intelligibility for both NH and HI listeners, when compared to any other pair of passages (Cox et al., 1987; 1988). Two passage pairs were given in each condition. Verbal responses were scored live and verified with the recorded versions of the test session, as needed.

### **3.3. Hearing Aids**

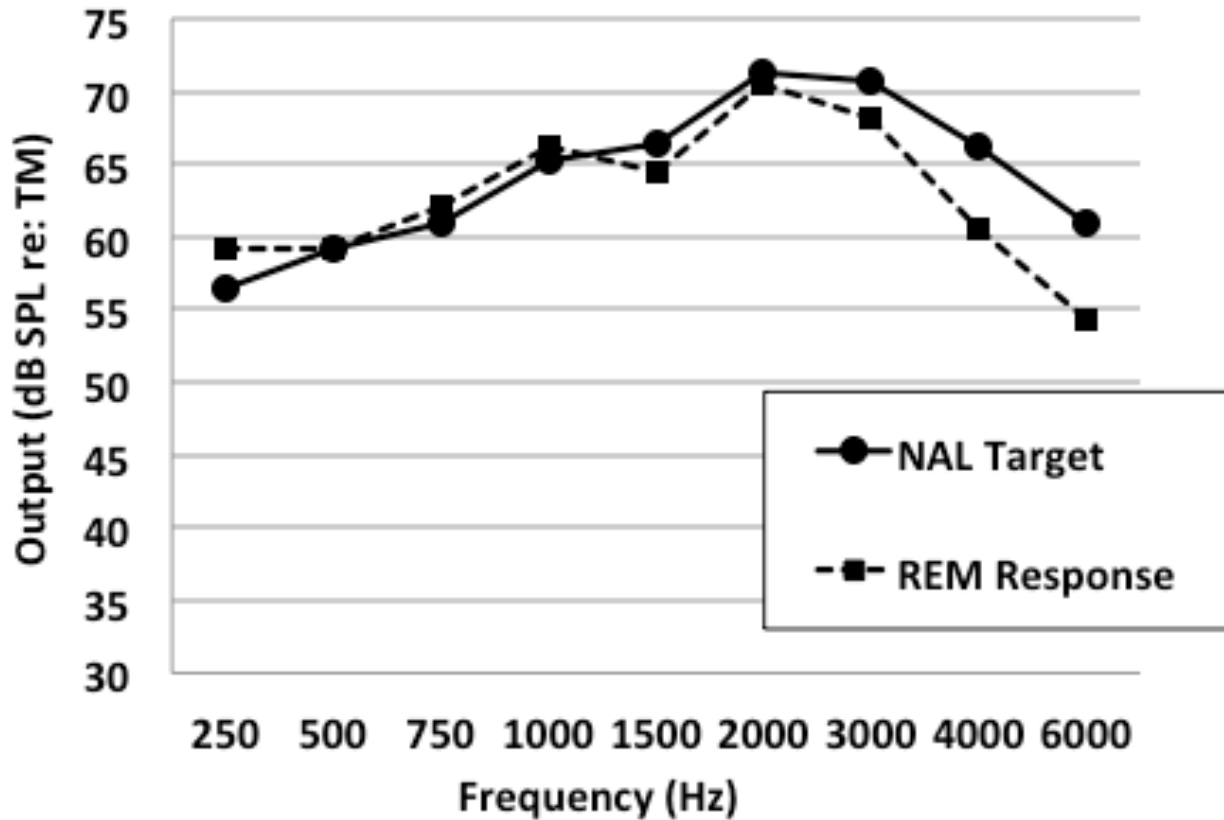
The same three behind-the-ear hearing aids were used for all 50 subjects. Different manufacturers made the hearing aids, and we attempted to choose the most current models of aids we had available in our facility. The hearing aids used were: Oticon Acto Pro (8 channels, 118 dB SPL maximum output, 51 dB maximum gain), Phonak Ambra (16 channels, 129 dB SPL maximum output, 60 dB maximum gain) and Widex Mind 440 (15 channels, 124 dB SPL maximum output, 56 dB maximum gain). All

aids could be programmed for linear or compression processing, and had noise reduction strategies. As discussed previously, manufacturers implement algorithms differently; therefore, it was possible that the implementation of a noise reduction or WDRC algorithm in one brand of hearing aid would create different results than an algorithm implemented by another brand. One way implementation of WDRC algorithms often varies between manufacturers is in how fast the compression occurs. The Oticon Acto Pro changes compression characteristics, including attack and release time, based on the “identity” you (or the software) choose for that individual. The time constants range from 75 ms to 1000 ms, depending on the identify (Oticon,2009). We chose the middle identity, “active,” for every subject. The Phonak Ambra uses an attack time of 1 ms and a release time of 20 ms (Phonak, 2010). The Widex Mind 400 uses adaptive time constants, with attack times ranging from 10 ms to 2 seconds and release times ranging from 10 ms to 20 seconds, which also vary between channels (Widex, 2013). To better understand how the noise reduction algorithms behaved in this study, we recorded the output of each noise reduction algorithm with linear processing on KEMAR using white noise at 85 dB SPL (Figure 12). The activation time was measured as the time it took for the output to stabilize within 3 dB of the steady-state output (Bentler and Chiou, 2006). The onset times were 8.54, 0.04, and 22.66 seconds for the Oticon, Phonak, and Widex aids, respectively. The gain reduction was measured by comparing the average levels pre- and post-onset of each noise reduction algorithm. Average gain reduction was 6.28, 3.48, and 9.63 for the Oticon, Phonak, and Widex aids, respectively.

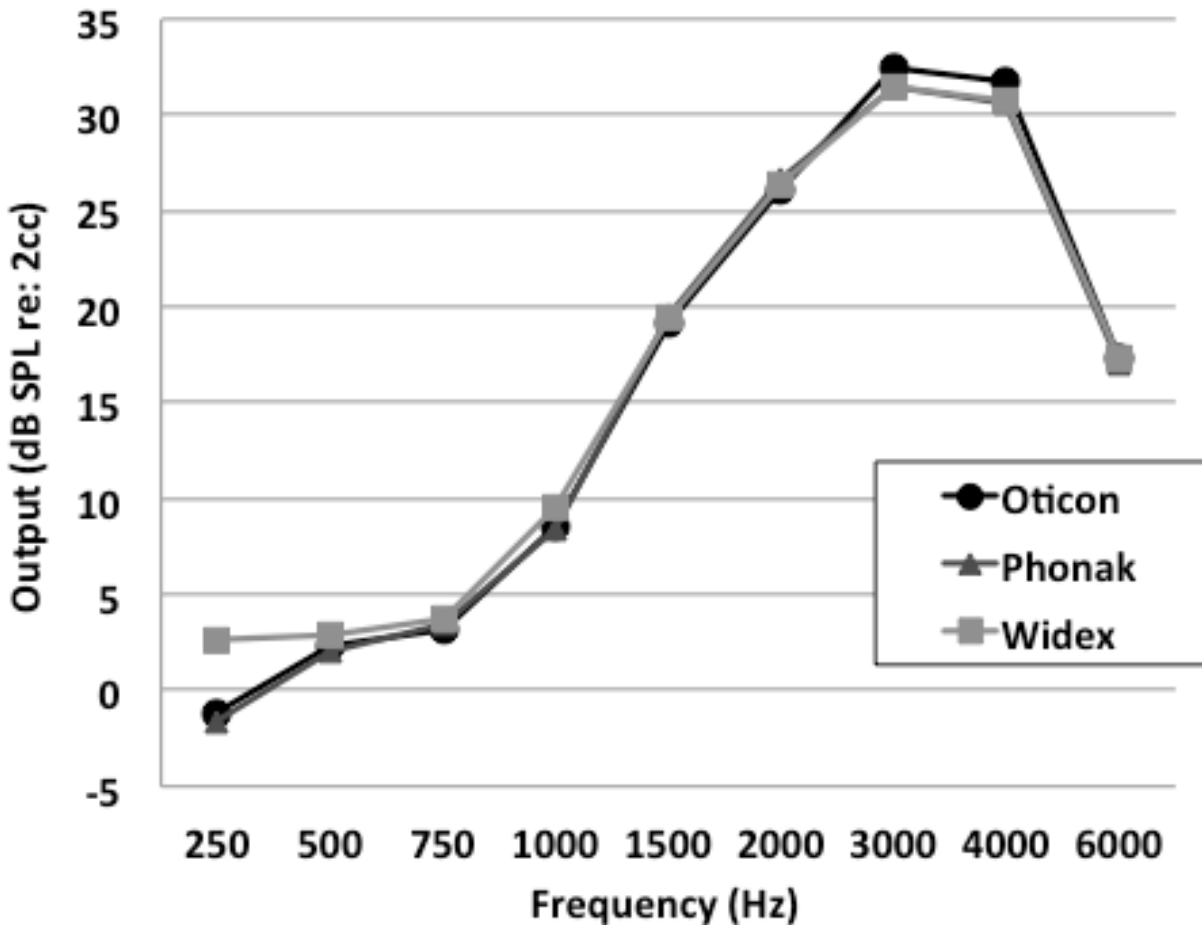


**Figure 12. Activation times and amount of gain reduction provided by the Oticon (A), Phonak (B), and Widex (C) noise reduction algorithms programmed for a flat 50 dB HL audiogram. The stimulus was an 85 dB SPL white noise signal.**

The hearing aids were programmed to NAL-NL1 targets for all participants (regardless of NH or HI group status), using individual threshold levels for their better hearing ear and real ear measures. The Frye FP35 was used for real ear measures and coupler matching of a digital speech signal to a 65 dB SPL NAL-NL1 target. The procedure of programming the aids were as follows: 1) enter audiogram and date of birth into manufacturer's software and request NAL-NL1 programming, if available; 2) enter audiogram into Frye test box and generate NAL-NL1 targets for a monaural fitting; 3) one randomly chosen aid among the three was coupled to a comply tip; 4) program the first aid using real ear measures to match the target for a 65 dB SPL signal as close as possible in the WDRC program; 5) run 90 dB SPL pure tone sweep to compare output levels to predicted uncomfortable levels, and subjectively verify the aid is not too loud; 6) measure the real ear response of the linear program and adjust to match to WDRC response; 7) put the first aid into a test box in a 2cc coupler, and match other two aids to the frequency response of aid one using the same digital speech signal at 65 dB SPL. All aids consisted of four programs: LIN, LIN+NR, WDRC, WDRC+NR. These were programmed in the same memory position for all aids, but randomly chosen for order of presentation to each subject. Feedback cancellation was disabled, unless an adequate match to target could not be obtained without it, which occurred with four subjects. In these cases, the feedback canceller was set as mild as possible for behavioral testing, and disabled again for acoustic testing. Figure 13 shows the average (across subjects) results of the match to NAL-NL1 target and Figure 14 shows the average (across subjects) matched output levels in the 2cc coupler for all three aids. If sounds were uncomfortable, the MPO was reduced until comfort was achieved. In reality this was not a problem for any subject throughout the testing.



**Figure 13. Mean match to NAL-NL1 target achieved by real ear measures for a 65 dB SPL digital speech signal across all 50 subjects. The match between the target and real ear response was within 5 dB from 250-3000 Hz, but deviated by a greater amount in the high frequencies as often seen with hearing aid output levels.**



**Figure 14. Mean match between all aids for a 65 dB SPL digital speech signal in a 2cc coupler, showing close output levels between the aids.**

### **3.4. Procedure**

#### **Behavioral Testing**

The participants were seated in the center of a double-walled, 8 foot x 8.8 foot, sound-treated booth. Both speech and babble were presented from a speaker at 0 degrees azimuth, vertical and horizontal, at 1 meter from the subject's head. A microphone was placed near the listener's mouth to ensure adequate audibility of verbal responses. Another microphone and digital voice recorder were placed around the listener's neck to record responses. An earplug was placed in the subject's non-test ear.

The speech was presented at 65 dB SPL, as measured at the location of the center of the subject's head with a sound level meter without the subject present. The SNR of the stimuli was chosen individually for each listener, at approximately their 50% correct level. Previous research suggests that relative performance within a subject does not change whether performance is measured at the SNR for 29.3%, 50%, or 70.4% correct (Dirks, Morgan, and Dubno, 1982); therefore, we chose the most clinically applicable measure. The SNR-50 was measured in two stages. One hearing aid was chosen randomly and set to the linear program. First, the listening passages of the CST were used to grossly estimate the level where the listener scored 50% correct (SNR-50). The order of the listening passages was randomized for each subject. The listening passages started at an SNR of +15dB and dropped to an SNR of -10 dB. One list was presented at a favorable SNR (+15 to +10 dB), and then reduced in SNR until the listener started missing a majority of words. On average, each listener heard 4-5 listening passages. Second, the four CST practice pairs were used to fine-tune the SNR-50 level. The first pair was presented at a level estimated to be their SNR-50 based on the listening passages. Subsequent levels were presented at better and poorer SNRs to find the point where approximately 50% of the words were understood. If a choice had to be made, the SNR where the listener scored better than 50% was chosen. For example, if a listener understood 28/50 words at 0 dB SNR, and 22/50 words at -1 dB SNR, then the 0 dB SNR level was chosen for testing. The purpose of this procedure was to avoid hitting the floor or ceiling of an individual's performance as the SNR was modified by the hearing aid.

The hearing aid test conditions were performed with the input SNR at the individual's approximate SNR-50. The hearing aid brands (Oticon, Phonak, Widex)

were blocked and randomly presented, and the conditions (LIN, LIN + NR, WDRC, WDRC +NR) were randomized within each block, for each listener. The listener heard two pairs of CST lists in each hearing aid condition. The order of CST test sentences was randomized for each listener (sample in Appendix B). The average was taken between the two pairs of CST lists in each condition, which should be sensitive enough to identify a 1-2dB SNR difference in speech perception (Cox et al., 1987, 1988). The listener was encouraged to walk around between each aid tested (every 1/2 hour).

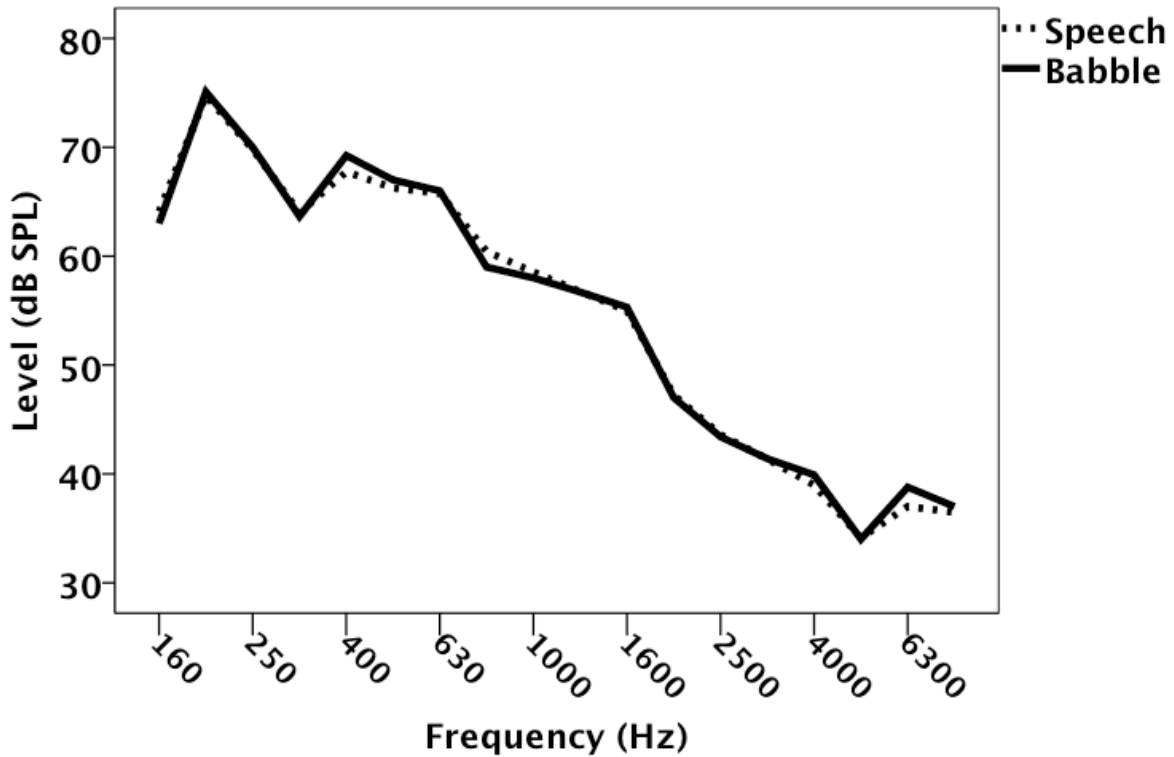
### **Acoustic Testing**

To obtain the SNR at the hearing aid output, a modified version of Hagerman's phase-inversion technique was used to separate the speech and babble signals at the output of the hearing aid. Four versions of the CST stimuli were generated for this technique, which vary in phase only:

- 1) Original-phase speech + original-phase noise (+Sp +N)
- 2) Phase-inverted speech + original-phase noise (-Sp +N)
- 3) Original-phase speech + phase-inverted noise (+Sp -N)
- 4) Phase-inverted speech + phase-inverted noise (-Sp -N)

The original description of this phase-inversion technique given by Hagerman & Oloffsson (2004) used only two versions of stimuli (equations 1-2 above). Souza, Jenstad, and Boike (2006) included equation 3 to add to the stability of the measurement (Jenstad; personal communication). We included equation 4 to allow us to measure a type of error produced when using this technique with non-linear processing (described later). A one-minute segment of the CST speech and babble was used for acoustic testing, which was generated from two passages (topics: vegetables

and owls) with the silent portions between sentences removed. The first 30 seconds were used to allow the hearing aid algorithms time to “settle,” while the next 30 seconds were use for the analysis. The spectrum of the babble was filtered to match the spectrum of the speech for the one-minute sample from 160 to 6300 Hz (Figure 15).

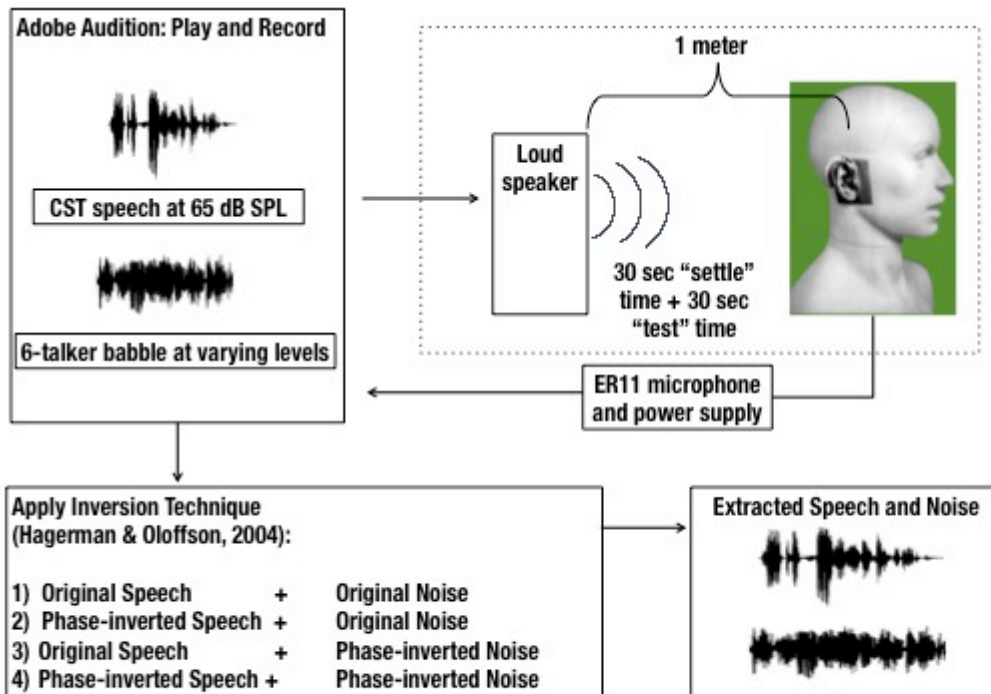


**Figure 15. Long-term average levels of speech and babble stimuli used for acoustic testing of aids. A close match in output levels was achieved for all frequencies tested.**

### *Recording of Stimuli*

Because Hagerman’s phase-inversion technique requires several identical recordings (except for the phase of the input signal) to be made sequentially, real ear measures were ruled out for this procedure. After the subject left, each hearing aid was placed on a mannikin’s left ear with a comply tip. The manikin used was the Knowles

Electronic Manikin for Acoustic Research (KEMAR; Knowles Electronics, 1972), which is commonly used in hearing aid research for taking in-situ measurements. KEMAR has average measurements for a human person and a simulated pinna, outer ear canal, and a Zwislocki coupler for outer ear simulations. Differences between output responses measured on KEMAR and those measured on real ears are constant across input level and frequencies (i.e., a linear transformation). In other words, a change in SNR measured on a real ear would give the same change in SNR measured on KEMAR. KEMAR was situated in the same position as the subjects during speech perception testing. The output of the Zwislocki coupler in KEMAR's ear was connected to an Etymotic Research ER-11 1/2 microphone system, which was routed back to the computer. Adobe Audition was used to present the stimuli from the loudspeaker in the booth, and simultaneously record the output of the hearing aid from KEMAR's ear (see Figure 16 for design).



**Figure 16. Set up for acoustic recording of hearing aid output. Adobe audition was used to play and record sounds to KEMAR’s ear, and to implement the phase-inversion technique for speech and noise extraction at the output of the hearing aid.**

*Extraction of Speech and Noise Signals*

In each hearing aid condition, the recordings were added in Adobe Audition to extract the speech or the babble as follows:

- 1) extracted noise = (+Sp +N) + (-Sp +N)
- 2) extracted speech = (+Sp +N) + (+Sp -N)

Subsequent analysis on the extracted speech and noise signals was done using a Matlab program created for this study (Lewis, 2012). Each speech and noise file was bandpass filtered using a bank of 18 1/3-octave band filters, according to the approximate frequencies used in the ANSI S3.5-1997 SII standard, from 160-6300 Hz. The root

mean square (RMS) levels in each band, and across bands, were calculated using 120 ms time windows from which the SNR was calculated in each condition. The wideband SNR was used as the SNR value at the output of the hearing aid. Then the narrowband levels were passed through an implementation of the SII procedure to yield the SII value for each condition.

### **Stability of Hagerman's phase-inversion technique**

Hagerman's phase-inversion technique was designed for use with linear or quasi-linear systems. Due to the use of this technique with non-linear processing in this study, it's important to quantify the integrity of the measures and account for any "error" generated from the inversion technique itself. To quantify the error, the 180 degree inverted recordings are combined with the 0 degree phase recordings to obtain a "residual noise" estimate from using the Hagerman technique:

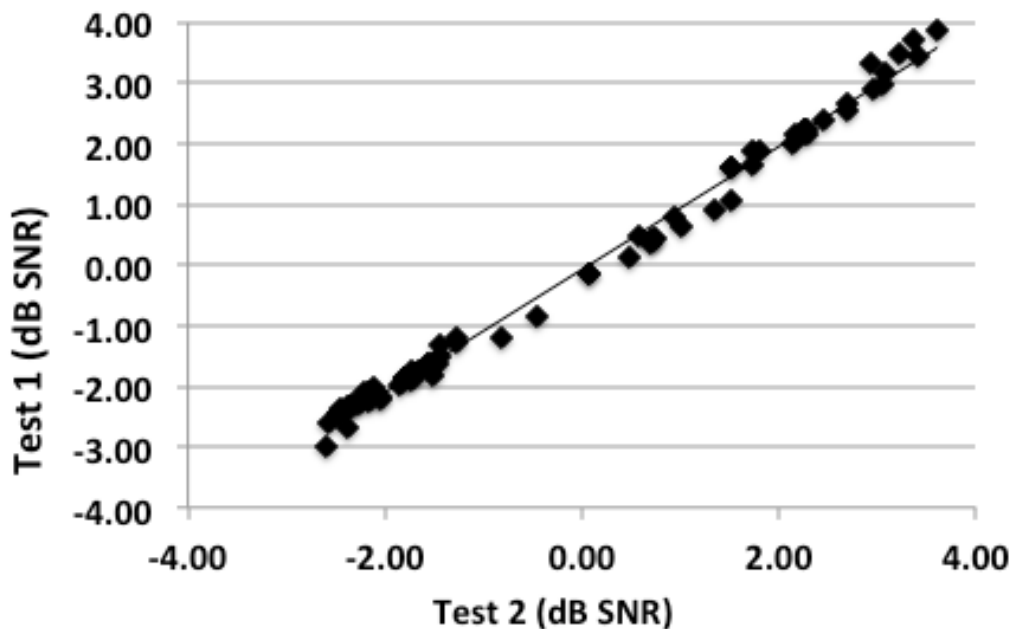
$$1) \text{ error} = (+Sp + N) + (-Sp - N)$$

The residual noise should be well below the softest level of the extracted speech or noise signal (Naylor & Johannesson, 2010). One way to estimate the difference between the residual noise and the extracted signal of interest is through measuring the amount of attenuation obtained (Hartling et al., 2012). We measured the long-term RMS level of the speech, babble, and error signals using 120 ms windows, averaged over the second 30-seconds of the recording. The difference between the level of the error and the level of the speech and noise was calculated. As discussed in the introduction, previous results using these techniques have shown attenuation values of 20 dB or greater as

providing sufficient separation of the extracted signals (Naylor & Johannesson, 2009; Jenstad & Zakis, 2011; Hartling et al., 2012). Therefore, a criterion attenuation value was adopted to evaluate the stability of the measurements. If the difference was less than 20 dB, then the recordings for that condition were considered unreliable, and were rechecked for errors. If an error was made in the extraction, it was corrected. If not, the recording was made again and the levels were re-extracted. Following this procedure, if a recording did not pass the criterion attenuation value of 20 dB, then the recording and results were discarded from further analysis.

### **Reliability of Hagerman's Technique**

To verify the reliability of using Hagerman's separation technique with non-linear processing, test re-test reliability of the SNR measurement was performed on 6 subjects (3 NH and 3 HI) with the highest average attenuation values. Test retest reliability for SNR quantification is shown in Figure 17 and the correlation was found to be high ( $r=.99$ ).

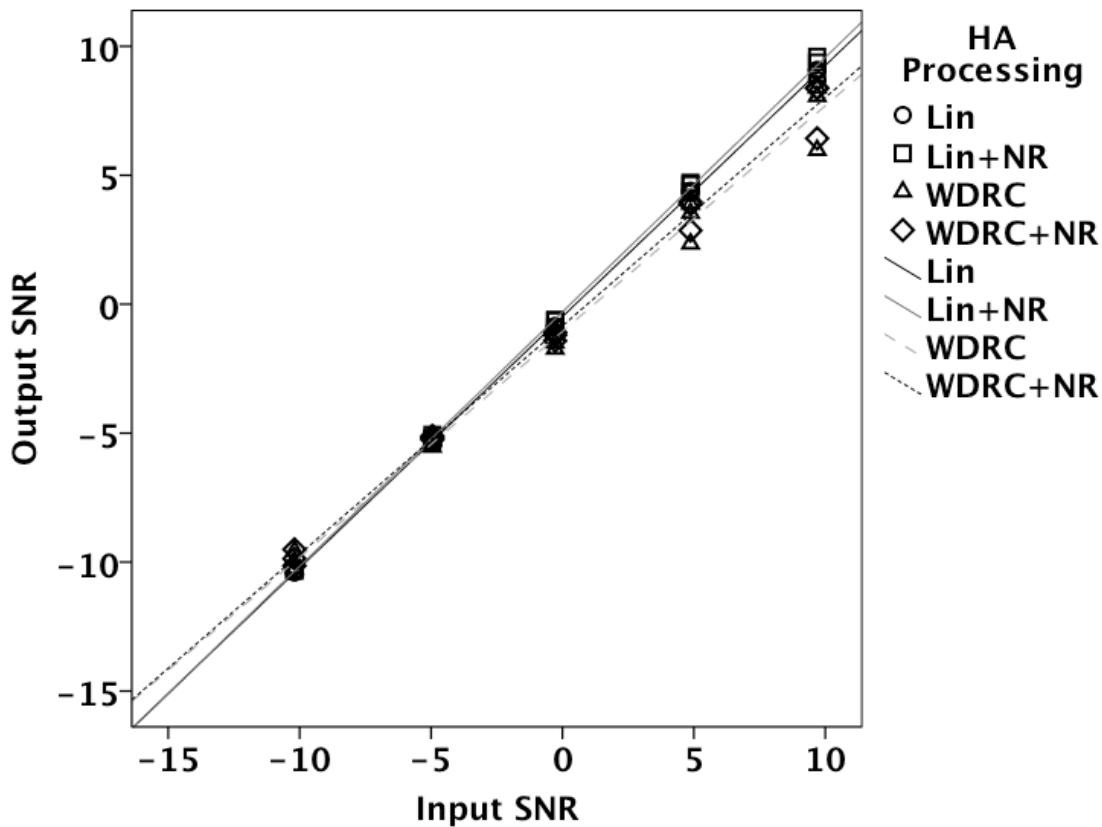


**Figure 17. High test-retest reliability ( $r=0.99$ ) of SNR measured post-Hagerman inversion was found for six subjects who were retested on a different day.**

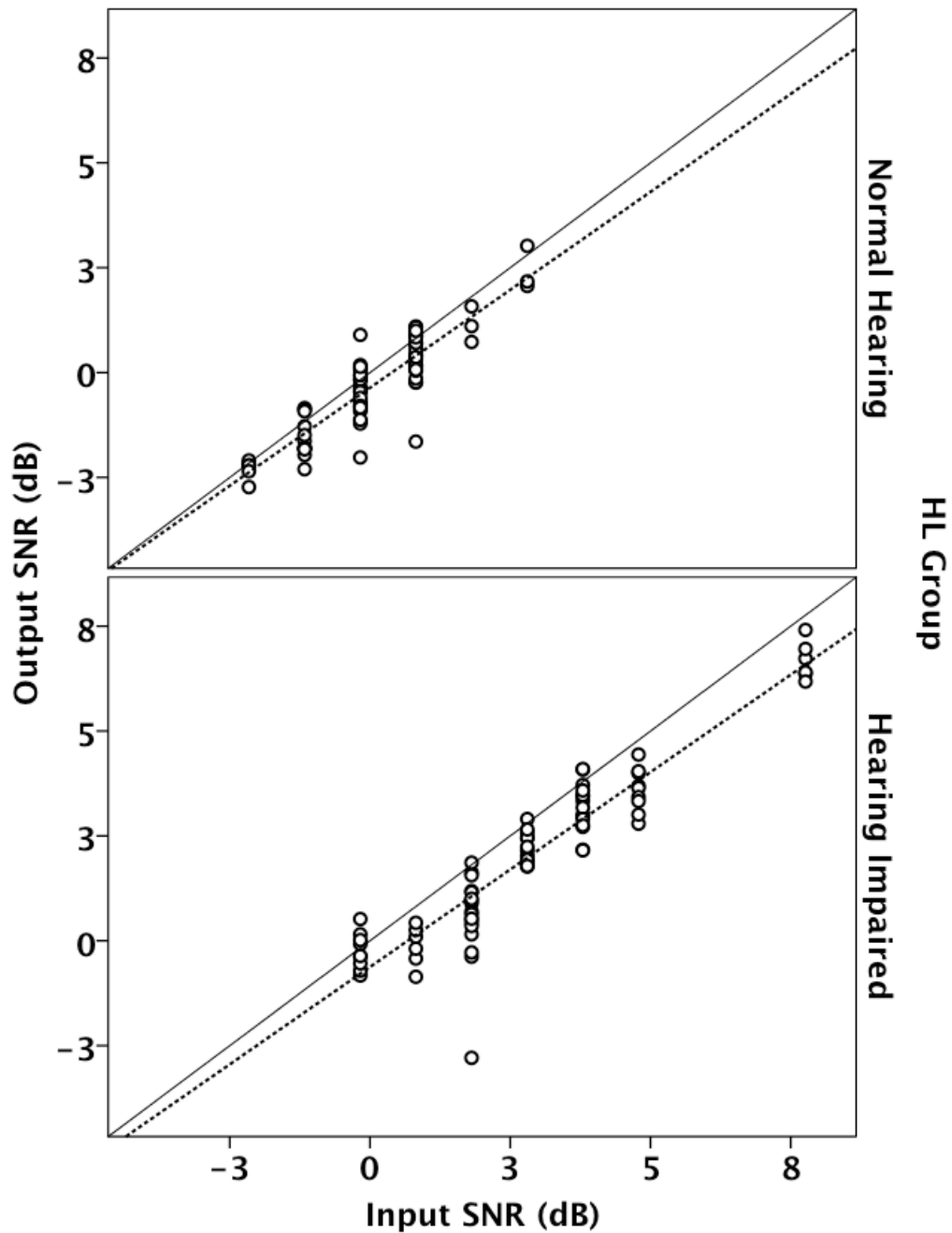
### **Validity of SNR Calculation post-Hagerman Processing**

One way to evaluate the validity of this procedure is to measure the SNR in the linear condition, with the expectation that the input SNR will equal the output SNR. During pilot testing, we programmed each aid for a flat 50 dB HL audiogram using linear processing. Using the Hagerman extraction technique, and quantification of SNR, our pilot results were plotted as an input-output SNR graph (Figure 18, dark solid line). A linear regression analysis showed  $R^2$  values for the conditions of Lin, Lin+NR, WDRC, and WDRC+NR of 1.00, .999, .991, and .994, respectively, as expected. However, after the experiment, when we plotted the linear results for all of our subjects, the regression line deviates from linear (Figure 19, dotted line). The linear regression fit

for the NH and HI groups were  $R^2 = .797$  and  $.872$ , respectively. It's possible that those aids with higher internal noise result in a greater deviation from linear. Although our ANSI (2003) testing reported equivalent input noise levels between 17-26 dB, the level of the internal noise will vary depending on the hardware of the aid, gain settings, input level, and SNR (Lewis et al., 2010). Therefore, it's difficult to specifically predict which conditions will result in higher internal noise settings than others.



**Figure 18. Input-Output SNR graph for all conditions with aids programmed to a flat 50 dB HL audiogram during pilot testing. Linear regression analysis showed  $R^2$  values for the conditions of Lin, Lin+NR, WDRC, and WDRC+NR of 1.00, .999, .991, and .994, respectively.**



**Figure 19. Linear conditions for all hearing aids and all subjects. The solid black line is the linear reference line, while the dotted line is the regression line for the data. The correlation for the NH and HI group was .89 and .93, respectively.**

## Chapter 4. Results

### 4.1 Efficacy of Hagerman Separation

First, the efficacy of the Hagerman separation was analyzed for each condition by assessing the error levels and attenuation values. Figures in Appendix G plot the extracted speech and noise levels compared to the error levels (extracted silence) across frequency for the NH and HI groups (collapsed across subject). The figures in Appendix G show that the error in the high frequencies is greater than the error generated in the low frequencies. This is likely due to softer input sound pressure levels and poorer hearing thresholds in the high frequencies causing greater gain requirements. In turn, this will increase the level of the noise floor in that frequency region.

**NH Attenuation Values**

|                 |         | HA Brand |        |        |        |       |        |
|-----------------|---------|----------|--------|--------|--------|-------|--------|
|                 |         | Oticon   |        | Phonak |        | Widex |        |
|                 |         | Noise    | Speech | Noise  | Speech | Noise | Speech |
| HA<br>Condition | LIN     | 24.68    | 22.57  | 26.11  | 24.29  | 29.76 | 28.07  |
|                 | LIN+NR  | 30.18    | 28.04  | 28.24  | 26.54  | 30.49 | 28.79  |
|                 | WDRC    | 30.71    | 28.46  | 28.33  | 26.21  | 30.72 | 28.82  |
|                 | WDRC+NR | 30.61    | 28.58  | 28.27  | 26.27  | 30.35 | 28.48  |

**Table 1. Mean attenuation values for the NH group.**

**HI Attenuation Values**

|              |         | HA Brand |        |        |        |       |        |
|--------------|---------|----------|--------|--------|--------|-------|--------|
|              |         | Oticon   |        | Phonak |        | Widex |        |
|              |         | Noise    | Speech | Noise  | Speech | Noise | Speech |
| HA Condition | LIN     | 23.27    | 23.76  | 23.84  | 24.61  | 27.98 | 28.94  |
|              | LIN+NR  | 27.75    | 28.40  | 24.88  | 25.81  | 28.51 | 29.49  |
|              | WDRC    | 28.03    | 28.42  | 25.36  | 25.17  | 28.43 | 29.12  |
|              | WDRC+NR | 28.12    | 28.51  | 24.95  | 24.90  | 28.49 | 29.23  |

**Table 2. Mean attenuation values for the HI group.**

The level of the error was subtracted from the levels of the extracted speech and noise to obtain the amount of “attenuation” achieved (Appendix H). The mean attenuation across frequency is shown in Tables 1 and 2. We used an attenuation value of 20dB as our criterion for adequate separation of speech and noise at the output of the hearing aid, which is comparable to those reported by other researchers who found attenuation values of 20 dB or more (Naylor & Johannesson, 2009; Jenstad & Zakis, 2011; Hartling et al., 2012). The attenuation was measured in each condition for each subject and compared to the criterion of 20dB. Twenty-one conditions (out of 600 analyzed) did not pass this inspection and were re-checked. Of those, the incorrect sound files were combined when extracting the error using the Hagerman inversion technique for four conditions. After combining the correct files, the attenuation passed the 20 dB criterion. For seven conditions, the Oticon Linear condition was not passing the criterion attenuation amount, even after re-checking the analysis, meaning that the level of error was too high, which may contaminate the extracted speech and noise signals. If we extracted the error files by a different combination of recordings (sp+n- and sp-n+, instead of sp+n+ and sp-n- as in the original analysis), the error improved by about 5 dB. These results suggest that perhaps the Oticon aid, in the linear condition,

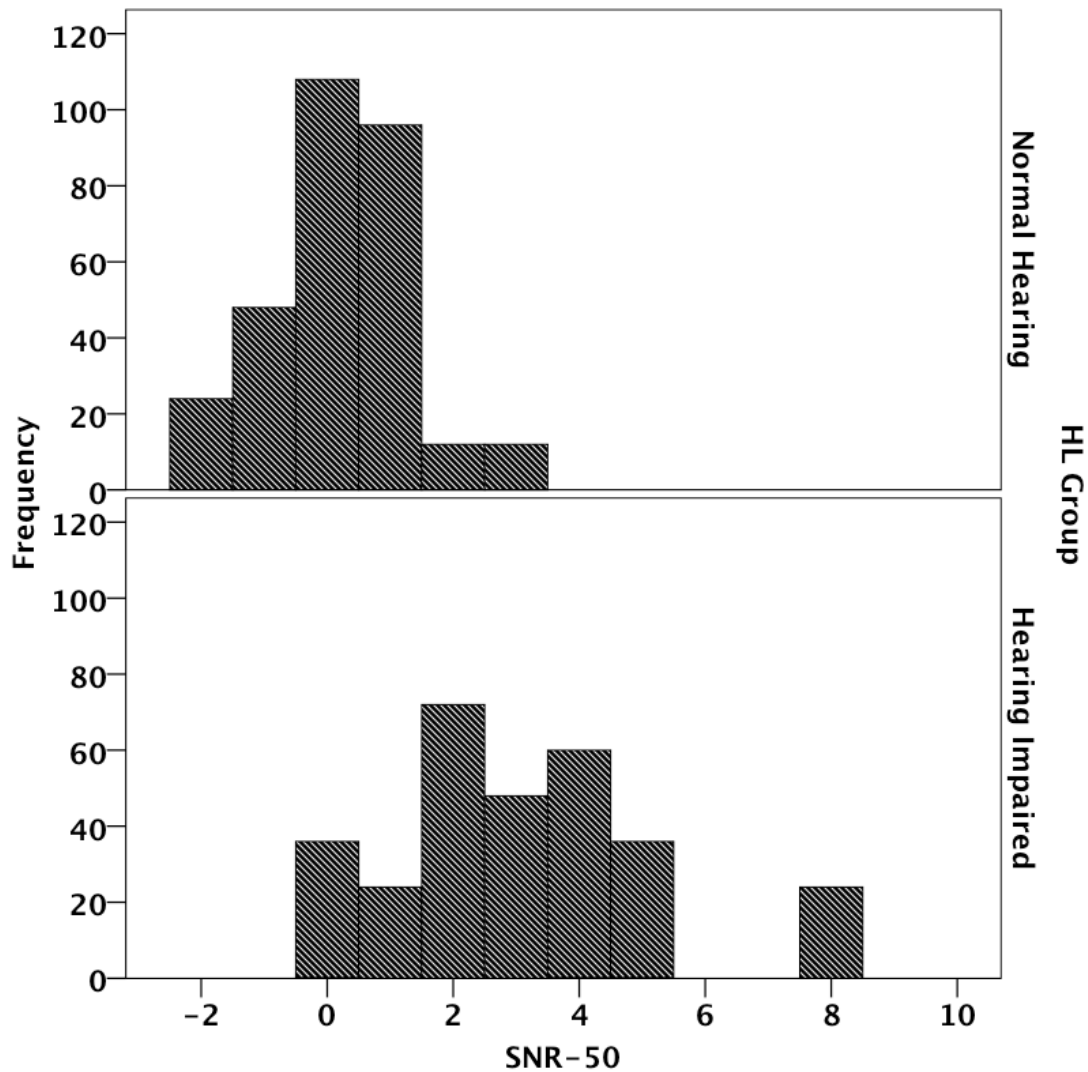
was not reacting to the signals the same way every time (e.g., time variant processing), which is an assumption of using Hagerman's phase-inversion technique. Six of the seven conditions passed the criterion when we used these error levels instead. The remaining one was discarded (subject 51, condition Oticon Linear). The final twelve conditions that did not pass were all from one subject (subject 3), who had one of the worse hearing losses of the group, and had a reverse slope nature to it. Even after re-recording these conditions, the attenuation did not improve. Therefore, her output SNR data were discarded from the remainder of the analysis.

The results from Hartling et al. (2010) showed instability such as this in some of their conditions also, where some aid models and conditions had higher attenuation values than others. Therefore, it's likely that specific aid conditions will result in variations of attenuation, and that this must be measured for each case. Further work in this area needs to be done to understand the interactions and consequences of instability in processing. Upon further testing, it was found that the Oticon aid was always in fixed directional microphone mode at certain frequencies, even when the manufacturer's software said it was in omni-directional mode. While this was unlikely to influence our results given that speech and noise were presented from 0 degrees azimuth, it suggests that there may be algorithms running in the aid regardless of what is shown in the fitting software.

#### **4.2. To what extent do hearing aid algorithms change the SNR when programmed for an individual?**

Subjects were tested at different input SNRs, based on their approximate SNR-50.

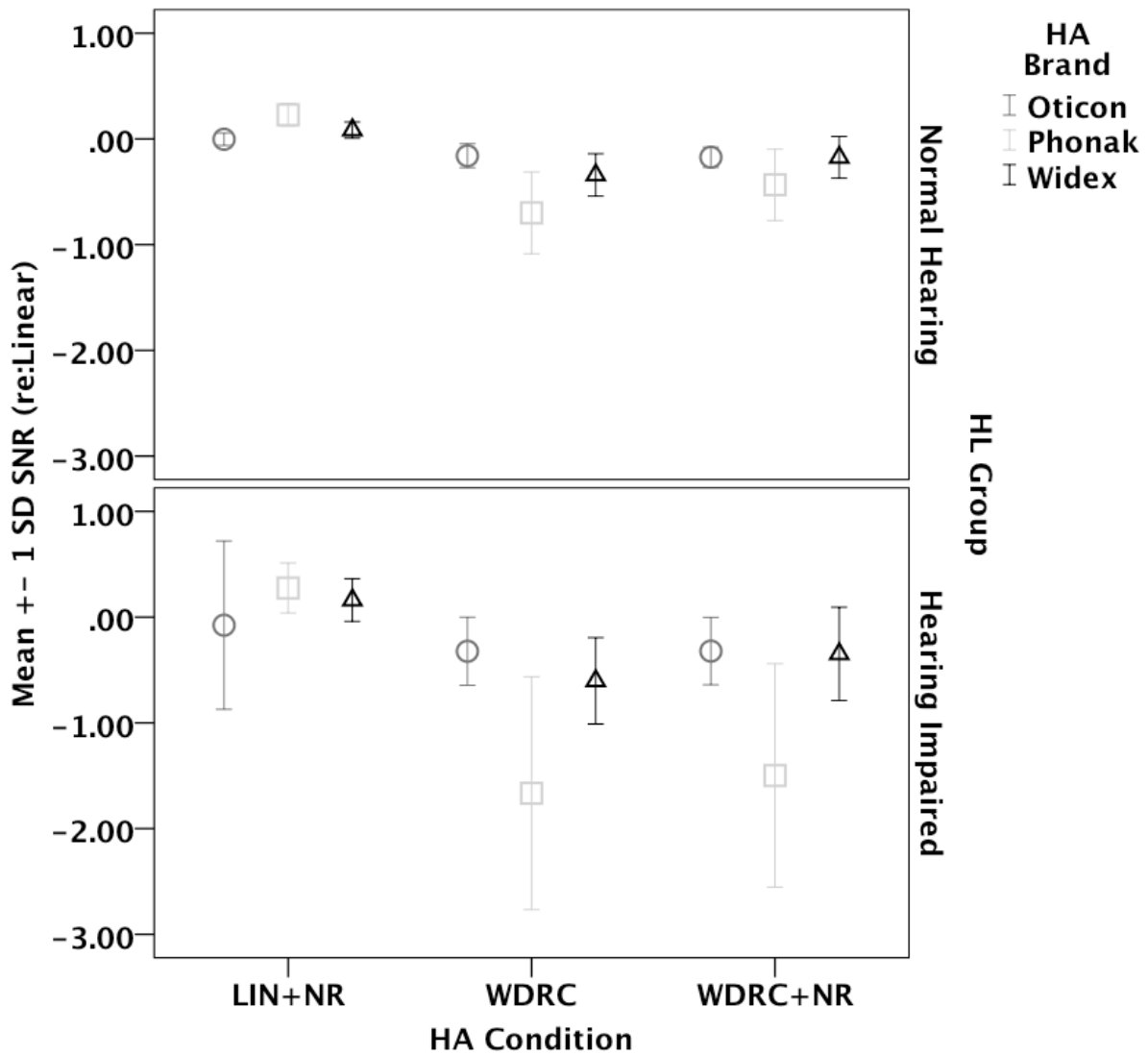
Figure 20 shows the distribution of SNR-50 results for each group.



**Figure 20. Distribution of input SNRs (SNR-50) in both groups.**

Due to the differences in input SNR between subjects, we referenced the extracted speech and noise levels to those for the linear condition (re: LIN) for each HA brand and subject, for all plots and data analysis, unless otherwise specified.

To answer our first question of whether hearing aid processing modifies the SNR from the input to the output of the hearing aid, we plotted the mean change in SNR (re: LIN) at the output of the aid (Figure 21; Table 3). Individual data is plotted for each condition (Appendix I).



**Figure 21. Mean change in SNR (re: LIN).**

|              |         | HL Group       |        |       |                  |        |       |
|--------------|---------|----------------|--------|-------|------------------|--------|-------|
|              |         | Normal Hearing |        |       | Hearing Impaired |        |       |
|              |         | Oticon         | Phonak | Widex | Oticon           | Phonak | Widex |
| HA Condition | LIN+NR  | .00            | .23    | .09   | -1.10            | .25    | .20   |
|              | WDRC    | -.14           | -.70   | -.34  | -1.32            | -1.75  | -.63  |
|              | WDRC+NR | -.17           | -.43   | -.17  | -1.32            | -1.57  | -.35  |

**Table 3. Mean change in SNR (re: LIN).**

To test whether changes in SNR were statistically different, a mixed design, repeated-measures ANOVA was performed with a between group variable of hearing status, and within group variables of HA brand and condition. A between group variable of hearing status was included to assess whether the effects of SNR change were due to gain or compression setting differences. Mauchly's test of sphericity suggested that the variance between the two groups was unequal; therefore transformations of the data were explored. No transformation lead to normally distributed results, so ANOVAs were conducted within each group separately.

A two-way, repeated-measures ANOVA was performed within each group to test whether differences in SNR (re: LIN) existed between brand (Oticon, Phonak, or Widex), or type of processing (LIN+NR, WDRC, WDRC+NR). For the NH group, Mauchly's test of sphericity indicates that the assumption of sphericity had been violated for type of processing ( $X^2(2)=53.966$ ,  $p<.0001$ ) and brand \* type of processing ( $X^2(9)= 98.357$ ,  $p<.0001$ ) of this model are not met, therefore the degrees of freedom were corrected using the Greenhouse-Geisser estimates of sphericity. All main and interaction effects were reported as significant at  $p<.05$ . The main effect of HA brand ( $F(1.576, 37.828)=12.446$ ,  $p<.0001$ ) was further evaluated using pairwise comparisons with a Bonferroni adjustment for multiple comparisons, and showed that the differences in mean SNR (re: LIN) were significant between Oticon and Phonak ( $p=.001$ ) and between Phonak and Widex ( $p=.009$ ), but not between Oticon and Widex ( $p=.557$ ). The main effect of HA type of processing ( $F(1.050, 25.206)=156.882$ ,  $p<.0001$ ) was also further evaluated using pairwise comparisons with a Bonferroni adjustment for multiple comparisons showed that the difference in mean SNR (re: LIN) between all conditions was significant ( $p<.0001$ ). The interaction term between brand and type of processing

( $F(1.484, 35.626)=51.697, p<.0001$ ) indicates that the type of processing had different effects on the output SNR depending on the brand of the HA. To break down this interaction, contrasts were performed comparing all brands and types of processing to each other, which revealed significant interactions between all comparisons except one ( $p<.005$ ; see Appendix J for details). The interaction suggests that the extent of changes to SNR made by HA processing depends on the HA brand.

For the HI group, a two-way, repeated-measures ANOVA was also used to evaluate differences in SNR (re: LIN) between HA brand (Oticon, Phonak, Widex) and type of processing (LIN+NR, WDRC, WDRC+NR). For the NH group, Mauchly's test of sphericity indicates that the assumption of sphericity had been violated for brand ( $X^2(2)=8.134, p<.017$ ) and brand \* type of processing ( $X^2(9)= 85.514, p<.0001$ ), therefore the degrees of freedom were corrected using the Greenhouse-Geisser estimates of sphericity. All effects were reported as significant at  $p<.05$ . The main effect of HA brand ( $F(1.528, 35.139)=55.925, p<.0001$ ) was further evaluated using pairwise comparisons with a Bonferroni adjustment for multiple comparisons showed that the difference in mean SNR (re: LIN) were significant between Oticon and Phonak ( $p<.001$ ) and between Phonak and Widex ( $p<.001$ ), but not between Oticon and Widex ( $p=.509$ ). The main effect of HA type of processing ( $F(1.933, 44.451)=95.072, p<.0001$ ) was also further evaluated using pairwise comparisons with a Bonferroni adjustment for multiple comparisons showed that the difference in mean SNR (re: LIN) between all types of processing was significant ( $p<.0001$ ), except between WDRC and WDRC+NR. The significant interaction term between brand and type of processing ( $F(1.981, 45.572)=26.879, p<.0001$ ), indicates that type of processing had different effects on the output SNR depending on the brand of the HA. To break down this interaction,

contrasts were performed comparing all brands and types of processing to each other, which revealed that six out of nine comparisons reached statistical significance (see Appendix J). The interaction suggests that the extent of changes in SNR made by HA processing depended on the HA brand.

In summary, the changes in SNR made by hearing aid processing, relative to the linear condition, are between 0.25 and -1.74 dB with a mean change of -0.25. Many significant differences were found between brands and types of processing in both groups, regarding the extent of SNR change made at the output of the aid. While the patterns between groups (Figure 20) seem similar, a 3-way interaction between brand, processing type, and hearing status group was significant ( $F(2,146)=9.290, p<.0001$ ) suggesting that the extent of SNR change depended on hearing status and hearing aid settings.

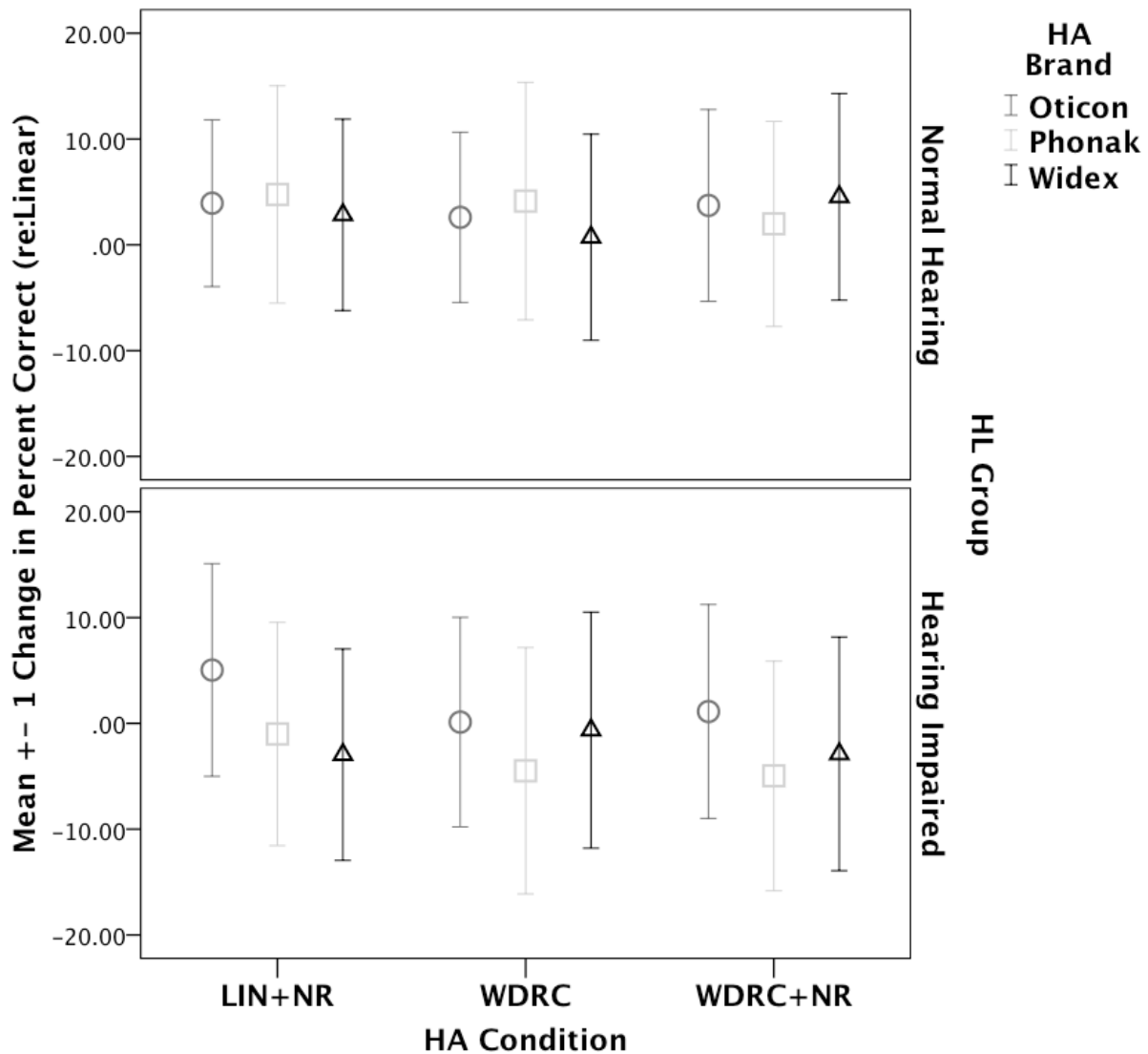
#### **4.3. Does the HA modified-SNR relate to speech perception?**

Our second question was whether changes in SNR at the output of the hearing aid could predict changes in speech perception. Speech perception scores for each group and HA condition are shown in Table 4 (and graphed in Appendix K), while Figure 22 shows the same data referenced to the linear condition (re: LIN). The mean of the NH group (59.8%) was slightly higher than the HI group (47.3%). Percent correct scores were arcsine transformed to stabilize the error variance (Studebaker, 1985). A repeated-measures ANOVA within the NH group revealed no significant results. For the HI group, the ANOVA showed a significant main effect of condition ( $F(2,48)=4.031; p=.024$ ), but the brand and interaction term were not significant. Post-hoc

comparisons revealed that the only significant difference was between the LIN+NR and WDRC + NR condition ( $F(1,24)=9.771$ ;  $p=.005$ ).

|                 |         | HL Group       |        |       |                  |        |       |
|-----------------|---------|----------------|--------|-------|------------------|--------|-------|
|                 |         | Normal Hearing |        |       | Hearing Impaired |        |       |
|                 |         | Oticon         | Phonak | Widex | Oticon           | Phonak | Widex |
| HA<br>Condition | LIN     | 54.84          | 58.51  | 58.76 | 48.48            | 49.84  | 46.24 |
|                 | LIN+NR  | 58.77          | 63.27  | 61.60 | 53.52            | 48.84  | 43.28 |
|                 | WDRC    | 57.44          | 62.64  | 59.48 | 48.60            | 45.36  | 45.60 |
|                 | WDRC+NR | 58.57          | 60.49  | 63.30 | 49.60            | 44.88  | 43.36 |

**Table 4. Mean percent correct scores in each condition.**



**Figure 22. Mean change in percent correct (re: LIN) for NH and HI groups, respectively.**

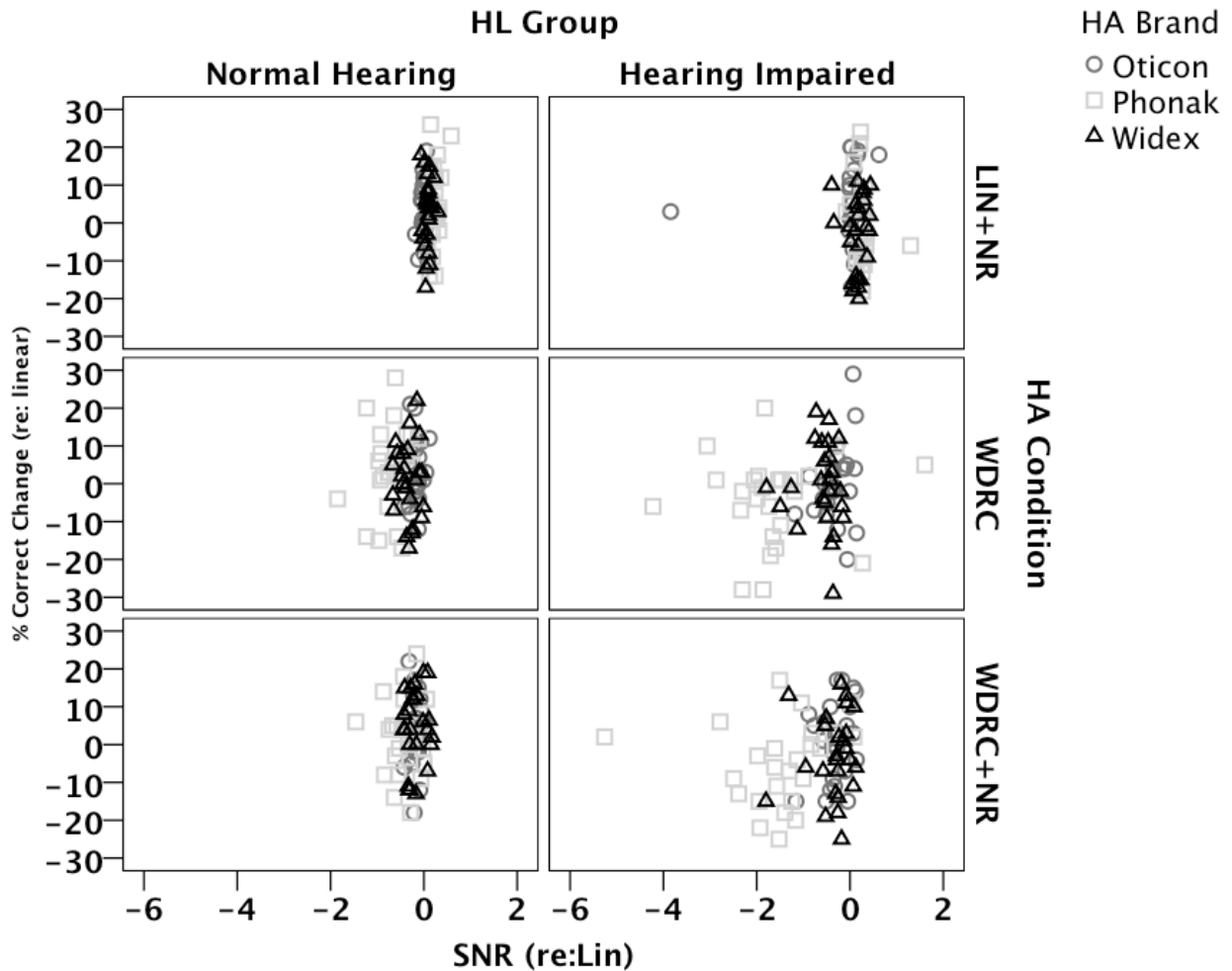
The SNR (re: LIN) was plotted relative to change in speech perception (re: LIN) in Figure 23. To assess whether speech perception changes (re: LIN) could be predicted, a multiple linear regression analysis was performed within each group. The initial model used predictors of primary interest: SNR (re: LIN; continuous variable), HA brand (dummy variable), and HA processing (dummy variable) as predictors. A second model

also included predictors of each subject (coded as dummy variables) to observe any changes to primary predictors when variance within subjects were accounted for in the model.

For the NH group, the first model revealed processing condition as a significant predictor, but SNR and brand were not significant. All processing conditions were significant compared to the LIN condition: LIN+NR ( $t=2.599$ ;  $p=.010$ ), WDRC ( $t=2.369$ ;  $p=.019$ ), and WDRC+NR ( $t=2.941$ ;  $p=.004$ ). The total amount of variance explained from this model was 4.1%. When subjects were added into the model, the significance of condition predicting speech perception changes changed only slightly in that WDRC was no longer significantly different than the LIN condition ( $t=1.813$ ;  $p=.071$ ). The effect of LIN+NR ( $t=2.933$ ;  $p=.004$ ) and WDRC+NR ( $t=2.610$ ;  $p=.010$ ) did not change from the first model. The effect of SNR at the hearing aid output remained insignificant when these other variables were controlled. The total amount of variance in speech perception explained with this model was 21%.

For the HI group, the first model revealed that brand of hearing aid was a significant predictor of speech perception, but that SNR and processing condition were not significant. Both Widex ( $t=-2.435$ ;  $p=.015$ ) and Phonak ( $t=-2.521$ ;  $p=.012$ ) were significantly different from Oticon. The total amount of variance in speech perception scores explained by this model was 5.7%. When subjects were added as predictors into the second model, the significance of primary predictors did not change in that brand stayed significant (Widex:  $p=.009$ ; Phonak:  $p=.008$ ), while the others remained insignificant. The total amount of variance in speech perception explained by this model was 25.4%.

In summary, it appears that changes in speech perceptions scores (re: LIN) were predicted by the type of hearing aid processing for NH subjects and by hearing aid brand for HI subjects. The SNR (re: LIN) was not a significant predictor of speech scores for either group.

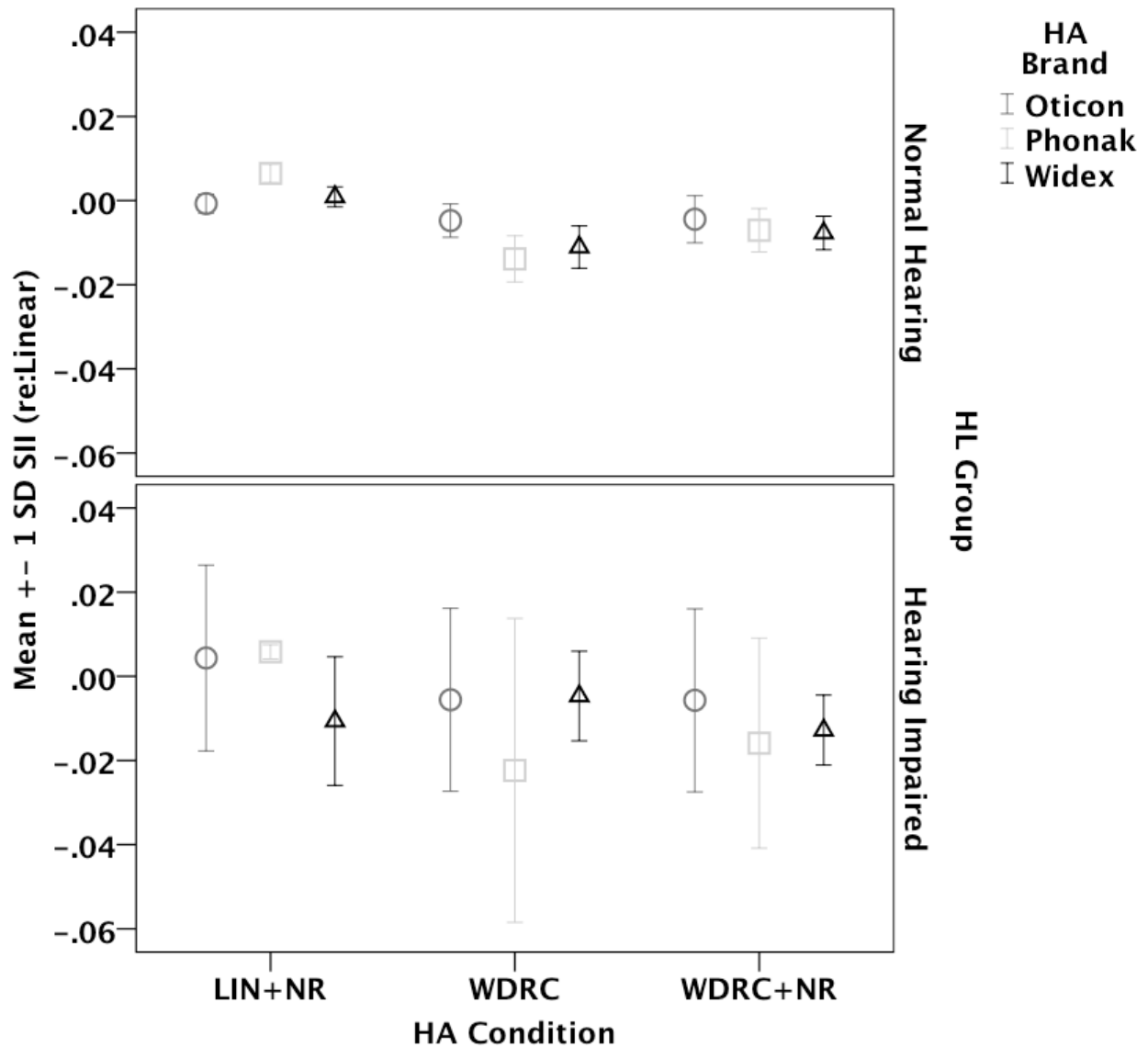


**Figure 23. Relation between SNR (re: LIN) and percent correct (re: LIN).**

**No significant correlations were found in any condition.**

#### **4.4. Role of Audibility**

It's possible that the changes in SNR may or may not be creating changes in audibility, depending on the listener's hearing threshold levels. Therefore, SNR was also calculated using the SII in each condition, which takes into account individual threshold levels, and uses frequency importance weighting (Figure 24 and Table 5). A repeated-measures ANOVA was conducted within each group. For the NH group, the main effects of brand ( $F(2,47) = 5.016$ ;  $p=.011$ ), condition ( $F(1,30) = 192.338$ ;  $p<.0001$ ), and the interaction term ( $F(2,57)=43.855$ ;  $p<.0001$ ) were significant. Paired comparisons were conducted to explore the significant interaction between brand and condition. All comparisons were significant with corrections for multiple comparisons. For the HI group, the main effect of condition ( $F(1.62, 38.868) = 8.896$ ;  $p=.001$ ) and the interaction between brand and condition ( $F(2.252, 54) = 9.877$ ;  $p<.0001$ ) were significant. Paired comparisons were conducted to explore the interaction term (Appendix L), with four out of nine comparison reaching statistical significance. These results suggest that the extent of the effect of processing on SII depends on the HA brand, for both groups.



**Figure 24. SII values in each HA condition.**

|              |         | HL Group       |        |       |                  |        |       |
|--------------|---------|----------------|--------|-------|------------------|--------|-------|
|              |         | Normal Hearing |        |       | Hearing Impaired |        |       |
|              |         | Oticon         | Phonak | Widex | Oticon           | Phonak | Widex |
| HA Condition | LIN     | .37            | .39    | .41   | .37              | .37    | .37   |
|              | LIN+NR  | .37            | .40    | .41   | .38              | .38    | .35   |
|              | WDRC    | .37            | .38    | .40   | .37              | .35    | .36   |
|              | WDRC+NR | .37            | .39    | .40   | .37              | .36    | .35   |

**Table 5. Mean SII values for each condition.**

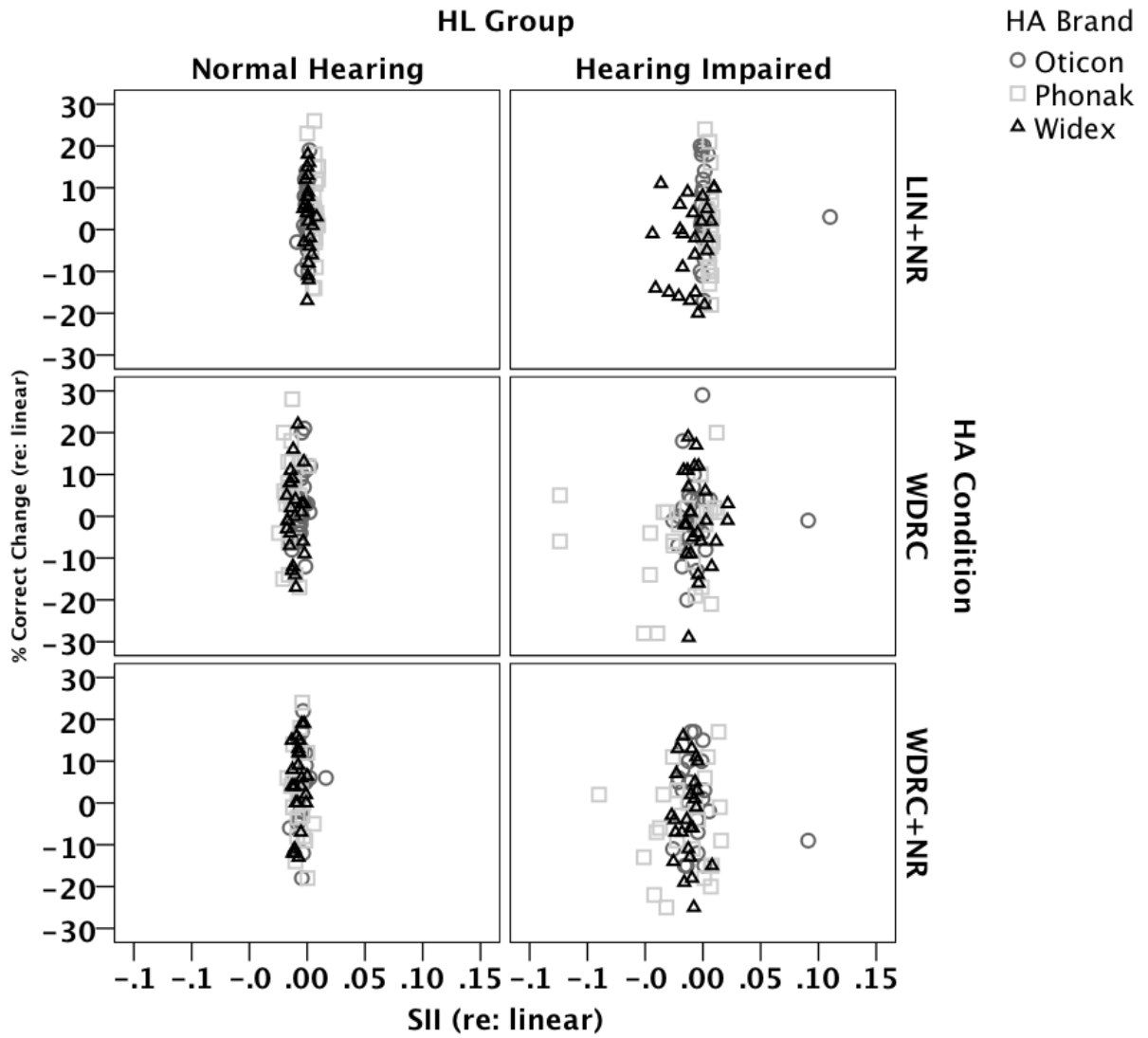
The relationship of most interest to our research was whether SII changes could better predict speech perception changes than SNR. This is plotted in Figure 25. To assess whether speech perception changes could be predicted, a multiple linear regression analysis was performed within each group. The initial model used predictors of primary interest: SII (re: LIN; continuous variable), brand (dummy variable), and processing (dummy variable) as predictors. A second model also included predictors of each subject (coded as dummy variables) to observe any changes to primary predictors when variance within subjects were accounted for in the model.

For the NH group, the first model revealed that SII and brand were not significant predictors, but that processing condition was significant. All processing conditions were significant compared to the LIN condition: LIN+NR ( $t=2.644$ ;  $p=.009$ ), WDRC ( $t=2.134$ ;  $p=.034$ ), and WDRC+NR ( $t=2.777$ ;  $p=.006$ ). The total amount of variance explained from this model was 3.8%. When subjects were added into the model, the significance of condition predicting speech perception changes changed only slightly in that WDRC was no longer significantly different than the LIN condition ( $t=1.781$ ;  $p=.076$ ). The effect of LIN+NR ( $t=2.910$ ;  $p=.004$ ) and WDRC+NR ( $t=2.583$ ;  $p=.010$ ) did not change from the first model. The effect of SII at the hearing aid output remained insignificant when these other variables were controlled. The total amount of variance in speech perception explained with this model was 21%.

For the HI group, the first model revealed that brand of hearing aid was a significant predictor of speech perception, but that SNR and processing condition were not significant. Oticon was significantly different from both Widex ( $t=-2.308$ ;  $p=.022$ ) and Phonak ( $t=-3.028$ ;  $p=.003$ ). The total amount of variance in speech perception

scores explained by this model was 5.3%. When subjects were added as predictors into the second model, the significance of primary predictors did not change in that brand stayed significant (Widex:  $p=.010$ ; Phonak:  $p=.001$ ), while the others remained insignificant. The total amount of variance in speech perception explained by this model was 24.7%.

In summary, it appears that changes in speech perceptions scores (re: LIN) were predicted by the type of hearing aid processing for NH subjects and by hearing aid brand for HI subjects. The SII (re: LIN) was not a significant predictor of speech scores for either group. These results are nearly identical to those for the SNR (re: LIN) predictions. The inclusion of audibility, as measured by the SII, did not appear to change the results or ability to predict speech perception changes.



**Figure 25. The relationship between SII (re: LIN) and percent correct (re: LIN) for each group and condition.**

## **Chapter 5. Discussion**

This research aimed to answer two questions: 1) to what extent do hearing aid algorithms change the input SNR when programmed for an individual? 2) Does a change in the hearing aid modified SNR correspond to a change in speech perception? Because hearing impaired listeners often do not benefit from changes in SNR as much as normal hearing listeners, these questions were addressed in both groups. On average, WDRC and noise reduction processing modified the SNR from the input to the output of the aid between -1.75 to 0.25 dB, with a mean change of -0.25. The changes in SNR were not correlated to changes in speech perception. Furthermore, audibility changes (quantified by the SII) from hearing aid processing also did not correlate to changes in speech perception. The lack of correlation to speech perception may have been due to 1) little variability in SNR or SII made by hearing aid processing, 2) the CST not being sensitive enough to capture the small changes in SNR or SII at the output of the hearing aid, and/or to 3) the HA processing modifying the long-term SNR, but not the instantaneous SNR when speech was present. These reasons and implications of these results are discussed. Furthermore, individual differences are addressed.

### **Little Variability in SNR from Input to Output**

The changes in SNR from hearing aid input to output were slightly smaller than we predicted, based on previous research using this method to estimate the SNR and our own pilot data showing a 1-5 dB change. The modifications to SNR in our study depended on HA brand and type of processing, with WDRC showing the greatest effect on SNR. The differences between previous results (e.g., Hagerman and Oloffson, 2004; Souza et al., 2006; Naylor and Johannesson, 2010 Chong and Jenstad, 2010) and ours

are likely due to differences in test parameters, such as input SNR and type of background noise. The input SNR varied based on individual needs (i.e., SNR-50), which could have concealed more systematic effects had we tested at multiple fixed SNRs. Most of our listeners required an SNR for 50% (i.e., input SNR) correct between 0 and 4 dB SNR. Previous work showed greater deviations in SNR from linear when using WDRC processing when the SNR input was more favorable (Souza et al., 2006; Naylor and Johannesson, 2009). One option would have been to choose input SNRs that reflect real world listening. Pearsons and colleagues (1977) suggest that SNRs in real world environments vary from -2 dB in trains and airplanes to +9 dB inside urban homes (and even more favorable SNRs in suburban environments and school classrooms). It's also possible that our choice of background noise prevented greater changes in output SNR to be observed. Naylor & Johannesson (2009) showed the effect of noise type on SNR changes caused by WDRC. They found that speech backgrounds tended to have the least effect on SNR than steady-state noise backgrounds.

There is very little previous work assessing changes in SNR made by noise reduction algorithms. Data from Chong and Jenstad (2010) using Hagerman's phase-inversion technique suggests that noise reduction algorithms reduce the level of steady-state and cafeteria noise by 5-7 dB on average, while not changing or even enhancing the level of fricatives. Indirectly, this suggests that SNR may be improving with noise reduction processing. Furthermore, Hagerman and Oloffson (2004) used a simulated noise reduction algorithm and showed SNR improved by 3-5 dB with all five algorithms tested, but only in a steady-state background, not in multi-talker babble. In fact, in multi-talker babble, only a -1 to +1 dB change in SNR was found. Furthermore, for speech in a background of babble in one hearing aid tested, the authors found the SNR

was only improved by noise reduction for very unfavorable SNRs (all worse than 0 dB SNR). Therefore, the effect of input SNR and type of noise on the WDRC results in this study also hold true for the noise reduction results, likely explaining why the noise reduction algorithms only showed an average of 0.11 dB SNR improvement in both groups. Further, on average, WDRC processing showed the strongest effect, while WDRC+NR had a slightly more favorable SNR. In other words, under these test conditions, the effect that the noise reduction algorithms has on the SNR is very small, regardless of combining it with linear or WDRC processing.

### **Sensitivity of the Connected Speech Test**

The relationship on the CST between SNR and speech perception has been reported for NH listeners as 2 dB SNR / 23% change in speech perception and for HI listeners as 2 dB SNR / 17% change in speech perception using two passage pairs (Cox et al., 1987; 1988). The changes we saw in SNR caused by HA processing on average was -1.75 dB at the most. It's possible that the stimulus we chose was not able to measure the perceptual changes we were interested in measuring. Furthermore, the CST was chosen due to how it reflects real world situations, where the listener usually knows the topic, but subsequent details about that topic are being interpreted. This means that we provided a lot of context to our listeners, which may have masked any small changes in improvement due to audibility. Listeners vary widely in their ability to use contextual cues (e.g., Pichora-Fuller et al., 1995). If we had chosen a test that was low in context and forced a high reliance on audibility (e.g., non-sense syllables or words in isolation), we might have seen a stronger relationship between changes in SNR and speech perception.

## **Acoustic Changes to Instantaneous SNR and other Speech Cues**

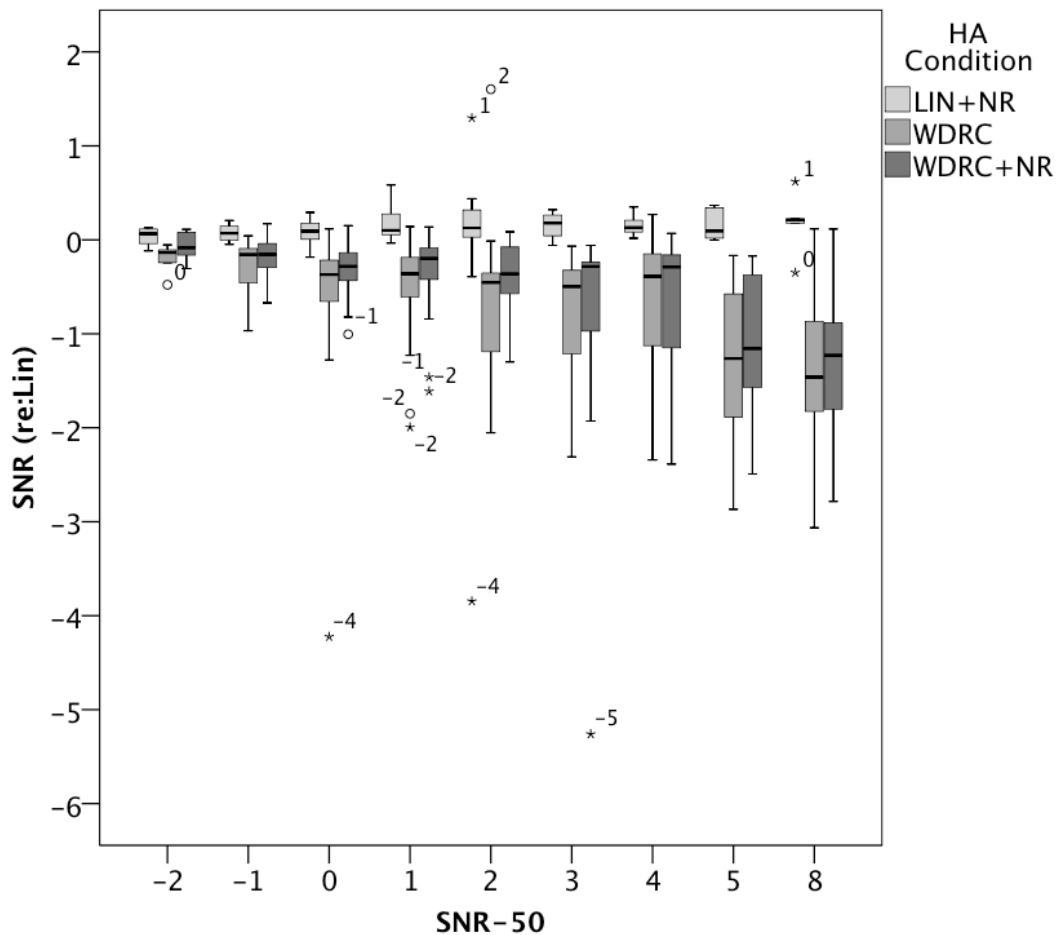
We measured the changes in SNR over the long-term (e.g., averaged over 30 seconds of speech), which is the method often used in research for this purpose. The long-term SNR has been shown to relate to speech perception when modified in the environment. For example, when modifying SNR by increasing or decreasing the speech or noise level, then presenting the combined signals to the listener, the relationship to speech perception is strong (e.g., 1 dB SNR / 10% change in speech perception; Nilsson et al., 1994). While the WDRC and noise reduction processing are modifying the long-term SNR, they may not be modifying the instantaneous SNR. In other words, when speech is present, the gain for both speech and noise may be reduced (WDRC) or unmodified (noise reduction), but the relative difference between speech and noise at that moment in time is unchanged. However, during a pause in speech, the gain may be reduced (noise reduction) or increased (WDRC), but this gain change is applied to the noise signal only. Therefore, over the long-term, the noise only segments are changing in SNR; however the speech plus noise segments are not changing. This has the effect of modifying the long-term SNR, but not the instantaneous SNR. If the SNR when speech is present is unchanged, then speech perception is not expected to change either.

It is also likely that the algorithms used in this study were making other acoustic modifications to the signal that may have contributed to the variability in speech perception scores (Rhebergen, 2009), including changes to the temporal envelope. Work by Jenstad & Souza (2005) has shown that WDRC does modify the temporal envelope if settings are severe enough, and that greater distortions in temporal envelope can predict declines in speech perception benefit with hearing aid processing. Typically

only fast release times (e.g., less than 100 ms) and high input levels (65 dB SPL or greater) create a significant amount of envelope distortion (Jenstad & Souza, 2005). Both of these criteria were in the scope of the hearing aid parameters used in this study, particularly for the HI group. This could be an explanation for why most of our listeners did not benefit from changes in SNR. It's also unknown what effect noise reduction algorithms have on the temporal envelope. Future work may include quantifications of temporal envelope changes, to observe whether these modifications contributed to the lack of benefit in SNR observed by the majority of HI listeners.

### **Individual Variability in Benefit from SNR Changes**

Our methods were designed for clinical purposes, to evaluate whether SNR changes, measured post-Hagerman processing, would vary in a way which could be useful for clinical purposes (i.e., improving speech perception). If we could identify which listeners would benefit from SNR improvements, clinical recommendations and outcomes could be more focused. Our impressions from these results are that the SNR changes very little by WDRC or noise reduction algorithms, at the listener's approximate SNR-50. However, if we plot the SNR changes at the HA output by the subject's SNR-50, we do see some trends consistent with results from other researchers (Figure 26). It appears that while input SNR (i.e., the listener's SNR-50) doesn't affect the LIN+NR condition very much, the effect is much stronger for WDRC and WDRC+NR processing. As a listener requires a better SNR for 50% correct, the more likely their hearing aids are to make the SNR worse. This seems counterintuitive, given that people with worse speech in noise abilities are those who need improvements in SNR from their aids the most.

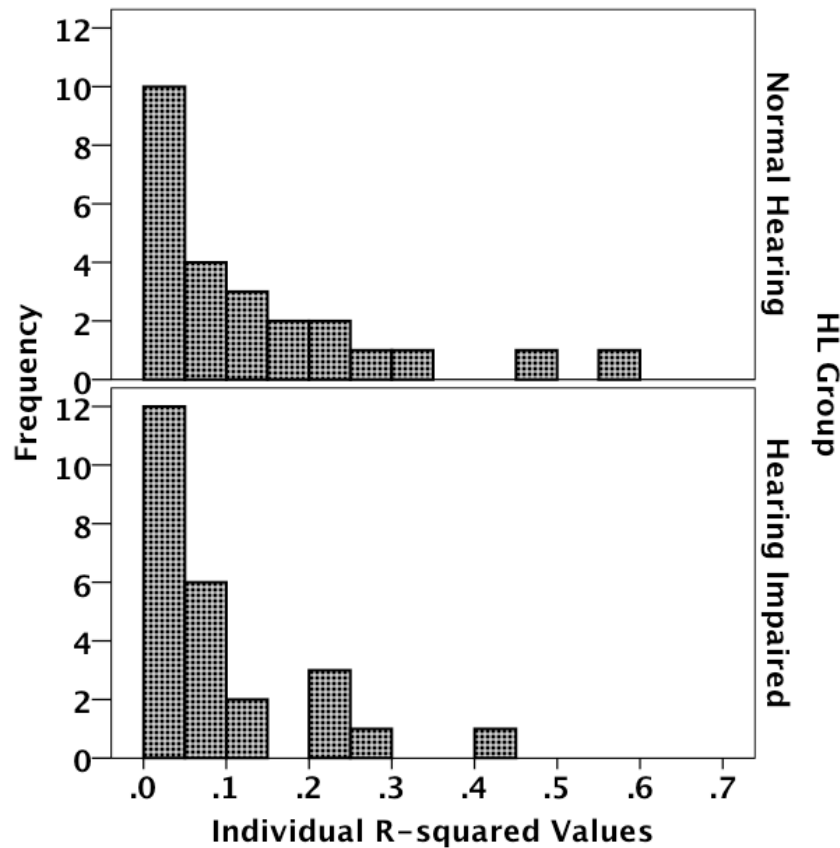


**Figure 26. SNR changes (re: LIN) made by hearing aid processing as a function of input SNR, or the listener’s approximate SNR-50. Boxes show the median (dark line) and 50% interquartile range, with whiskers stretching to the most and least extreme cases. The stars and circles labeled with specific SNR (re: LIN) numbers are data points deemed as outliers by the SPSS statistical program. Consistent with prior research, as the input SNR becomes more favorable, the output SNR becomes less favorable for WDRC processing. The effect of noise reduction seems to improve the SNR by a small amount, particularly at more favorable SNRs, regardless of combining it with linear or WDRC processing.**

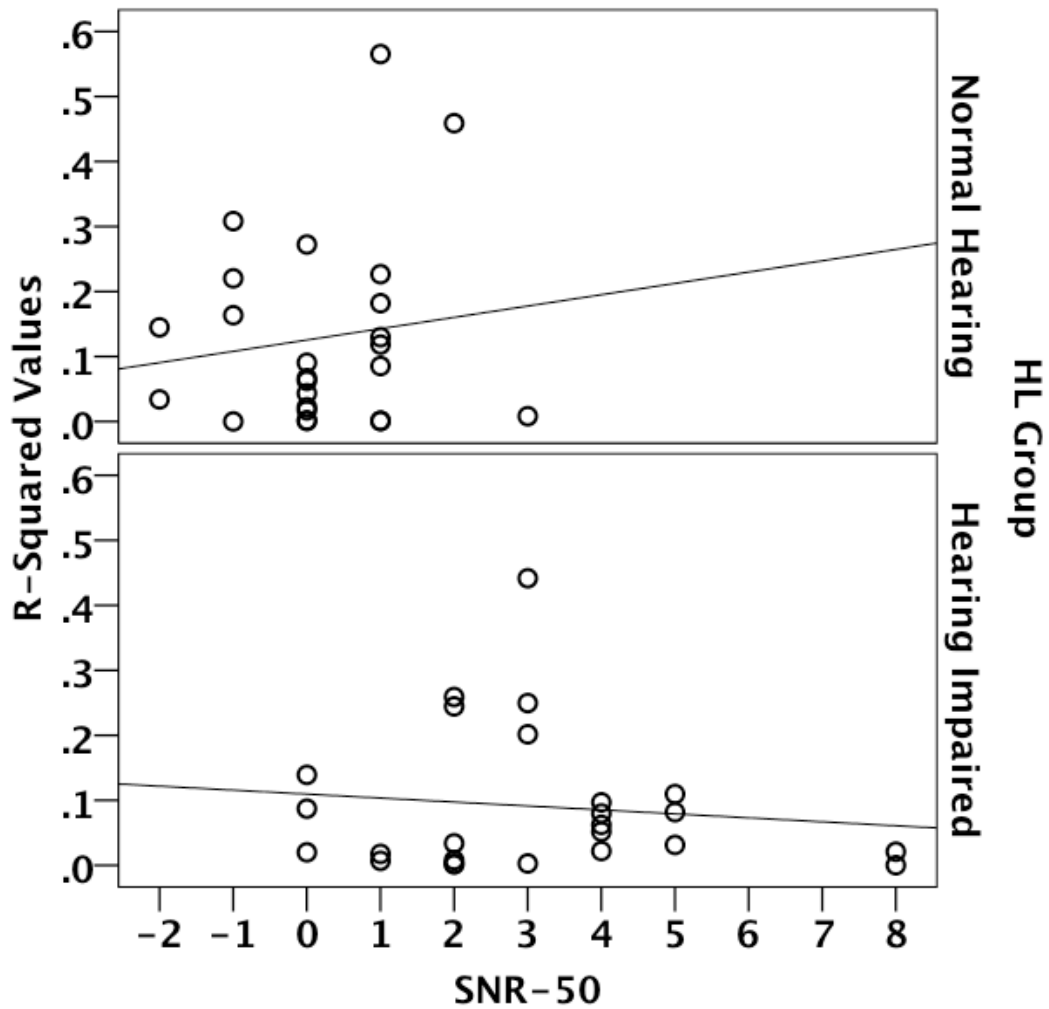
To further explore the idea that those with greater SNR needs (e.g., more favorable SNR-50) would experience greater declines in the output of their hearing aid, the data were analyzed on an individual level. We wanted to examine whether listeners who benefited more from changes in SNR could be identified from those who didn't benefit. Each subject's speech perception scores (re: LIN) were plotted relative to the output SNR (re: LIN). A linear regression analysis was performed to determine the slope of each individual's performance intensity function, as done in the development of the CST (Cox et al., 1988, 1988). The histogram in Figure 27 shows the distribution of  $R^2$  values for each group. None of the correlations were statistically significant while accounting for multiple comparisons. These distributions illustrate that while the effect size of changes in SNR at the HA output on changes to speech perception is small (e.g., the majority of values are less than .5), the effect size appears to be similar between groups since their distributions largely overlap. In fact, a t-test between the mean  $R^2$  value for each group was not significant ( $t=1.034$ ;  $p=.306$ ). Therefore, it appears that NH listeners are not able to benefit any more than HI listeners from changes to SNR, despite previous research showing differences in slope between the two groups. Specifically for the CST, the slope for NH and HI listeners has been reported as a 23% and 17% change in speech perception, respectively, for every 2 dB change in SNR (Cox et al., 1987, 1988). We did not observe differences in slope between groups in our study, likely due to the inability of WDRC and noise reduction processing to modify the SNR by 2dB.

We hypothesized that perhaps listeners who required greater SNR-50 values would experience a greater change in SNR at the HA output, and possibly they would experience a stronger relationship between SNR and speech perception changes. To

address this hypotheses, we plotted individual  $R^2$  values against their SNR-50, and found very little correlation for the NH group ( $r=0.136$ ;  $p=.516$ ) or the HI group ( $r=-0.117$ ;  $p=0.577$ ). In other words, as SNR needs increased, there was no correlation to how well a listener could benefit from SNR changes; despite greater changes in HA output SNR (Figure 28). Other factors about the individual (e.g., working memory abilities, biological encoding of SNR) may better differentiate between listeners who can benefit from SNR changes and those who can't benefit, and will be considered in future studies.



**Figure 27. The distribution of individual  $R^2$  values between SNR (re: LIN) and percent correct (re: LIN) for both groups. None of the individual correlations were statistically significant.**



**Figure 28. Each data point represents an individual’s  $R^2$  value between SNR (re: LIN) and speech perception scores (re: LIN), plotted against their SNR-50. The linear regression lines within each group showed little correlation between these two variables (NH:  $r=0.136$ ;  $p=0.516$ ; HI:  $r=-0.117$ ;  $p=.577$ )**

Previous research suggests that many other factors may contribute to speech perception in noise, including audibility (e.g., Humes, 2002), listening effort (e.g., Rudner et al., 2012), and working memory (e.g., Humes, 2002). We quantified the

contribution of audibility using the SII, and found no significant correlations to speech perception, under the test conditions of this study. It could also be argued that asking listeners to perform at 50% correct for an extended period of time might be too hard for some listeners (i.e., require too much effort), supported by results from Rudner et al. (2012) showing a significant correlation between environmental SNR and effort ratings. However, post-hoc analysis showed no systematic decline in speech perception over time for either group. A linear regression analysis between the presentation order of conditions (e.g., to represent the length of test time) and speech perception performance showed very little relationship in either the NH ( $R^2=.0004$ ) or HI ( $R^2=.013$ ) group. Working memory differences may have contributed to individual variability in the ability to take advantage of SNR improvements (Gatehouse et al, 2003; Lunner and Sundewall- Thoren, 2007). The influence of working memory on speech perception in noise is particularly highlighted in more complex background noises, such as the 6-talker babble used in this study (Rudner et al., 2012). Future work might include measures of working memory and individual tolerance of noise.

## **Chapter 6. Conclusion**

Based on the hearing aid brands and SNR conditions used here, the SNR is only slightly affected by hearing aid processing and it does not significant affect speech perception. The influence of SNR on other hearing aid outcomes (e.g., self-report questionnaires) is still unanswered. Future studies should include SNR measurements that reflect real world listening situations, working memory quantification, individual tolerance to noise, as well as outcomes sensitive to hearing aid use and satisfaction in every day life.

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## **Appendix A**

### **Interleaved Harmonic Method (Bell et al., 2010)**

Although not reported, only minimal overlap between main and side lobes of each harmonic is likely to occur due to the use of an integer number of periods within each window of the FFT. If less than whole number of periods were used, the spectral leakage into adjacent frequency bands would have been greater. Error terms generated by using this technique with non-linear hearing aid processing are of concern and this method needs to be tested with more modulated signals. The authors used Matlab simulations to estimate the distortion imposed by compression processing. In the Matlab program, the compression parameters of a single channel hearing aid could be specified (ratio, kneepoint, gain, maximum output, time constants), as well as the modulation rate and depth of the input signal. With linear gain and an un-modulated signal, the spectral leakage was found to be -78 dB between bands. The maximum spectral leakage occurred with a compression ratio of 3:1, attack time of 25 ms, and using a modulated input signal at a rate of 5 Hz, where the leakage was -15 dB between adjacent bands. When both a modulated and un-modulated signal was presented simultaneously, the spectral leakage was not reported. However, the authors stated that the difference between the SNR estimated by the Matlab simulation and the SNR measured using the interleaved harmonics approach with real hearing aids was less than 0.05 dB and increasing to 1 dB at an SNR of +15 dB. Unfortunately this simulation only informs us of the effects of compression and signal type on the results, without any information on distortion arising from the actual measurement technique. The between-subject variability in real ear measures using the technique described by Bell et al. (2010) is quite high, even when the same hearing aid was used. This may be due to

the way real ear insertion gain was applied. One subject was used to establish the gain settings, and then the gain settings in the aid were kept the same for all subjects. The variability within real ear unaided responses can be as high as 20 dB at some frequencies (Valente, Valente, & Goebel, 1991) and insertion gain may vary between subjects by up to 5 dB in the higher frequencies (Dillon, 2001), which may have contributed to the high between-subject variability observed in this study

## **Appendix B. Olofsson and Hansen (2006)**

In the method suggested by Olofsson and Hansen (2006), two output signals are compared, which are linearly related by a specific type of filter, the Hilbert transform. These two output signals are called a Hilbert pair due to this relationship. After passing these two signals through a hearing aid, how much the output signals deviate from being an exact Hilbert pair is calculated as the non-linearity or distortion of the system. The authors argue that this measure is sensitive to fast acting non-linearities, but not to slow-acting automatic gain control or similar processing strategies. For a purely linear system, the two output signals result in a perfect Hilbert pair. Using this technique, Olofsson and Hansen (2006) correlated the distortion measure results to perceptual measures of distortion. Specifically they simulated 12 hearing aid conditions, all had a compression ratio of 2:1, with variable compression channels (1, 3, or 15), time constants (instantaneous, fast, or slow), and kneepoints (0 or 25 dB HL). The stimuli were a male and a female voice, guitar and piano music presented at 65 and 75 dB SPL. The distortion of each system was characterized using the Hilbert pairs of each signal, and ten normal hearing listeners were asked to rate the clarity of each sound using a sliding scale from 0-10 (0=very strong distortion; 1=very unclear; 5=midway; 9=very clear; 10=totally free of distortion). The results of the subjective impressions showed that as the time constants increased, perceived distortion or reduced clarity increased. This was also reflected in the objective measure of distortion, with slower time constants yielding less distortion. A correlation between the subjective and objective measures was only moderate ( $r=0.677$ ), and the significance of this correlation was not mentioned. However, the authors of this paper suggest the relationship is strong

between objective and subjective measures, and recommend the use of this method for estimating distortion in non-linear systems.

## Appendix C

### Rhebergen et al. (2009)

Rhebergen et al. (2008, 2009) proposed a different method for speech and noise separation, designed to overcome the potential distortion caused by using Hagerman's phase inversion technique with non-linear systems. They proposed that the speech and noise signals are separately compressed by the gain factor in every sample of the speech and noise stimuli, and then the change in output is measured. To do this, first the speech plus noise signal is fed to a compressor to obtain the gain factors for each sample given by that system. These gain factors are stored, then applied to the speech and noise signals separately. The output of this manipulation is analyzed with respect to the original speech and noise signals. Using this technique, they found that WDRC reduced the SNR at more favorable input SNRs, and slightly increased it at less favorable SNRs in interrupted noise (Rhebergen et al., 2009). Increasing the compression ratio further exacerbated this effect. In interrupted noise, this effect was stronger, showing a much greater improvement in SNR at negative SNR inputs. The author attributes the difference between noise types to which signal is dominating the behavior of the compressor. At negative input SNRs, steady-state noise is likely dominating the compressor decision compared to interrupted noise in this condition. Comparing the acoustic SNR measures to the speech thresholds in noise (SNR-50) for eight normal hearing listeners, the agreement was very strong for all compression conditions in steady-state noise, and for the two most mild compression conditions in interrupted noise (Rhebergen et al, 2009). Once the compression ratio became strong (2:1 and 4:1) in interrupted noise, other distortions caused by compression processing seem to dominate and perception was much worse than the acoustic SNR predicted. The

acoustic results from Rhebergen et al. (2009) are in agreement with others using Hagerman's phase-inversion technique to measure the change in SNR from input to output (Hagerman & Olofsson, 2004; Souza et al., 2006; Naylor & Johannesson, 2010).

## **Appendix D. AI/SII and STI**

While the AI is useful to predict the audibility of speech, which can be filtered or masked speech, it does not handle reverberation or temporally distorted speech. Correction factors have been recommended, but more often another prediction of speech intelligibility is used, called the Speech Transmission Index (for discussion, see Appendix ?). Currently, there are two methods for predicting speech intelligibility: the Audibility Index or Speech Intelligibility Index, and the Speech Transmission Index. Both methods were used to predict the intelligibility of subjects with normal and impaired hearing in a variety of listening situations in a retrospective analysis (Humes et al., 1986). The same speech-shaped noise spectrum was used as the input signal to both models, and the results were compared to speech intelligibility tested in various studies. The listening environment conditions used a 1) reverberated speech signal, 2) filtered speech, 3) For each listening condition, the environment that the listeners were tested in was modeled using either the AI or STI to determine what value was predicted.

In the reverberated speech condition, the SNR for 50% correct was found for both groups of listeners. This SNR and listening situation was then applied to each model, and the predicted AI or STI value was produced. Theoretically, the models should produce the same value across reverberation conditions since the criterion was always 50% correct. The STI did a better job of predicting speech intelligibility for both normal and hearing-impaired subjects across different reverberation times. However, the STI model produced higher values for the normal hearing listeners than the hearing impaired listeners, suggesting that some form of distortion in the hearing impaired listeners was not being accounted for by the STI model. On the other hand, the AI

model predicted similar values for short reverberation times (less favorable SNR), but did worse as the reverberation time increased in length (more favorable SNR).

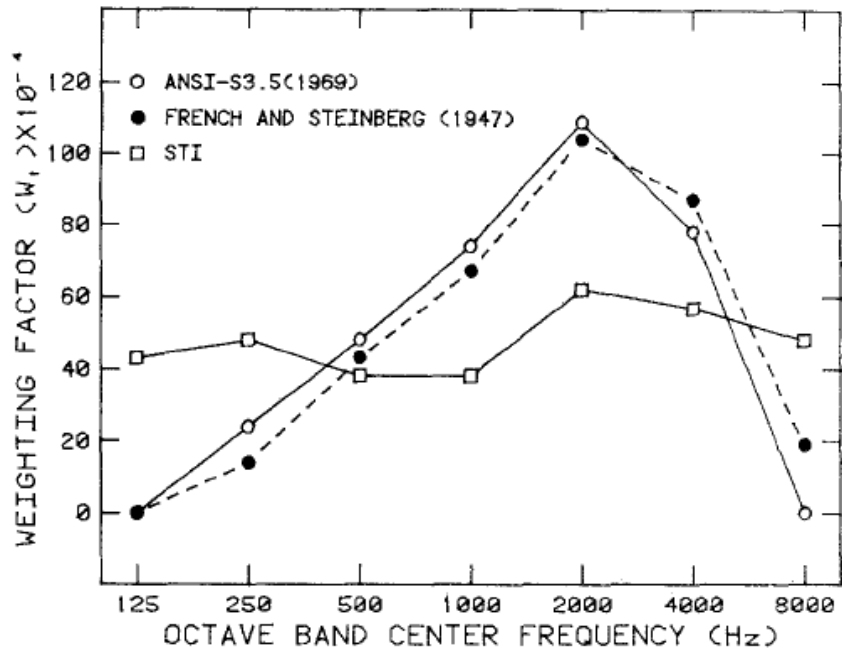


FIGURE 2. Octave-band weighting factors for two versions of the Articulation Index (ANSI-S3.5, 1969, and French and Steinberg, 1947) and the Speech Transmission Index (STI). Weighting factors have already been divided by 30.

**Figure . The importance of each band to the overall speech intelligibility predicted by the AI and STI models from Humes et al., 1986.**

In the filtered speech condition, 42 and 88 dB SPL levels were used with low and high passed speech. The AI predicted the intelligibility better than the STI for both groups of listeners using filtered speech. Humes and colleagues (1986) attributed this to the small number of bands used in the STI method or due to inappropriate weightings for sharply changing spectral input shapes. The authors concluded that the STI was better for modeling temporally distorted speech and the AI was better for modeling spectrally distorted speech.

To test this hypothesis, normal hearing listeners were asked to identify speech in six conditions of non-sense syllables in babble, and five filtered speech conditions. In general they found that octave bands were too gross to predict intelligibility when abrupt spectral changes occur in the signal (e.g., filtered speech). Humes and colleagues recommend using 1/3-octave bands. Both models predicted scores reasonably well under these controlled conditions. However, the AI was worse at predicting temporal distorted speech and the STI was worse at predicted spectrally distorted speech. These contradictory results between models highlight the trade off of spectral and temporal cues used for speech intelligibility, and the complexity of the interaction between them.

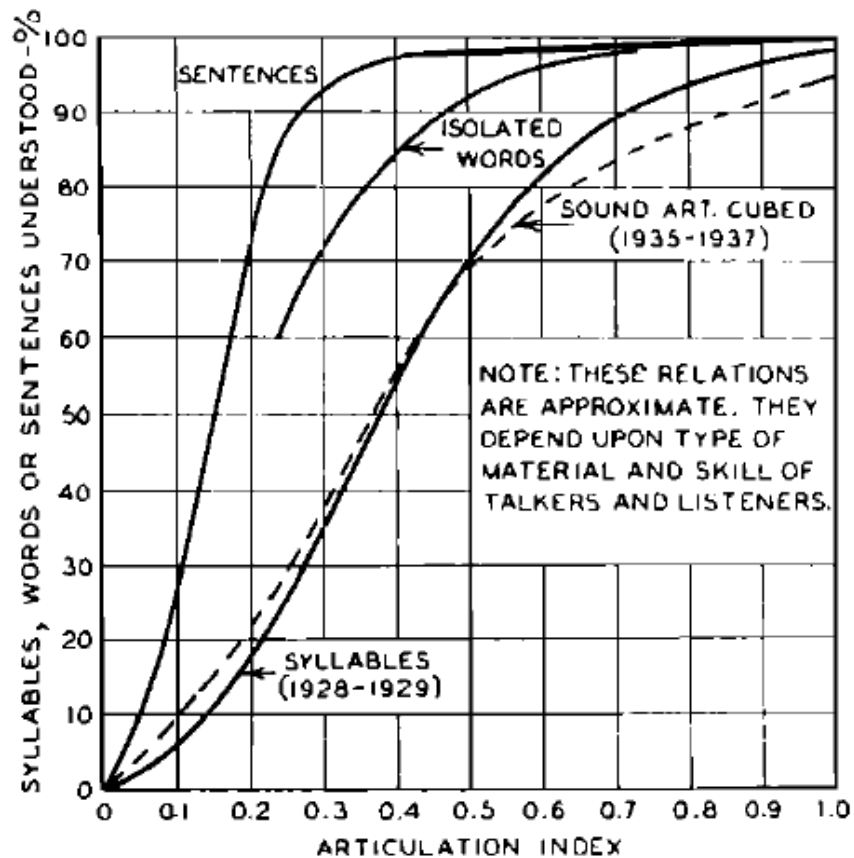


FIG. 23. Approximate relations between articulation index and subjective measures of intelligibility.

Figure . Transfer functions between Articulation Index and speech intelligibility (% correct) from French and Steinberg, 1947.

The Speech Transmission Index (STI) was originally developed based on principles of the AI (Steeneken & Houtgast, 1980), but differs from the AI in the way the SNR is calculated. The STI calculates reductions in the modulation index at the output of a system. It assumes the input signal is speech-shaped random noise, which is 100% modulated at a rate of 0.63-12.5 Hz (similar to the rate of running speech). The preservation of the modulation characteristics in seven octave bands is calculated at the output of a “system”, which can include a room, filtering, a hearing-impaired listener, a hearing aid, or noise conditions (Humes et al., 1986). The resulting modulation transfer function is then converted to SNR, and the STI is calculated as:

$$STI = \sum_{i=1}^N W_i [(SNR_i + 15)/30]$$

where  $i$  is the band number,  $W_i$  is the weighting applied to that band, and  $SNR_i$  is the SNR converted from the modulation transfer function for a given band. The weighting factors are quite different in the STI, compared to the AI, which are attributed to the stimuli used to develop them (Figure 14; Humes et al., 1986; Appendix ?).

## Appendix E. Subject Details

| Subject | HL Group | Age (yrs) | HA use | Aid Use (yrs) | Test Ear | LF PTA (dB HL) | HF PTA (dB HL) | SNR-50 (dB) |
|---------|----------|-----------|--------|---------------|----------|----------------|----------------|-------------|
| 1       | NH       | 23        | No     | 0             | right    | 3              | 2              | -2          |
| 3       | HI       | 23        | Yes    | 20            | left     | 45             | 53             | 2           |
| 4       | HI       | 82        | Yes    | unknown       | left     | 42             | 47             | 8           |
| 5       | HI       | 78        | Yes    | unknown       | right    | 35             | 42             | 0           |
| 6       | NH       | 67        | No     | 0             | left     | 15             | 17             | 1           |
| 7       | HI       | 60        | Yes    | unknown       | right    | 32             | 40             | 1           |
| 8       | HI       | 78        | Yes    | 5             | right    | 37             | 45             | 2           |
| 9       | NH       | 46        | No     | 0             | right    | 0              | 0              | -1          |
| 10      | NH       | 78        | No     | 0             | right    | 15             | 15             | -1          |
| 11      | HI       | 66        | Yes    | 38            | left     | 35             | 42             | 2           |
| 12      | HI       | 73        | No     | unknown       | right    | 13             | 27             | 4           |
| 13      | HI       | 65        | Yes    | 3             | left     | 10             | 13             | 0           |
| 14      | HI       | 77        | Yes    | 5             | left     | 33             | 45             | 5           |
| 15      | NH       | 78        | No     | 0             | right    | 8              | 10             | 1           |
| 16      | HI       | 87        | Yes    | 11            | left     | 35             | 38             | 2           |
| 17      | NH       | 70        | No     | 0             | left     | 8              | 10             | 0           |
| 18      | HI       | 73        | No     | 0             | right    | 15             | 18             | 3           |
| 19      | HI       | 86        | Yes    | 6             | right    | 40             | 40             | 1           |
| 20      | HI       | 85        | Yes    | unknown       | left     | 45             | 55             | 8           |
| 21      | NH       | 68        | No     | 0             | right    | 3              | 5              | 0           |
| 22      | HI       | 70        | Yes    | unknown       | left     | 33             | 45             | 4           |
| 23      | HI       | 76        | No     | 0             | right    | 12             | 18             | 2           |
| 24      | HI       | 78        | Yes    | 5             | left     | 40             | 47             | 3           |
| 25      | HI       | 80        | No     | 0             | left     | 12             | 22             | 4           |
| 26      | HI       | 72        | Yes    | 9             | right    | 25             | 35             | 4           |
| 27      | HI       | 68        | Yes    | 14            | right    | 48             | 50             | 3           |
| 28      | HI       | 29        | Yes    | 8             | right    | 53             | 65             | 5           |
| 29      | NH       | 34        | No     | 0             | left     | 13             | 10             | -2          |
| 30      | HI       | 46        | Yes    | unknown       | left     | 40             | 42             | 0           |
| 31      | HI       | 53        | Yes    | unknown       | left     | 48             | 43             | 4           |
| 32      | HI       | 80        | Yes    | unknown       | right    | 38             | 53             | 5           |
| 33      | HI       | 74        | No     | 0             | left     | 32             | 38             | 3           |
| 34      | HI       | 29        | No     | 0             | right    | 57             | 55             | 2           |
| 35      | NH       | 59        | No     | 0             | right    | 12             | 17             | 1           |
| 36      | NH       | 75        | No     | 0             | left     | 15             | 15             | 2           |
| 37      | NH       | 58        | No     | 0             | right    | 15             | 15             | 1           |
| 38      | NH       | 65        | No     | 0             | right    | 17             | 20             | 1           |
| 39      | NH       | 65        | No     | 0             | left     | 8              | 8              | 0           |
| 40      | NH       | 70        | No     | 0             | left     | 20             | 22             | 1           |
| 41      | NH       | 56        | No     | 0             | left     | 7              | 7              | 1           |
| 42      | NH       | 64        | No     | 0             | right    | 22             | 20             | -1          |
| 43      | NH       | 65        | No     | 0             | left     | 12             | 12             | 0           |
| 44      | NH       | 49        | No     | 0             | right    | 10             | 12             | 0           |
| 45      | NH       | 77        | No     | 0             | right    | 22             | 20             | 3           |
| 46      | NH       | 68        | No     | 0             | right    | 7              | 8              | 0           |
| 47      | NH       | 65        | No     | 0             | left     | 22             | 22             | 0           |
| 48      | NH       | 60        | No     | 0             | left     | 10             | 10             | 1           |
| 49      | NH       | 56        | No     | 0             | left     | 17             | 18             | 0           |
| 50      | NH       | 66        | No     | 0             | right    | 12             | 10             | -1          |
| 51      | NH       | 68        | No     | 0             | left     | 15             | 12             | 0           |

## Appendix F. CST Sample Passages

### Test Passage Pair 1 (Window/Glove)

#### Psg: 1 WINDOW

Windows **PROVIDE LIGHT** and air to **ROOMS**.  
Windows were **ONCE COVERED** with **CRUDE SHUTTERS**.  
Later, oiled **PAPER** was **USED** for windowpanes.  
**GLASS** windows **FIRST** appeared in ancient Rome.  
**COLORED** glass was used in European **WINDOWS**.  
**SOME CHURCHES** were **FAMOUS** for their **BEAUTIFUL** windows.  
These windows **DISPLAYED PICTURES** from the **BIBLE**  
**PIECES** of glass were **HELD** together by lead.  
**SUCH** windows **MAY** be seen in French cathedrals.  
English churches also contain **STAINED** glass windows.

#### Psg: 2 GLOVE

Gloves are **CLOTHING WORN ON** the **HANDS**.  
The **WORD "GLOVE"** **MEANS** "palm of the hand."  
**CRUDE GLOVES** were **WORN** by **PRIMITIVE MAN**.  
Greeks wore **WORKING** gloves to **PROTECT** their hands.  
The **ROMANS USED** gloves as a sign of **RANK**.  
Knights used to fasten gloves to their helmets.  
The gloves **SHOWED** their **DEVOTION** to their **LADIES**.  
A glove thrown on the **GROUND SIGNALLED** a challenge.  
Knights threw them at their enemy's feet.  
**FIGHTING STRTED WHEN** the enemy picked up the glove.

### Test Passage Pair 2 (Umbrella/Giraffe)

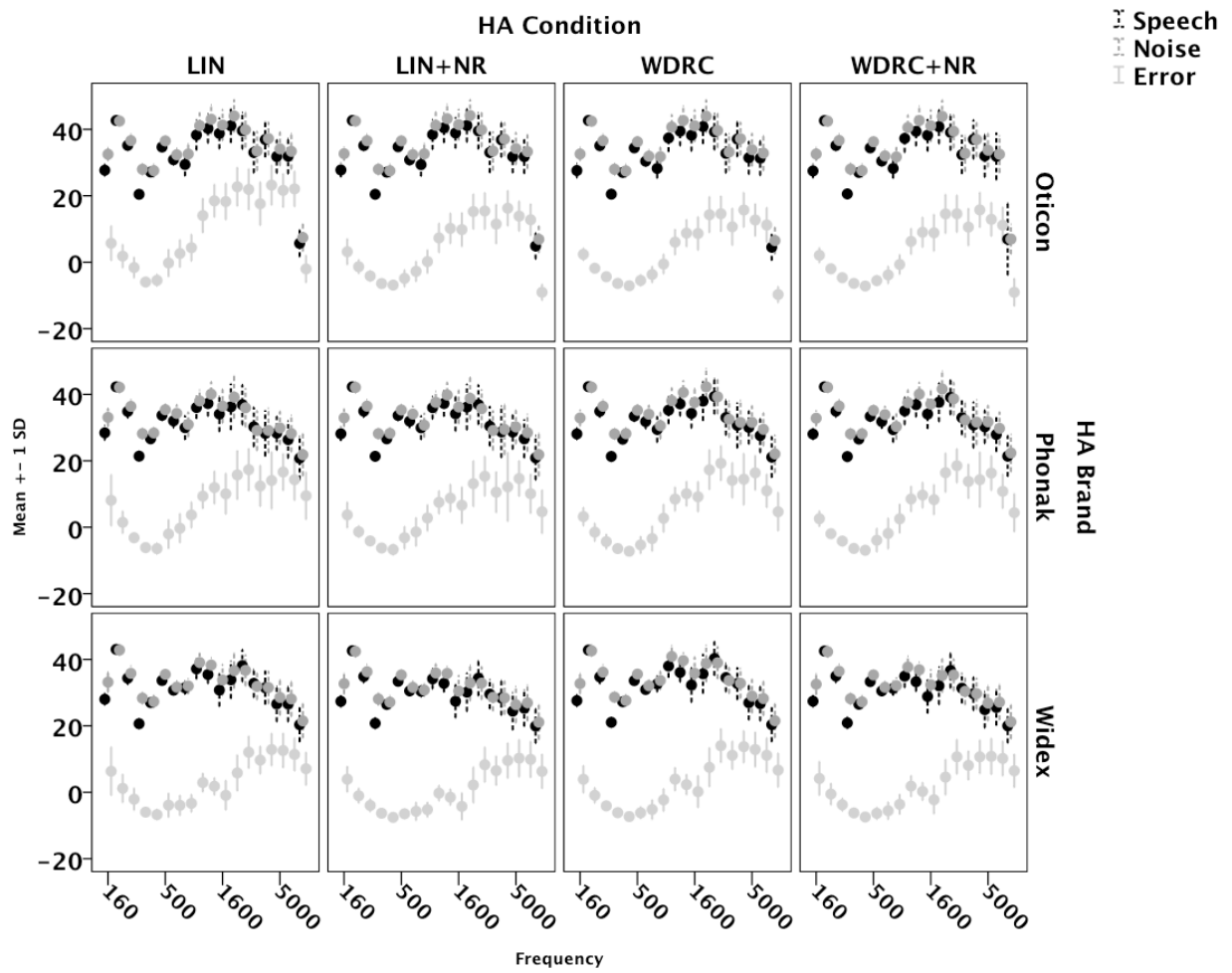
#### Psg: 3 UMBRELLA

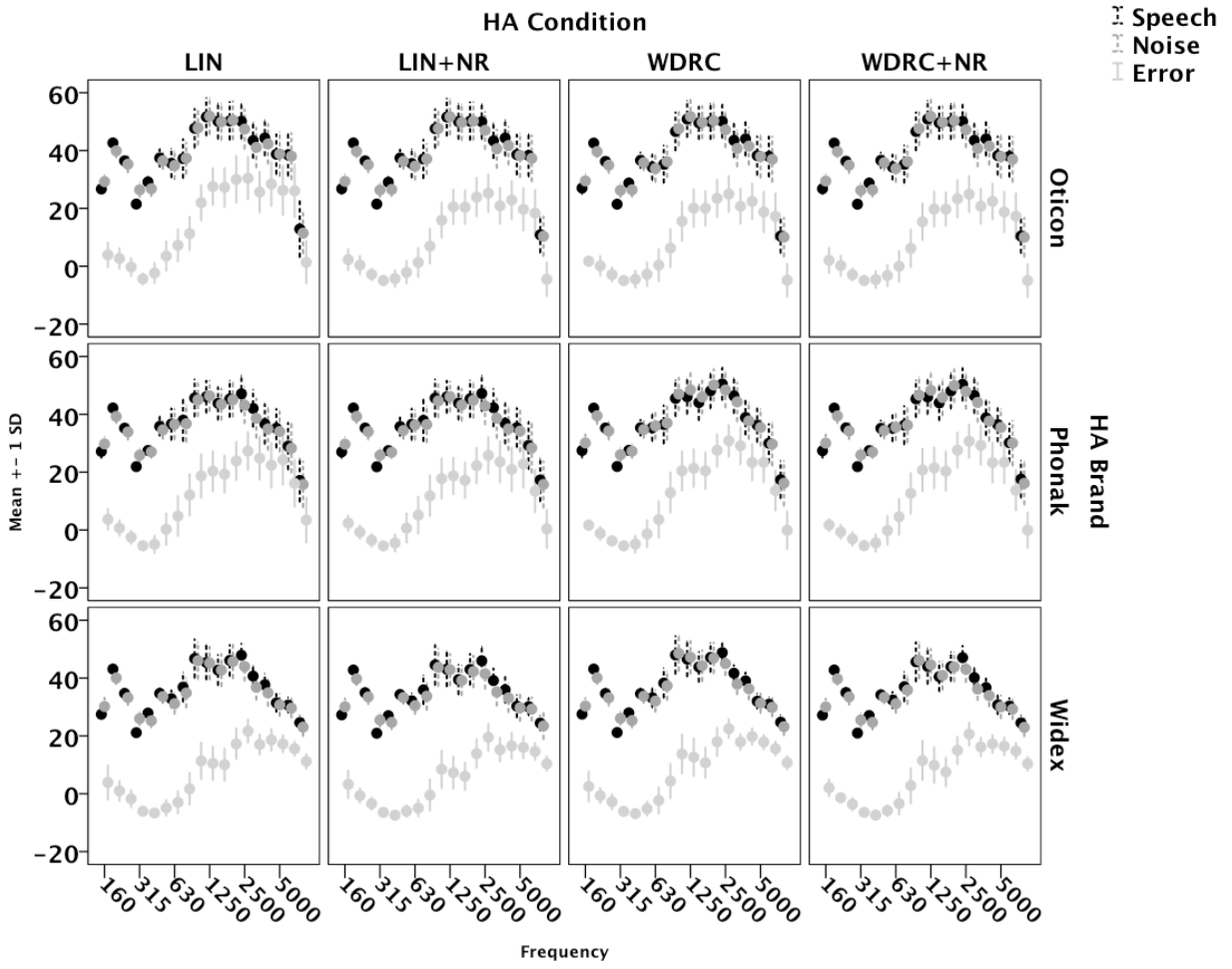
The **NAME** "umbrella means small shadow."  
Umbrellas **WERE** first used in **ANCIENT** Egypt.  
**THEY GAVE** protection **FROM** the fierce **SUNSHINE**.  
**SLAVES** held **UMBRELLAS** over their **MASTERS**.  
In Egypt today, many people **CARRY** umbrellas.  
In **EARLY** Rome, **ONLY WOMEN** used umbrellas.  
**IF** a **MAN** did, he **WAS CONSIDERED** a sissy.  
Umbrellas were **USED** by both **SEXES** in **ENGLAND**.  
**TODAY**, people use umbrellas to keep **OUT** the **RAIN**.  
Umbrellas **USED** as sunshades are called parasols.

#### Psg: 4 GIRAFFE

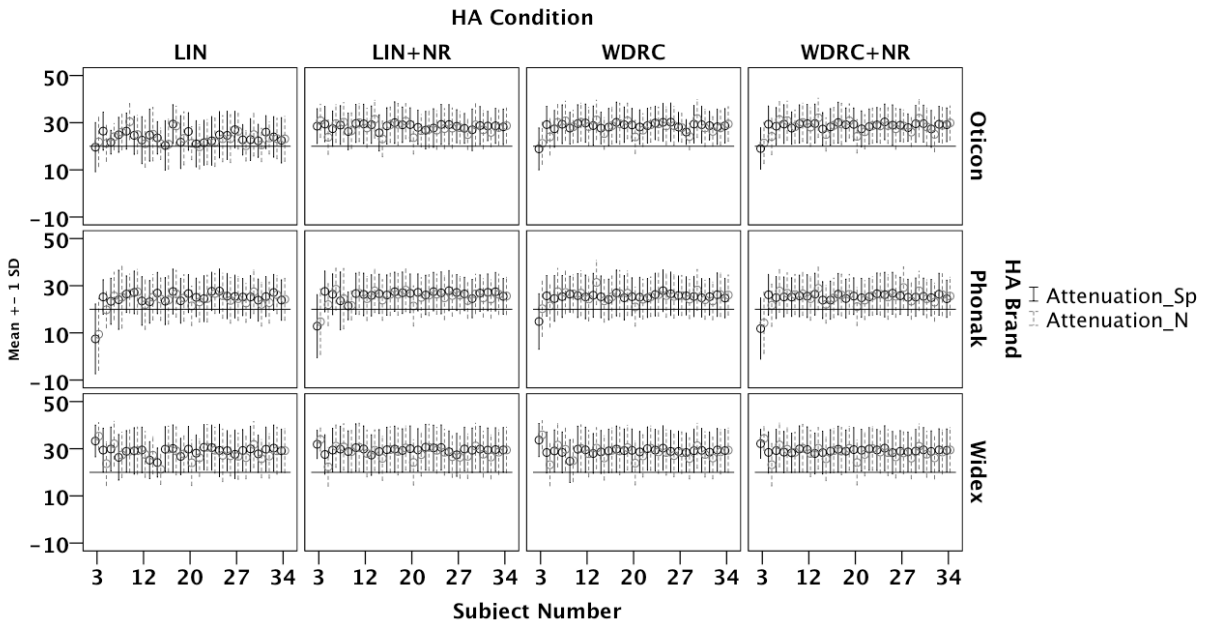
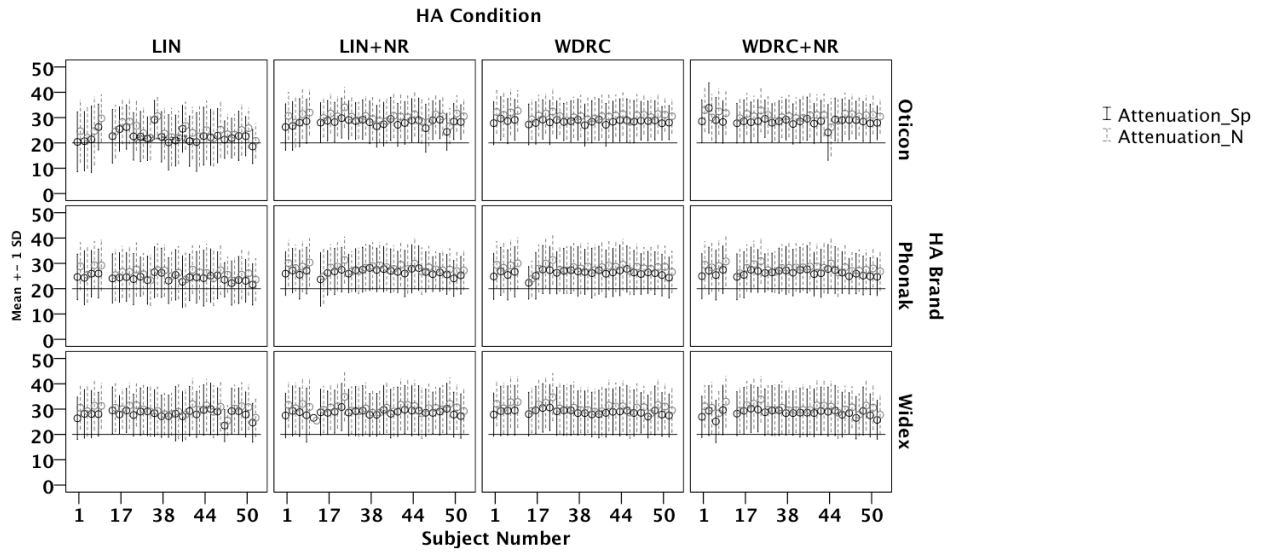
The giraffe is the tallest wild **ANIMAL**.  
It is three time taller than a man.  
A full grown giraffe is eighteen **FEET** high.  
The giraffe has an extremely **LONG NECK**.  
The neck **HAS ONLY** seven **NECKBONES**.  
The **GIRAFFE'S BODY** is about the **SIZE** of a **HORSE'S**.  
The **BODY** is **SHAPED LIKE** a triangle.  
Africa is the only **COUNTRY WHERE** giraffes **LIVE WILD**.  
**LARGE GROUPS** of them are **FOUND ON** the **PLAINS**.  
They live there with **LIONS** and **ELEPHANTS**.

## Appendix G. Extracted Levels for NH and HI group.

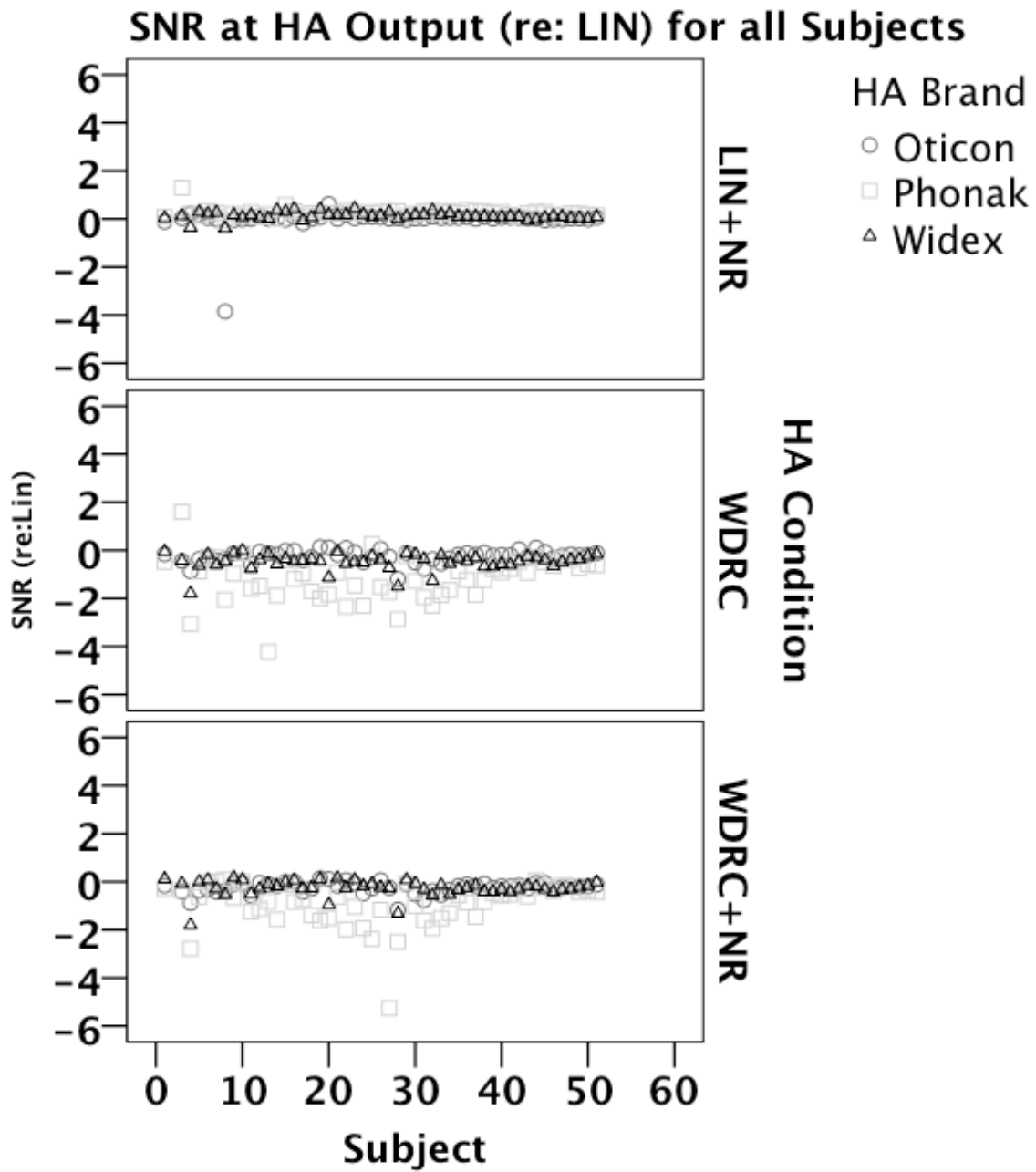




**Appendix H. Individual attenuation values for the NH (top graph) and HI groups (bottom graph).**



**Appendix I. Individual SNR changes at the HA output re: LIN.**



**Appendix J. Contrasts for Interaction of Brand and Condition in the change in SNR re: linear scores.**

**NH Group Interactions**

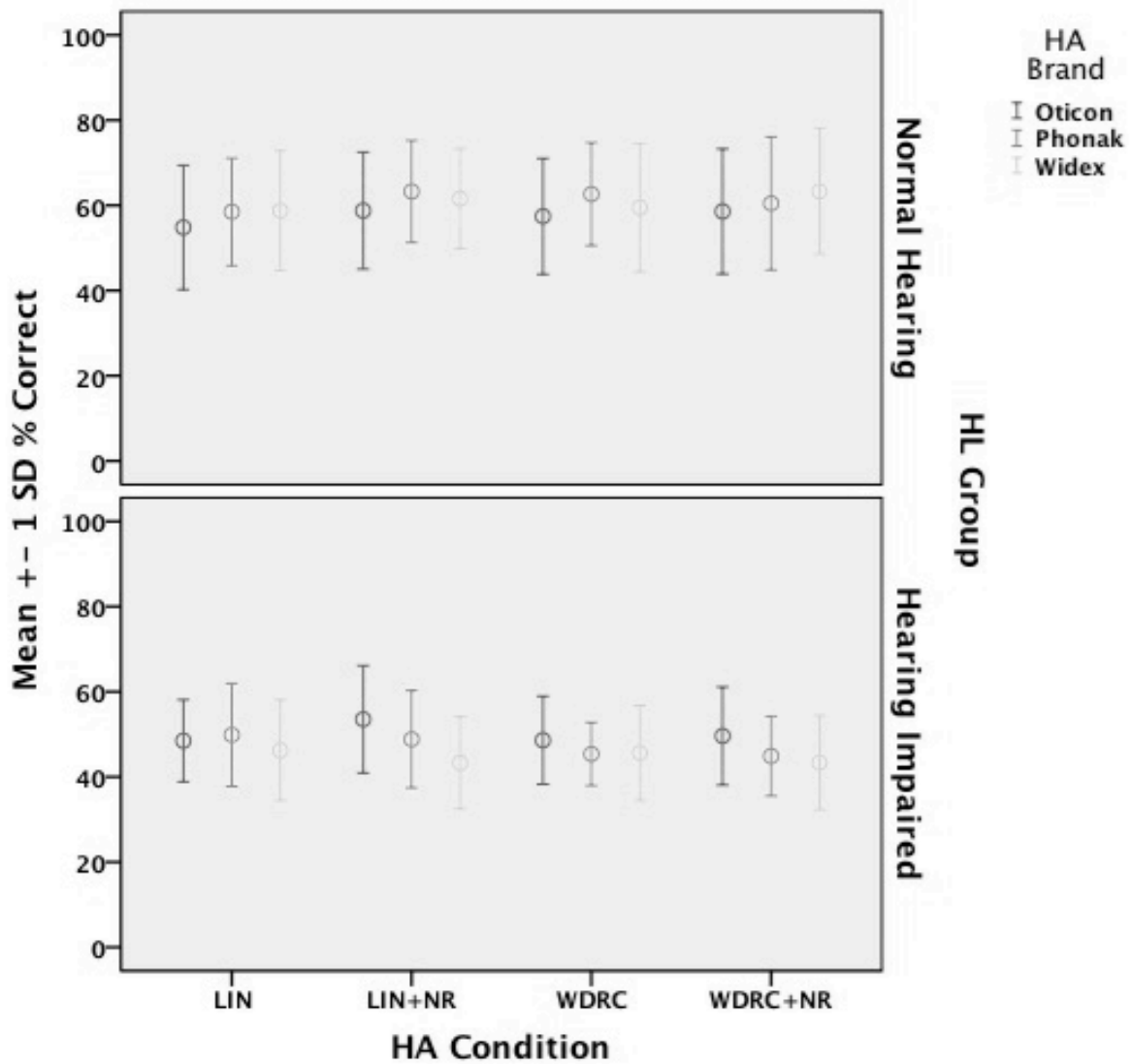
| HA Brand      |               | HA Condition |         | F       | p       |
|---------------|---------------|--------------|---------|---------|---------|
| <b>Oticon</b> | <b>Phonak</b> | LIN+NR       | WDRC    | 49.255  | <.0001* |
|               |               | WDRC         | WDRC+NR | 236.141 | <.0001* |
|               |               | WDRC+NR      | LIN+NR  | 45.567  | <.0001* |
| <b>Phonak</b> | <b>Widex</b>  | LIN+NR       | WDRC    | 35.051  | <.0001* |
|               |               | WDRC         | WDRC+NR | 24.010  | <.0001* |
|               |               | WDRC+NR      | LIN+NR  | 28.347  | <.0001* |
| <b>Widex</b>  | <b>Oticon</b> | LIN+NR       | WDRC    | 89.763  | <.0001* |
|               |               | WDRC         | WDRC+NR | 118.574 | <.0001* |
|               |               | WDRC+NR      | LIN+NR  | 8.072   | .009    |

**HI Group Interactions**

| HA Brand      |               | HA Condition |         | F      | p       |
|---------------|---------------|--------------|---------|--------|---------|
| <b>Oticon</b> | <b>Phonak</b> | LIN+NR       | WDRC    | 68.203 | <.0001* |
|               |               | WDRC         | WDRC+NR | .503   | .485    |
|               |               | WDRC+NR      | LIN+NR  | 45.727 | <.0001* |
| <b>Phonak</b> | <b>Widex</b>  | LIN+NR       | WDRC    | 37.349 | <.0001* |
|               |               | WDRC         | WDRC+NR | .185   | .671    |
|               |               | WDRC+NR      | LIN+NR  | 42.055 | <.0001* |
| <b>Widex</b>  | <b>Oticon</b> | LIN+NR       | WDRC    | 28.179 | <.0001* |
|               |               | WDRC         | WDRC+NR | 43.460 | <.0001* |
|               |               | WDRC+NR      | LIN+NR  | 5.032  | .035    |

\*significant at a .0005 level.

**Appendix K. Mean speech perception scores.**



**Appendix L. Contrasts for Interaction of Brand and Condition in the change in SII re: linear scores.**

**NH Group**

| HA Brand      |               | HA Condition |         | F       | p       |
|---------------|---------------|--------------|---------|---------|---------|
| <b>Oticon</b> | <b>Phonak</b> | LIN+NR       | WDRC    | 166.984 | <.0001* |
|               |               | WDRC         | WDRC+NR | 96.468  | <.0001* |
|               |               | WDRC+NR      | LIN+NR  | 52.696  | <.0001* |
| <b>Phonak</b> | <b>Widex</b>  | LIN+NR       | WDRC    | 24.707  | <.0001* |
|               |               | WDRC         | WDRC+NR | 31.906  | <.0001* |
|               |               | WDRC+NR      | LIN+NR  | 10.955  | .0003*  |
| <b>Widex</b>  | <b>Oticon</b> | LIN+NR       | WDRC    | 31.200  | <.0001* |
|               |               | WDRC         | WDRC+NR | 19.911  | <.0001* |
|               |               | WDRC+NR      | LIN+NR  | 11.63   | <.0001* |

**HI Group**

| HA Brand      |               | HA Condition |         | F      | p       |
|---------------|---------------|--------------|---------|--------|---------|
| <b>Oticon</b> | <b>Phonak</b> | LIN+NR       | WDRC    | 6.170  | .020    |
|               |               | WDRC         | WDRC+NR | .881   | .357    |
|               |               | WDRC+NR      | LIN+NR  | 6.467  | .018    |
| <b>Phonak</b> | <b>Widex</b>  | LIN+NR       | WDRC    | 27.776 | <.0001* |
|               |               | WDRC         | WDRC+NR | 4.447  | .047    |
|               |               | WDRC+NR      | LIN+NR  | 18.156 | <.0001* |
| <b>Widex</b>  | <b>Oticon</b> | LIN+NR       | WDRC    | 12.646 | .002*   |
|               |               | WDRC         | WDRC+NR | 10.427 | .004*   |
|               |               | WDRC+NR      | LIN+NR  | 9.058  | .006    |

\*Significant at .0005 level.

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