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# Electroacoustic and Behavioural Evaluation of Hearing Aid Digital Signal Processing Features

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A thesis submitted in partial fulfillment of the requirements for the degree in Doctor of Philosophy

# ELECTROACOUSTIC AND BEHAVIOURAL EVALUATION OF HEARING AID DIGITAL SIGNAL PROCESSING FEATURES

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by

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# Abstract

Modern digital hearing aids provide an array of features to improve the user listening experience. As the features become more advanced and interdependent, it becomes increasingly necessary to develop accurate and cost-effective methods to evaluate their performance. Subjective experiments are an accurate method to determine hearing aid performance but they come with a high monetary and time cost. Four studies that develop and evaluate electroacoustic hearing aid feature evaluation techniques are presented. The first study applies a recent speech quality metric to two bilateral wireless hearing aids with various features enabled in a variety of environmental conditions. The study shows that accurate speech quality predictions are made with a reduced version of the original metric, and that a portion of the original metric does not perform well when applied to a novel subjective speech quality rating database. The second study presents a reference free (non-intrusive) electroacoustic speech quality metric developed specifically for hearing aid applications and compares its performance to a recent intrusive metric. The non-intrusive metric offers the advantage of eliminating the need for a shaped reference signal and can be used in real time applications but requires a sacrifice in prediction accuracy. The third study investigates the digital noise reduction performance of seven recent hearing aid models. An electroacoustic measurement system is presented that allows the noise and speech signals to be separated from hearing aid recordings. It is shown how this can be used to investigate digital noise reduction performance through the application of speech quality and speech intelligibility measures. It is also shown how the system can be used to quantify digital noise reduction attack times. The fourth study presents a turntable-based system to investigate hearing aid directionality performance. Two methods to extract the signal of interest are described. Polar plots are presented for a number of hearing aid models from recordings generated in both the free-field and from a head-and-torso simulator. It is expected that the proposed electroacoustic techniques will assist Audiologists and hearing researchers in choosing, benchmarking, and fine-tuning hearing aid features.



# Keywords

Digital Hearing Aids, Speech Quality, Digital Noise Reduction, Electroacoustic Measures, Reference-free Speech Quality Metric, Speech Intelligibility.

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# Nomenclature

| | |
|---|---|
| ANIQUE+ | Auditory Non-Intrusive Quality Estimation Plus |
| ANOVA | ANalysis Of VAriance |
| ANSI | American National Standards Institute |
| BW | Bandwidth |
| CC | Cepstrum Correlation |
| CR | Compression Ratio |
| DHA | Digital Hearing Aid |
| CST | Connected Speech Test |
| DNR | Digital Noise Reduction |
| DSL | Desired Sensation Level |
| DSP | Digital Signal Processing |
| DTC | Dedicated Test Chamber |
| FBR | Front-to-Back Ratio |
| FIR | Finite Impulse Response |
| HA | Hearing Aid |
| HASQI | Hearing Aid Speech Quality Index |
| HATS | Head And Torso Simulator |
| HI | Hearing Impaired |
| HL | Hearing Loss |
| I/O | Input/Output |
| IEEE | Institute of Electrical and Electronics Engineers |
| IHC | Inner Hair Cell |
| IIT | Impulse Invariant Transform |
| ITU | International Telecommunication Union |
| KEMAR | Knowles Electronic Manikin for Acoustic Research |
| LLR | Log-Likelihood Ratio |
| LPC | Linear Predictive Coding |
| MUSHRA | Multiple Stimulus with Hidden Reference and Anchors |
| OHC | Outer Hair Cell |
| PC | Personal Computer |
| PEMO-Q | PErceptual MOdel - Quality |
| SII | Speech Intelligibility Index |
| SNHL | SensoriNeural Hearing Loss |
| SNR | Signal-to-Noise Ratio |
| SPL | Sound Pressure Level |
| SRMR-HA | Speech-to-Reverberation Modulation Ratio - Hearing Aid |
| SSN | Speech shaped Noise |
| SSQ | Speech, Spatial and Qualities |
| USB | Universal Serial Bus |
| VAD | Voice Activity Detection |
| WDRC | Wide Dynamic Range Compression |

# Chapter 1

# 1 Introduction

Hearing impairment is a condition that affects many people throughout the world. Hearing loss is the third most common chronic disability observed in older adults [1]. It is estimated that about 10% of the general population suffers from hearing impairment and both the incidence and prevalence of hearing impairment increases with age [1]. As the average age of the population increases, hearing impairment will become an increasingly common problem. Accurate and comprehensive assessment of the auditory function and appropriate therapeutic intervention are crucial for enhancing the communicative ability and restoring good quality of life for affected persons.

## 1.1 Hearing Aids & Their Features

Hearing aids form the most common treatment modality for listeners with mild to severe degrees of hearing loss. Over the past four decades, hearing aids have evolved significantly from simple analog amplifiers to sophisticated and intelligent computing machines that incorporate an array of digital signal processing (DSP) features [2]. As an example, Figure 1-1, shows the block diagram of a pre-configured DSP system, AYRE SA3291 [3], recently introduced for use in commercial hearing aids.  Starting with the microphone inputs on the left (pins #1 and #2 respectively), the input to the hearing aid passes through many computational blocks before presentation to the impaired ear (the main signal path is highlighted in green in Figure 1-1). The focus of this research is on the impact of three hearing aid features shown in Figure 1-1, namely the adaptive directional microphone, the Wide Dynamic Range Compression (WDRC), and the noise reduction blocks. These feature blocks have been highlighted in red in Figure 1-1 to give an idea of where these features fit in with the overall signal processing, and their functional description is given in the following sections.

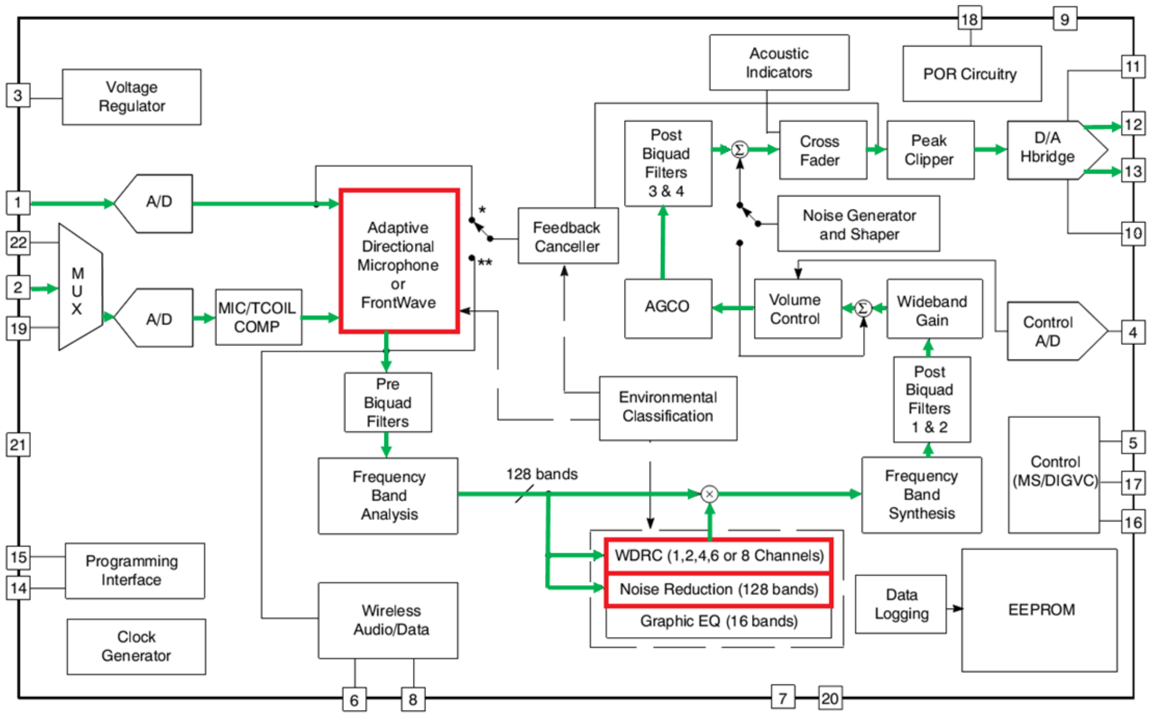



**Figure 1-1: A block diagram of the AYRE SA3291 DSP system for hearing aids [3].**

### 1.1.1 Multichannel Wide Dynamic Range Compression

Hearing loss typically involves a reduction in the ability to perceive lower level sounds, while the highest level that can be perceived without discomfort remains unchanged [4]. This means that the range of detectable sound levels for an individual with a hearing loss is reduced and the information contained in the undetected lower level sounds is lost. The purpose of a WDRC algorithm is to compress the range of levels that are detectable by normal hearing individuals into the detectable range of the Hearing Impaired (HI) individual. This is accomplished by applying a larger gain to low level sounds and reducing the gain applied as the level of the sound increases in such a way that the higher level sounds are never amplified to levels of discomfort. The result of this operation is that the HI individual will suffer less from a loss of the information contained in the low level sounds and will still be able to listen to the high level sounds comfortably [4], [5]. In a multichannel WDRC system, the level-dependent gain is applied independently in different frequency regions, based on the hearing loss profile of the wearer.

Several parameters characterize the functionality of a multichannel WDRC system, including: (a) the number of compression channels, (b) the input level threshold or knee-point for compressor activation, (c) the amount of compression applied, and (d) the reaction times to a sudden increase or decrease in input level (termed attack and release times respectively). The interested reader is referred to review articles by Dillon [4] and Souza [5] for a more detailed description of multichannel WDRC systems.

### 1.1.2 Multiband Adaptive Directionality

While multichannel WDRC has been shown to provide significant benefit in quiet listening environments [4], [5], it may also have a detrimental effect in noisy environments by increasing the levels of background noise [6]. Since HI listeners consistently rank poor understanding of speech in noisy environments as the number one problem associated with their hearing loss [7], modern digital hearing aids (DHAs) employ additional signal processing such as multiband adaptive directionality to improve the signal-to-noise ratio (SNR). The idea behind adaptive directionality is to use directional microphones to eliminate unwanted sound signals based on their angle of arrival at the listener. In modern DHAs, directional microphones are typically implemented by combining two or more omnidirectional microphones, as shown in Figure 1-2.

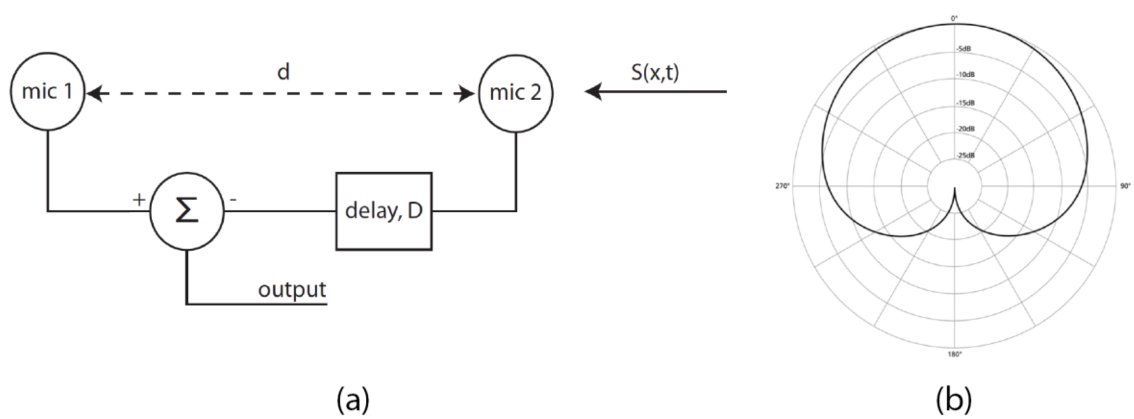



**Figure 1-2: (a) A simple representation of a directional microphone implemented using two omnidirectional microphones; (b) cardioid polar plot.**

Here, the incoming sound, $S(x,t)$, where $x$ is the position variable and $t$ is the time variable, experiences a natural delay between *mic 2* and *mic 1* given by $d/c$, where $c$ is the speed of sound in the given environment and $d$ is the distance between the microphones. By adjusting the value of the electronic delay, $D$, to be equal to $d/c$, the incoming signal is completely cancelled by the summation block since the two microphone outputs will be aligned. Due to the fact that signals arriving from other directions will have a different natural delay value, they will not be cancelled by the combination of the electronic delay and summation blocks. The ensuing directional response can be depicted using a polar plot [8], as shown in Figure 1-2b, which displays the amount of attenuation imparted by the directional system for sounds arriving from different incident angles.

In adaptive directionality, the electronic delay is adjusted by the DHA such that an optimal polar plot (one that attenuates the noise signal the most), is selected [8], [9]. Examples of common polar plots that adaptive directional hearing aids implement are shown in Figure 1-3. Multiband adaptive directionality is a further enhancement where separate and independent directional patterns can be realized in different frequency regions [9], which allows for simultaneous suppression of spatially- and spectrally-separated noise sources. For further exploration of the adaptive directional DHA technologies and related issues, the reader is referred to an excellent review by Ricketts [8].

### 1.1.3 Digital Noise Reduction

As directional microphone processing exploits the spatial separation between desired and undesired signals, an alternative strategy is required when the desired and undesired signals are spatially close. Furthermore, smaller hearing aid form factors (such as the completely in the canal (CIC) models) allow space for only a single microphone.

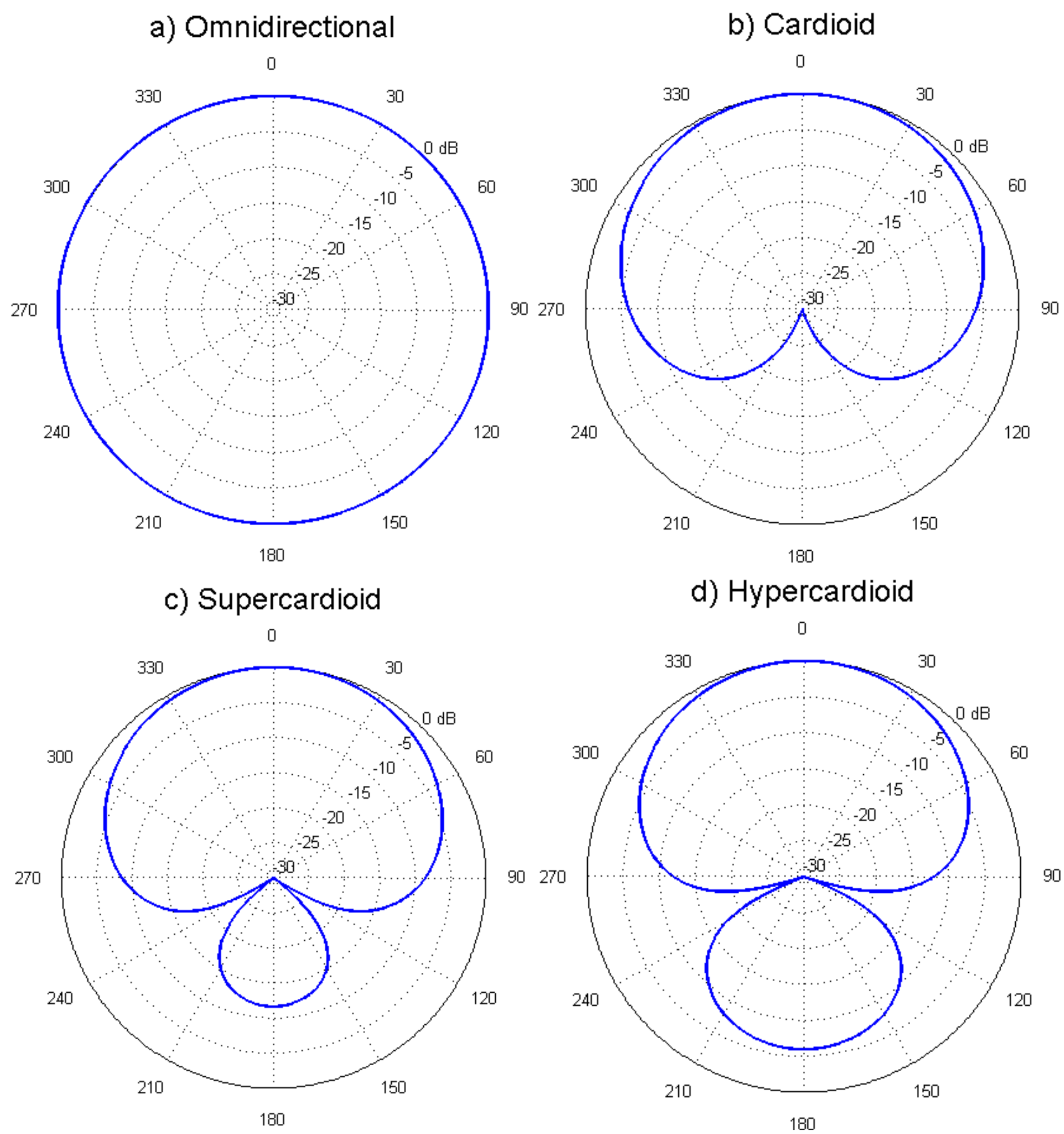



**Figure 1-3: Common polar plots used in hearing aid directionality: a) Omnidirectional, b) Cardioid, c) Supercardioid, d) Hypercardioid.**

The single-microphone algorithm typically employed by modern DHAs to reduce the noise energy is referred to as Digital Noise Reduction (DNR). Though the exact approach used in commercial DHAs is proprietary information, Figure 1-4 shows the block diagram of a typical DNR algorithm based on a spectral subtraction approach [10]. Here, the input mixture of speech plus noise signal is applied to a Fourier transform block and a Voice Activity Detection (VAD) block. The VAD block controls a switch which is open when speech is detected and closed when speech is not detected. The result of this configuration is that the averager will store an estimate of the noise spectrum which is subtracted from the speech-plus-noise magnitude spectrum even when speech is present in the source signal. To produce the output signal, the reduced noise magnitude spectrum and the phase output of the Fourier transform are applied to the inverse Fourier transform block. This algorithm is commonly implemented in a sub-band form in many modern hearing aids [10].

As mentioned before, implementation details of a DNR algorithm are proprietary and differences do exist among the DHAs on the voice activity detection procedure, the amount of noise reduction, the time constants for activation and deactivation of the noise reduction algorithm, and the interaction with other signal processing algorithms. The reader is referred to a review article by Bentler et al. [11] for further discussion of the DNR algorithms in modern DHAs.

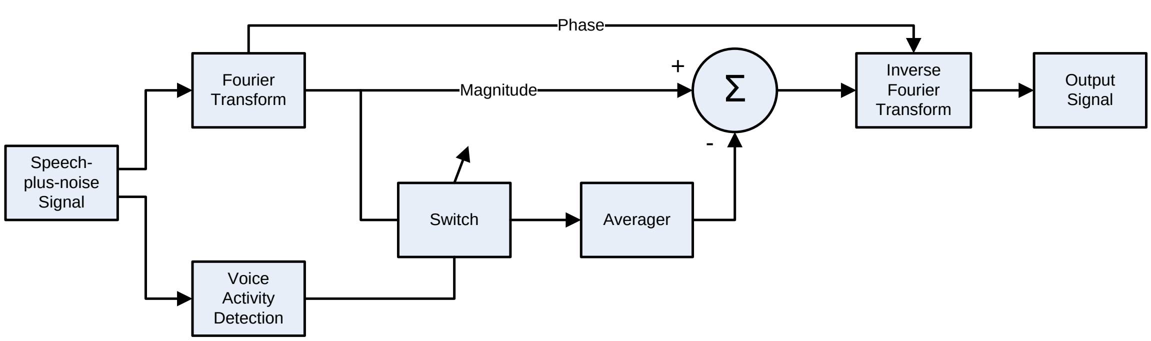



**Figure 1-4: Block diagram of a typical spectral subtraction system [10].**

### 1.1.4 Monaural and Bilateral Hearing Aids

It is pertinent to distinguish here between monaural and bilateral hearing aids. Hearing loss in only one ear is treated with a single hearing aid, which is termed as a monaural hearing aid fitting. In contrast, bilateral hearing aid fittings consist of a hearing aid on each ear. There is evidence that the proportion of bilateral fittings has increased over the past few years. A market survey presented in [12] has shown that bilateral hearing aid fittings constitute about 74% of all hearing aid fittings (up from 69.3% in 2004), and 86% of all bilateral hearing loss patients. A sub-class of the bilateral hearing aids, the so-called bilateral wireless hearing aids, have been introduced by a few hearing aid manufacturers (Oticon - "Binaural Broadband", Siemens - "e2e wireless"). Unlike the traditional bilateral hearing aids, where the two hearing aids apply the digital signal processing strategies independently, the bilateral wireless hearing aids communicate with each other wirelessly and collectively process the left and right acoustic inputs in a co-ordinated manner [13], [14].

Given the prevalence of bilateral fittings and the differences in the configuration and signal processing strategies in hearing aids from various manufacturers, it is imperative to measure and benchmark the performance of bilateral hearing aids, so Audiologists may prescribe, fit, and verify appropriate hearing aid technologies. In this research, the performance of unilateral or bilateral hearing aids is measured using parameters related to speech intelligibility, sound quality, and sound localization, which are introduced in the following section.

## 1.2 Impact on Speech Intelligibility, Sound Quality and Sound Localization

With the variety of signal improvement techniques discussed in Section 1.1, it is important to consider methods to identify and quantify the benefit that is provided to the hearing aid user.

As can be seen in Figure 1-5, each of the three hearing aid features introduced in the previous section can impact numerous aspects of hearing. This work focuses on the impact of hearing aid features on speech intelligibility, speech quality and sound localization. The methods of assessment shown in Figure 1-5 can be divided into two categories. Subjective assessment involves the participation of human subjects, whereas instrumental assessment can be accomplished through electroacoustic measurements of DHA performance. Figure 1-5 lists some common methods of assessment for both the subjective and instrumental categories. The remainder of this section will provide a brief overview of the subjective assessment of the impact of the three DHA feature blocks that are of interest to this thesis.

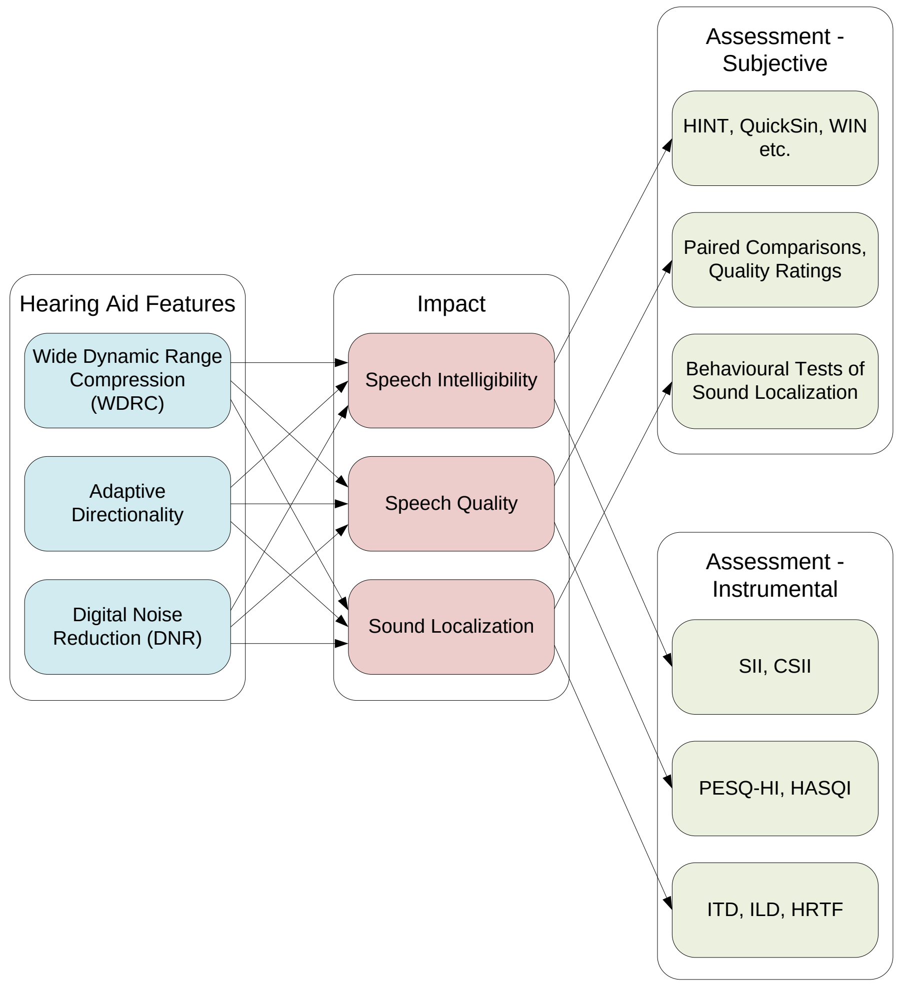



**Figure 1-5: An overall of picture of the impact that the relevant hearing aid features to this study have on hearing and common methods of impact assessment.**

### 1.2.1 Speech Intelligibility

Speech Intelligibility refers to the ability of an individual to comprehend a speech signal. Intelligibility can be a particular issue for HI individuals when significant portions of the speech signal energy fall outside of the audible range. A number of subjective measures of speech intelligibility have been developed and employed in past studies, and more commonly used methods are described below.

Speech intelligibility can be assessed at the sentence, word, or phonemic level. Within the context of DHAs, commonly used sentence-level speech intelligibility tests include: (a) the Hearing in Noise Test (HINT) [15], wherein sets of phonetically balanced sentences are presented in spectrally-matched background noise, and the SNR at which the subjects understand 50% of the sentences is the performance indicator; (b) the Connected Speech Test (CST) [16] wherein passages containing conversational speech sentences are presented in multi-talker babble, and the number of correctly identified scoring words is quantified as the performance metric; and (c) the Quick Speech-In-Noise (QuickSIN) test [17] and its longer version, SIN test [18], wherein sentences from the IEEE database [19] are presented in a background noise of 4-talker babble at varying SNRs, and the subject's performance is quantified as the "SNR-loss" - the SNR required by the HI individual above the SNR needed by a normal hearing individual to obtain 50% correct sentence identification. An example of the word-level intelligibility test is the Words-in-Noise test (WIN) [20]. Rather than testing sentence level recognition, the WIN test presents monosyllabic words combined with multi-talker babble which removes the contextual cues present in sentence level tests such as HINT. Wilson et al. [21], found that among the four different recognition tests, QuickSIN and WIN provided the greatest separation in recognition performance between normal hearing and HI individuals.

The impact of multichannel WDRC on speech intelligibility has been extensively investigated. A 2002 paper by Souza [5] included a review of the previous literature relevant to this topic. The overall observation of the author based on the reviewed studies was that WDRC was most effective in comparison to linear amplification for low-level speech in quiet. No clear advantage was identifiable for speech-plus-noise signals. Souza [5] also notes that increasing the number of compression channels may have a detrimental effect on speech intelligibility. Since this review, a number of further studies have been conducted to measure the effect of compression on intelligibility. Rosengard et al. [22], found that WDRC offered an improvement in intelligibility for moderate and flat simulated hearing losses. No improvement was observed for sloping, mild to moderate losses. In addition it was shown that increased compression ratios resulted in reduced sound quality leading to the assertion that to maximize satisfaction with WDRC both intelligibility and quality should be considered. Stone and Moore [23], investigated the effect of compression speed and showed that increased compression speed and channels caused a decrease in intelligibility. A follow-up study by Moore et al. [24], included an investigation of the effect of compression speed on intelligibility but for a competing-speech task. The results showed that for hearing impaired listeners, the slow acting compression resulted in mild but significant improvement in scores when compared to fast acting compression for spatially separated stimuli. This effect was not observed when the stimuli were co-located. To summarize, multichannel WDRC enhances speech intelligibility in quiet, but not in background noise. Moreover, the compression ratio, time constants, and number of compression channels can impact intelligibility.

There is substantial evidence that directional microphones enhance speech intelligibility, at least in laboratory environments (see reviews in [8], [25]). As an example, Blamey et al. [26], compared perception in noise results with DHAs in omnidirectional, supercardioid, and adaptive directional microphone configurations. The study included a number of noise conditions and found that in all cases the use of the adaptive directional microphone yielded the best speech perception scores. More recently, Magnusson et al. [27], found a modest but significant improvement in speech recognition with the use of directional microphones compared to the unaided case with open-fit DHAs. The use of an omni-directional microphone did not show a significant improvement in speech recognition in comparison to the unaided case. Mackenzie and Lutman [28] investigated speech recognition for bilateral hearing aid fittings where the adaptive directional systems are acting independently. The study found that use of the directional microphones still provided a benefit with respect to speech recognition, despite their independent operation.

Investigations into the impact of DNR on speech intelligibility have generally shown neither improvement nor degradation (a more detailed literature review is presented in Chapter 4). As an example, Hu and Loizou [29] investigated the effect of noise reduction algorithms on intelligibility with normal hearing listeners. The study found that for all but one noise condition (car noise at 5 dB SNR), no improvement in intelligibility was provided. Another important observation from this study was that algorithms that had previously performed well in terms of speech quality, were found to have worse performance than other algorithms in terms of intelligibility.

Kim and Loizou [30] conducted a study of the impact on intelligibility of specific types of distortion introduced by noise reduction algorithms. The authors suggest that by limiting the distortion caused by over-estimating the signal amplitude, speech intelligibility improvements may be achieved through the use of noise reduction algorithms.

In summary, a review of the literature has shown that WDRC can offer improvements in intelligibility for specific signal conditions, adaptive directionality can offer improved intelligibility for most signal conditions and there is limited evidence that suggests any improvement in intelligibility offered by DNR algorithms.

### 1.2.2 Sound Quality

While speech intelligibility is a measure of speech comprehension, sound quality refers more to the overall listening experience. Sound quality is quite subjective in nature and can therefore be challenging to accurately quantify. Examples of properties that affect sound quality include:

- Clarity of the sound
- Naturalness of the sound
- Richness or fidelity of the sound

When referring to the sound quality of speech signals, it is common practice to use the term *speech quality*. Subjective speech quality evaluation techniques have been used in this research for the purpose of comparison with electroacoustic measures.

Speech quality ratings can be obtained through markings on a visual analog scale representing a speech quality attribute [31], through paired comparisons [32], [33], and through the MUltiple Stimuli with Hidden Reference and Anchor (MUSHRA) [34]. The latter methodology is used in this thesis, which involves presenting a subject with a known reference signal, a hidden reference signal, a hidden anchor signal and multiple test signals such that the subject can compare and rate the quality of each signal in a relative fashion. The purpose of the hidden reference is for estimating the reliability of the ratings through comparison to the rating provided for the known reference and the purpose of the hidden anchor is to have a low quality reference that will deter the subject from giving low ratings to test signals with minor imperfections. MUSHRA allows for ratings between zero and one hundred which allows subjects to provide precise sound quality opinion scores.

Past studies have investigated the impact that hearing aid features have on sound quality. In the previously mentioned 2002 paper by Souza [5], the literature related to the effect of WDRC on speech quality was reviewed. The author noted that patients generally preferred simple signal processing techniques in comparison to more complicated techniques that incorporated a higher number of processing channels, greater compression ratios, and faster time constants. WDRC was more often preferred when compared to compression techniques that cause greater signal distortion such as peak clipping. The author found that increased speech quality ratings were correlated with increased speech intelligibility ratings which supports the idea that compression hearing aids can be fit clinically to maximize sound quality without detrimentally affecting speech intelligibility.

Bentler [25], reviewed nine previous studies that investigated the effectiveness of directional microphones in hearing aids. The overall conclusion of the review was that directional microphones offer an advantage over the use of amplification only and this advantage is maximised when a user controlled switch is included and users are trained on the environments that are best suited to directional microphone use. Mackenzie and Lutman [28] reported improved sound quality with the use of directional hearing aid modes. In particular, improvements in user ratings for comfort and clarity were observed. Amlani et al. [32] assessed the speech clarity associated with the DHA output when configured as an omnidirectional or directional microphone with hypercardioid or cardioid polar pattern.  Results showed better speech clarity judgements for the directional microphone condition (in either polar plot) over the omnidirectional condition.

The effect of DNR on sound quality has been investigated in a number of different studies. This is reviewed in more detail in Chapter 4, but the overall impression is that DNR offers improvements to sound quality in specific situations.

### 1.2.3 Sound Localization

Sound localization refers to the ability of a listener to determine the angle of incidence of a specific sound. In the horizontal plane, this is accomplished by exploiting differences between sound signals received at each of the two ears. For lower frequency sounds, below approximately 1500 Hz, the listener primarily exploits the time difference of arrival of the sound at each ear, termed the Interaural Time Difference (ITD). For sounds above 1500 Hz, it is the level difference, termed the Interaural Level Difference (ILD), that is exploited. The details of the sound localization process are further explained in 5.1.1, for now it is important to note that sound localization is an important part of listening as it allows for the focus of the listener to be adjusted to the appropriate direction.

A review of the literature on sound localization ability reveals no shortage of previous studies on this topic. Populin [35] reviewed past studies which made use of various methods to subjectively evaluate sound localization. The methods mentioned include verbal source location reporting, identification of sound sources, head pointing, pointing with an instrument such as a gun, or stylus and aiming a laser beam.

Hearing aid users that make use of the features outlined in 1.1, may find that their sound localization ability is impaired due to a distortion of ITD and ILD cues. Since WDRC clearly impacts the signal level in a non-linear fashion, independently acting hearing aids worn on each ear have the potential to distort the level differences that are important for sound localization. Keidser et al. [36], conducted a detailed study of the impact of WDRC, noise reduction and directional microphones on sound localization ability. Though it was found that WDRC and noise reduction impacted the ILD when users wore bilaterally fitted hearing aids, no significant difference in sound localization ability was found. In contrast, it was found that left/right confusions increased when there was a mismatch in the directional microphone attenuation pattern between the two ears for bilaterally fitted hearing aids. Table 1-1 outlines the effects that the various DSP features had on the studied sound localization cues.

**Table 1-1: An overview of the impact that hearing aid features have on cues used for sound localization. Reproduced from [36].**

| Signal Processing | ILD | ITD | Spectral Shape |
|---|---|---|---|
| Multi-channel WDRC | Yes | No | Yes |
| Noise reduction | Yes | No | Yes |
| Directional Microphone | Yes | Yes | Yes |
| Adaptive Directionality | Yes | Yes | Yes |

Based on these results, it appears that the greatest gain in the sound localization ability of hearing aid users could be achieved by refining the adaptive directionality feature in bilateral DHAs.

## 1.3 Need for Electroacoustic Measures

While it is customary to measure speech intelligibility, sound quality and sound localization performance through subjective listening tests as they have high face validity, they are also time and resource consuming. Electroacoustic (instrumental) measures that are obtained from hearing aid recordings and have a high degree of correlation with subjective data are therefore attractive.

The current standards for electroacoustic measurements of DHAs are primarily used for quality control and functional assessment [37], [38]. For example, the American National Standards Institute (ANSI) S3. 22 standard for hearing aids specifies procedures for measuring the maximum output sound pressure level (SPL), level-dependent frequency response characteristics, input/output functions, and attack and release time constants. Distortion in the DHA is quantified using Total Harmonic Distortion (THD) at specific frequencies. While these measurements ensure basic functionality of the DHA and that the DHA is performing within the limits of the manufacturer's specification, they do not proffer any information on the HI wearer's perception of the DHA performance. Similarly, ANSI S3.35 [39] describes procedures for mannequin based measurements of DHA performance, including the measurement of polar patterns and the directivity index. Once again, these measurements do not provide information or insight into the impact of DHAs on the aforementioned speech intelligibility, quality, and localization.

Standardized methods of advanced electroacoustic evaluation do exist for some cases. For example, the Speech Intelligibility Index (SII) is defined by the ANSI S3.5-1997 standard [40] which outlines a procedure for speech intelligibility prediction through analysis of the speech and noise spectrums of a signal of interest. Further details on the SII are provided in [40]. One disadvantage of the SII is that it does not account for signal distortions which may impact intelligibility. To address this issue, an electroacoustic measure known as the Coherence SII (CSII) was developed in [41]. This measure uses the coherence between a processed signal and a reference signal to compute a Signal-to-Distortion Ratio (SDR) and replaces the SII with the SDR in the SII calculation procedure. The study shows that by splitting the signal of interest into low, mid and high level regions, computing the CSII for each region and then linearly combining the three scores into a single overall score, high levels of correlation with subjective ratings of distorted signals are achieved.

While the International Telecommunication Union (ITU) has developed standards for measuring the quality of telephone-quality speech as well as broadband audio [42], [43], no such standards exist for hearing aids despite their attractiveness and need. Published research strategies for predicting the quality of hearing aid processed speech include: the HI version of the ITU standardized Perceptual Evaluation of Speech Quality [44], and the Hearing Aid Speech Quality Index (HASQI) [45], [46].

Standardized methods to electroacoustically evaluate sound localization performance are also lacking. As outlined in 5.1, methods to measure the attenuation pattern of hearing aids that make use of an adaptive directionality feature have been developed for specific hearing aid configurations. Based on the conclusion in [36] that adaptive microphones have the most significant impact on the sound localization ability of hearing aid users, it appears that these measurements should prove to be relevant in examining the degree to which the ILD and ITD cues are preserved and consequently predicting the impact that adaptive directionality has on sound localization ability.

## 1.4 Problem Statement & Thesis Scope

Since the introduction of DHAs, hearing aid manufacturers have continued to release updated models that incorporate new features and expand the capabilities of existing features. This has resulted in a relatively rapid evolution of DHAs to the point where there is currently a profusion of DHA models available to choose from. It is quite common for a particular manufacturer to offer DHAs at multiple price points, where the number of features and the sophistication of the features correlate with the offered price of purchase. For many HI individuals, the cost of purchasing DHAs is not insignificant. Therefore, it becomes quite important to quantify the benefit that is offered to the end user by any increased cost that is considered to obtain an enhanced DHA feature set.

As outlined in the previous sections, individual hearing aid features have the potential to degrade certain aspects of sound perception. For example, WDRC can reduce the user's sound localization ability by distorting the Interaural Level Difference (ILD) that is imperative to sound localization at frequencies above 1500 Hz. This further complicates the choice of DHA model and supports the need for advanced DHA evaluation techniques. Similarly, as discussed in previous sections, DHA features can impact both speech intelligibility and quality.

As introduced in section 1.3, electroacoustic measures of hearing aid performance are attractive due to their relatively low cost when compared to subjective based measures. The aim of this thesis is to address the electroacoustic evaluation of modern DHAs with a particular focus on three features; directionality (in some cases adaptive), DNR and bilateral WDRC. Four electroacoustic evaluation techniques are proposed as outlined in the following section.

## 1.5 Thesis Organization

Each of the following four chapters presents one of the developed electroacoustic methods for the evaluation of DHA performance. Chapter 2 outlines a study that focuses on the use of speech quality prediction algorithms to assess the performance of bilateral wireless hearing aids under a number of different operating conditions. Chapter 3 presents a similar study where a novel reference free electroacoustic hearing aid sound quality measure is presented and compared to subjective ratings under a variety of listening conditions. Chapter 4 describes a new electroacoustic approach to evaluating DNR performance. Chapter 5 details a turn-table based approach to evaluating the performance of DHA adaptive directionality algorithms. Finally, Chapter 6 includes a summary, key contributions and an overview of potential future work.

# Chapter 2

# 2 Bilateral Wireless Hearing Aid Sound Quality Assessment

This first study on the electroacoustic evaluation of DHA performance focuses on two bilateral wireless hearing aid models. The approach taken involves comparing sound quality estimates derived from two electroacoustic evaluation techniques with subjective sound quality scores. A particular point of interest for this study was to assess the impact of wireless synchronization of bilateral DHAs on sound quality.

## 2.1 Motivation

The quality of DHA processed sound is directly linked to the level of acceptance by DHA users. MarkeTrak surveys of HI listeners [7], [12], [47] have consistently ranked sound quality highly on the overall list of desirable DHA characteristics. For example, the most recent MarkeTrak survey [7] placed three aspects of sound quality, namely the clarity of sound, how natural sounding it is, and the fidelity of sound, within the top six most important DHA performance factors related to the user acceptance level with a particular hearing device. Based on this evidence, it is clear that the DHA sound quality plays an important role in wearer satisfaction and continued use of the device.

In past studies, the impact of a number of hearing aid processing characteristics on sound quality has been investigated. This includes additive noise and peak clipping [48], [49]; time constants, compression ratio, the number of channels in multichannel WDRC [50]–[52], and the impact of digital noise reduction (DNR), directional processing, speech enhancement (SE), and feedback cancellation [32], [33], [53], [54].

Given the subjective nature of sound quality, the most accurate form of measurement has traditionally been the collection of ratings from a group of HI subjects. The disadvantage of this approach is that it is quite time consuming and requires significant resources. In contrast, objective instrumental methods allow for convenient rating estimation. The challenge with sound quality estimation is to match the ratings provided by the subjective approach in an accurate and robust manner. Previously developed techniques take the approach of modeling the human auditory system, extracting features that are deemed to be relevant to sound quality and combining these features in an optimal fashion to produce an overall quality score. For other applications such as telephone speech and broadband audio, the ITU has developed standards to estimate speech and audio quality [43], [55]. These standards, for example, can be used to gauge the impact of speech and audio coders, noise reduction and echo cancellation algorithms, and telecommunication and broadcasting equipment on perceived sound quality. As of yet, no such standards exist for DHAs despite the significant potential benefits outlined above. Published research strategies for predicting the quality of DHA processed speech include: a metric based on the ITU standardized Perceptual Evaluation of Speech Quality (PESQ) [56], and the Hearing Aid Speech Quality Index (HASQI) [45], [46].

PESQ [27] is a widely used speech quality estimation method standardized by the ITU for telephony applications. The PESQ score is computed through comparing the test signal (i.e. the speech stimulus whose quality needs to be estimated) and a clean version of the test signal in feature space. This is accomplished through three steps: a time alignment step, a feature extraction step, and a feature mapping step. In the time alignment step, the test and reference signals are temporally aligned. Features are then extracted through a time– frequency analysis procedure incorporating two steps based on auditory perception: (a) transformation of the linear frequency axis to the Bark scale, which accounts for the finer frequency resolution at lower frequencies than higher frequencies, and (b) transformation of the amplitude values to "loudness" values according to Zwicker's loudness formula [27]. The differences in the resulting perceptual features from the test and reference signals are assimilated to produce the PESQ score. Beerends et al. [56] described a modified version of PESQ, termed PESQ-HI, for applications to hearing aids. These modifications include the adaptation of time-frequency processing and feature mapping models to better match "the behaviour of HI subjects" [56]. However, the details on how this was accomplished were not sufficiently explained.

More recently, Kates and Arehart [15] presented the HASQI as an alternative speech quality estimator. HASQI models the human auditory system for both normal and HI listeners. HASQI computes two indices after the application of a cochlear model of hearing loss to the test and reference signals: a nonlinear index that attempts to capture the impact of noise and nonlinear distortion, and a linear index that aims to capture the effects of spectral shaping. The final HASQI value is a product of these two indices. Arehart et al. [57] validated HASQI using speech quality ratings obtained from HI listeners for speech stimuli processed through a simulated hearing aid operating in a variety of linear, nonlinear, and noisy conditions. Arehart et al. [29] reported correlation coefficients of 0.96 between HASQI and subjective ratings, indicating a high degree of concurrence between the objective metric and subjective data. In a follow-up study, Arehart et al. [58] reported a correlation coefficient of 0.91 between HASQI and quality ratings of music stimuli obtained from HI listeners. In an independent study, Kressner et al. [59] compared the performance of HASQI to a number of the speech quality metrics (including PESQ) in predicting the quality ratings of speech processed by different noise reduction algorithms. Results from their study revealed that both HASQI and PESQ produced statistically similar performance. Thus, HASQI appears to be a viable solution for instrumental assessment of hearing aid speech quality, but its performance with data from real hearing aids incorporating state-of-the-art processing algorithms and operating in real environments has not been investigated.

One such processing strategy that has not undergone thorough scientific investigation is the synchronization of bilateral DHAs through wireless communication. As introduced in section 1.1.4, the bilateral wireless hearing aids communicate with each other and collectively process the left and right acoustic inputs in a coordinated manner [13], [14]. The rationale behind this co-ordination is to preserve the naturally occurring timing and level differences between the left and right DHAs, thereby conveying a more naturalistic acoustic scene to the listener. Smith et al. [13] conducted a study involving 20 HI listeners wearing Siemens bilateral wireless DHAs. After wearing the DHAs for eight weeks each in linked or unlinked mode, HI participants were asked to fill out the Speech, Spatial and Qualities (SSQ) of Hearing Scale. Analysis of the SSQ data revealed that most subjects preferred the linked condition over the unlinked condition. Sockalingam et al. [14] investigated the performance of Oticon bilateral wireless DHAs with 30 HI participants. These authors found significant improvements in sound localization and sound quality ratings in certain environments when the bilateral coordination was activated. It must be noted that the previous two studies were conducted by the respective DHA manufacturer and published in trade journals. As such, independent verification of the impact of bilateral wireless coordination on perceived speech quality is warranted.

In summary, the quality of DHA processed speech is of paramount importance for wearers, with implications on continued use of and satisfaction with DHAs. Speech quality is typically assessed through subjective testing; this was especially true for newer DHA processing strategies such as bilateral wireless communication. Instrumental measures of DHA speech quality offer several attractive features: efficient DHA testing, benchmarking different DHA processing algorithms and strategies, and fine tuning of DHA processing parameters. But before an instrumental metric can be relied upon, it must be proven to serve as a reasonable surrogate for subjective judgements accrued with different DHA settings. As such, this study was devised to answer the following questions: (a) Do speech quality judgments, as proffered by HI listeners, differ among brands of bilateral wireless DHAs and between linked and unlinked conditions? (b) What additional impact do variables such as DHA processing features, noise, and reverberation have on perceived speech quality? (c) Does a speech quality metric such as HASQI correlate with subjective judgments of speech quality by HI listeners with more realistic speech stimuli?

## 2.2 Speech Quality Metrics

This section provides a more detailed description of HASQI computational steps. Before embarking on that, a description of a traditional speech quality metric is given, which is used for comparative purposes.

### 2.2.1 Log-likelihood Ratio (LLR)

The LLR is a classic method used to measure the difference between two speech signals. It is based on the Linear Predictive Coding (LPC) representation of a speech signal. Given a clean input signal to the DHA, $x(n)$, and the corresponding DHA output signal, $y(n)$, the LLR is defined as follows:

$$LLR = log\left[\frac{a_y R_x a_y^T}{a_x R_x a_x^T}\right] \tag{2.1}$$

where $a_x$ is the LPC coefficient vector for $x(n)$, $a_y$ is the LPC coefficient vector for $y(n)$ and $R_x$ is the autocorrelation matrix of the original signal.

To interpret the meaning of the LLR, it is useful to consider the numerator and denominator of the fraction separately. As can be seen, the denominator is a function of the input signal autocorrelation matrix and the input signal LPC coefficient vector. This gives the energy of the error between the input signal and the LPC based estimation of the input signal. The numerator is similar to the denominator except that the input signal LPC coefficients are replaced with the output signal LPC coefficients. This gives the energy of the error that results from applying the input signal to the output signal LPC model which will always be greater than the denominator. This error can originate from any noise, distortion or non-linear processing within the DHA and its magnitude will be inversely correlated with the similarity between the input and output signal. The denominator is included as a normalizing factor to account for the fact that the similarity measure should be independent of the LPC performance [60].

### 2.2.2 HASQI

The HASQI speech quality metric seeks to model both linear and nonlinear effects on an input speech signal caused by hearing aid signal processing. As will be seen in this section, for the standard HASQI computation, linear and nonlinear models are designed to be independent of each other and are combined at the final stage of the quality estimation to produce an overall score.

Figure 2-1 displays the computational chain for the noise and nonlinear distortion index portion of the HASQI model.  As shown in the figure, the computation is carried out in two stages - a cochlear model stage and the cepstral correlation stage.

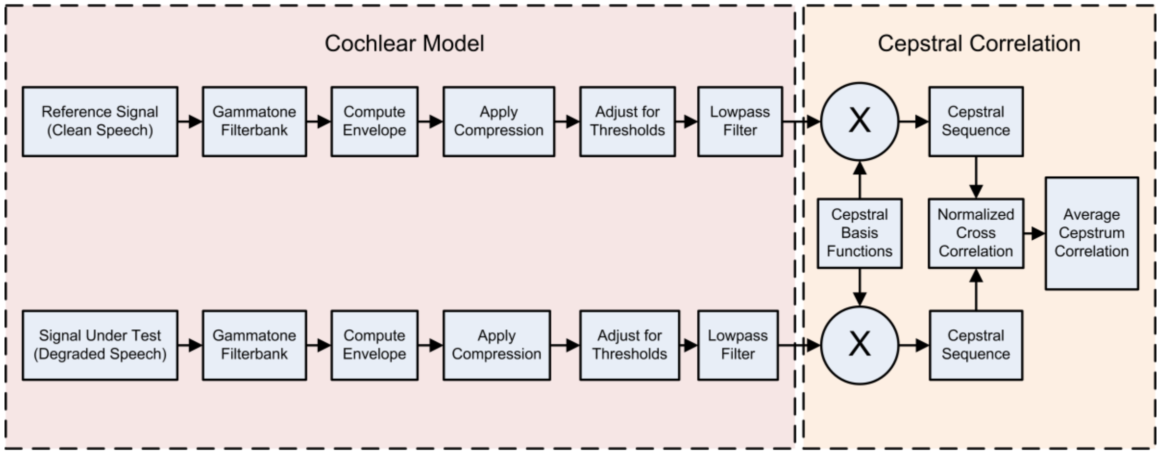



**Figure 2-1: Diagram of the HASQI computational procedure [45], [49].**

In the cochlear model stage, both the clean and processed speech samples are passed through a gammatone filterbank [61] - a parallel filterbank mimicking the auditory filtering behaviour. Broadening of the filter bandwidth due to Sensorineural Hearing Loss (SNHL), a common type of hearing loss typically associated with defects in the cochlear nerve or the inner ear, is incorporated into the model using the following equation:

$$BW_{Increase} = 1 + \left(\frac{OHC_{Loss}}{50}\right) + \left[\left(\frac{OHC_{Loss}}{50}\right)^6\right] \tag{2.2}$$

where $BW_{Increase}$ is broadened bandwidth (BW) and $OHC_{Loss}$ is the portion of total hearing loss due to outer hair cell (OHC) damage. The $OHC_{Loss}$ component also determines the model parameters simulating the compressive behaviour of the basilar membrane in each channel. Both the knee point and the compression ratio (CR) are computed independently in each gammatone channel based on the user's audiogram. The Input/Output (I/O) curve thus derived is applied to the envelope extracted from the filtered signal in each channel. The modified signal envelope is further attenuated by the loss due to Inner Hair Cell (IHC) damage.

The total hearing loss (HL), as specified in the Audiogram, is apportioned between OHC and IHC components as follows: (a) for mild to moderate hearing losses, 80% of the total HL is attributed to OHC damage, with the rest ascribed to IHC damage, and (b) for more severe losses, the OHC and IHC damage is limited across analysis frequencies as a function of the compression ratio. The attenuated envelope in each channel is converted to dB and its values below the hearing threshold are set to zero, which simulates perceived loudness and audibility respectively [45], [46].

The smoothed envelopes thus computed from the reference and processed speech samples are subsequently compared in the cepstral domain. The envelopes are fitted with a set of five cepstral bases functions, and the degree of correlation for each fitted basis function between the reference and processed envelopes is calculated. The average of these correlations represents the quality of the processed signal - a cepstrum correlation (CC) value of 1 indicates a perceptually indistinguishable processed signal from the clean reference, while a value of 0 represents a severely distorted processed signal. In an attempt to further the accuracy of the HASQI metric, the final stage is the application of a second-order regression to fit the computed CC value to a database of subjective speech quality ratings. This was done separately for normal hearing (NH) subjects, where the following relationship was found:

$$Q_{nonlinear} = 0.618 - 2.184CC + 2.566CC^2 \tag{2.3}$$

and HI subjects, where the result was:

$$Q_{nonlinear} = 1.175 - 4.361CC + 4.186CC^2 \tag{2.4}$$

As previously mentioned, the standard HASQI computation includes a linear index, the intent of which is to account for effects on the long term spectrum caused by hearing aid DSP. Much like the nonlinear computational chain, the linear computation includes a cochlear model, which in this case produces the compression adjusted, average envelope magnitude for each of the filterbank channels for both the reference signal and the processed signal. These long term spectra form the inputs to the linear model computation which quantifies the differences between the long term magnitude spectra and spectral slopes. Let $|X(k)|$ be the input signal long term magnitude spectrum produced by a $K$channel gammatone filterbank with $|Y(k)|$ defined similarly for the output signal. The two signals are first converted to dB values with respect to threshold, with the new signals represented by $\left|\hat{X}(k)\right|$ and $\left|\hat{Y}(k)\right|$. The following equations then yield the differences in spectra:

$$d_{spectra}(k) = \left|\hat{Y}(k)\right| - \left|\hat{X}(k)\right| \qquad 0 < k < K-1 \tag{2.5}$$

$$d_{slope}(k) = \left[\left|\hat{Y}(k)\right| - \left|\hat{Y}(k-1)\right|\right] - \left[\left|\hat{X}(k)\right| - \left|\hat{X}(k-1)\right|\right] \qquad 0 < k < K-1 \tag{2.6}$$

The standard deviations, $\sigma_{spectra}$ and $\sigma_{slope}$ of $d_{spectra}(k)$ and $d_{slope}(k)$, are linearly combined to fit the subjective speech quality ratings according to the following relationship for NH subjects:

$$Q_{linear} = 1 - 0.400\sigma_{spectra} - 0.628\sigma_{slope} \tag{2.7}$$

and the following relationship for HI subjects:

$$Q_{linear} = 1 - 1.109\sigma_{spectra} - 0.000\sigma_{slope} \tag{2.8}$$

As can be seen, for HI subjects, it was found that the $\sigma_{slope}$ term was independent of the quality ratings indicating that HI listeners have difficulty in identifying spectral slope differences [45].

After computation of both the linear quality rating, $Q_{linear}$, and the noise and distortion quality rating, $Q_{nonlinear}$, the overall quality rating is computed as [45]:

$$Q_{overall} = Q_{linear}Q_{nonlinear}. \tag{2.9}$$

## 2.3 Method

### 2.3.1 Hearing Impaired Participants and Hearing Aids

For this study, a group of 20 HI subjects were recruited to provide speech quality ratings of a number of different speech-in-noise signals. The participant gender division consisted of 5 females and 15 males, with an age range between 65 and 87 years, with a mean of 76 years. The hearing loss profile of all the participants was similar between the left and right ears and the severity ranged from moderate to severe. The mean participant audiograms for the left and right ears is shown in Figure 2-2, which exemplify typical high frequency sensorineural hearing loss configurations.

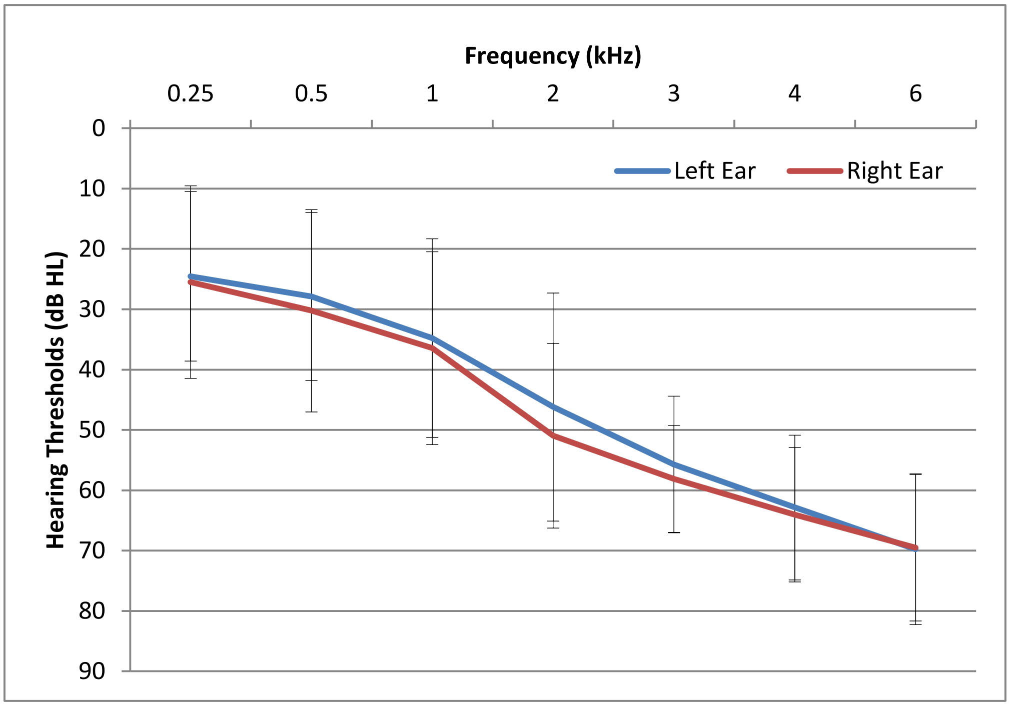



**Figure 2-2: Average left and right ear audiograms of the 20 HI participants. The error bars represent one standard deviation.**

Speech quality data were collected from two different bilateral wireless hearing aid models, *viz.* Oticon Agil and Siemens Motion. Key features of Agil include: a 10-channel wide dynamic range compression system with dual time constants that aims to preserve speech signal dynamics; and a spatial sound management that coordinates the bilateral compression and noise reduction systems such that naturally occurring spatial cues are preserved and speech perception in noise is optimized [62]. Salient features of Motion include: a 16-channel wide dynamic range compressor with syllabic time constants, and a wireless coordination strategy that synchronizes the directional and noise reduction features in the left and right hearing instruments. Both Agil and Motion incorporate multiband adaptive directionality to mitigate noise sources originating in the rear azimuths.

### 2.3.2 Hearing Aid Recordings

In order to collect the subjective speech quality ratings, a speech database was created which consisted of recordings from the experimental hearing aids under a variety of environmental conditions. The speech stimuli were recorded using a Head And Torso Simulator (HATS) wearing the hearing aids programmed to the specific hearing loss of each study participant. This allowed the recorded signals to be later presented to the subjects through a pair of insert earphones for speech quality ratings, without the explicit need for stopping the rating procedure to fit the second pair of hearing aids, changing the environment, or changing the hearing aid settings during the rating procedure. It must be noted here that ER-2 insert earphones were used for stimulus playback due to their flat frequency response and the ability to reproduce HATS recordings without any frequency shaping.

a) b)

**Figure 2-3: Hearing aid recording setup in the (a) low-reverberant and (b) high-reverberant environments.**

Hearing aid recordings were obtained in two different environments – in a hemi-anechoic chamber and in a reverberant chamber. The dimensions of the hemi-anechoic chamber were 12’ x 23’ x 18’ (L x W x H) and its broadband reverberation time ($RT_{60}$) was 40 ms. The reverberation chamber measured roughly 20’ x 13’ x 9’ with a broadband $RT_{60}$ of 890 ms. In both of these chambers, the HATS was placed at the centre of a circular array of speakers with a radius of 1.4 m. Figure 2-3 shows the recording setup for both the high reverberant and low reverberant environments.

In both of these environments, speech samples were presented from the speaker directly facing the HATS ($0^{o}$ azimuth), with uncorrelated background noise played out of speakers positioned at $90^{o}$, $180^{o}$, and $270^{o}$ degrees. Three specific Institute of Electrical and Electronic Engineers (IEEE) sentences spoken by each of a male and a female talker were played back consecutively as the speech material for all participants and conditions, at a level of 65 dB Sound Pressure Level (SPL). Two types of noise viz. multi-talker babble and traffic noise at overall Signal-to-Noise Ratios (SNRs) of 0 dB and 5 dB measured at the centre of the speaker array were used to create individual noise background settings. In addition to this symmetrical noise condition, an asymmetrical noise field was created. This experimental condition was included to probe the performance of wireless co-ordination between the hearing aids, as it was reported that bilateral wireless hearing aids preserve speech clarity and naturalness in asymmetric listening environments [13] , [14]. For this particular condition, only female speech samples were played from the front speaker, with speech-shaped stationary noise played back from a speaker positioned at $120^{o}$ azimuth. Thus a total of 16 symmetric (2 talkers x 2 noise types x 2 SNRs x 2 chambers) and 4 asymmetric (1 talker x 1 noise type x 2 SNRs x 2 chambers) speech-in-noise conditions were realized.

For each of these noise conditions, bilateral pairs of Agil and Motion were placed on the HATS in turn and stereo recordings were obtained for different hearing aid signal processing settings. First, the hearing aids were programmed to match the targets prescribed by the Desired Sensation Level (DSL) 5.0 algorithm [63] for each HI participant and verified in the Audioscan Verifit. Then, 4 different combinations of microphone/noise reduction and wireless modes were setup for each bilateral pair: omnidirectional microphone and wireless communication off, omnidirectional microphone and wireless communication on, adaptive directional and noise reduction with wireless communication off, and adaptive directional and noise reduction with wireless communication on. With this, a grand total of 3200 stimuli (20 noise conditions as described in the previous paragraph x 4 hearing aid settings x 2 hearing aids x 20 HI subjects) were recorded to constitute the database used for speech quality ratings. In addition, recordings of speech samples in quiet conditions (i.e. all noise sources turned off) were gathered in each room for each of the hearing aid signal processing settings. Furthermore, the sound pressure levels of the hearing aid recordings were noted, which were subsequently used in the speech quality ratings task, as described below.

### 2.3.3 Quality Ratings Data Collection

The subjective data collection was mediated by a custom software application, whose screenshot is shown in Figure 2-4.

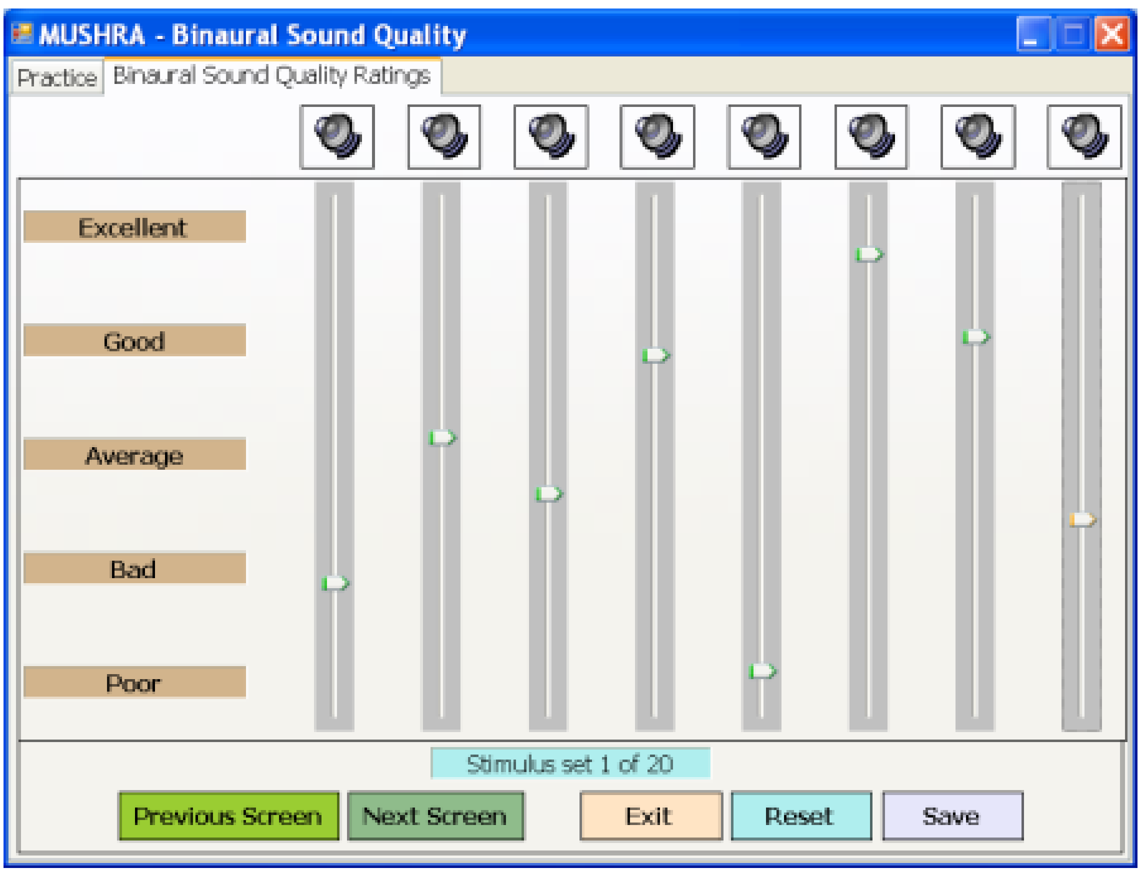


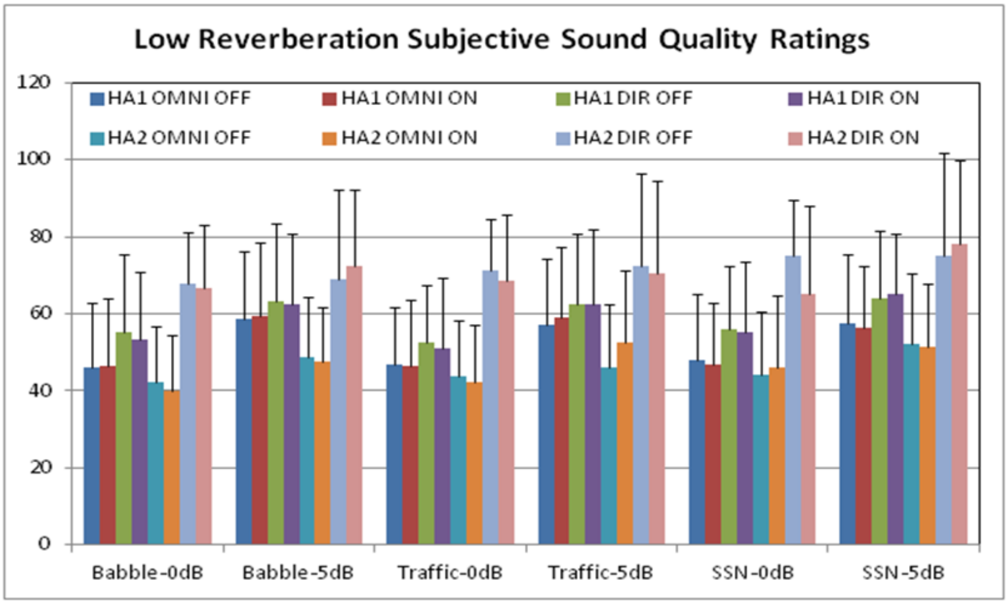

**Figure 2-4: The software user interface used to collect the subjective ratings.**

The quality ratings were obtained in a manner similar to the MUltiple Stimulus with Hidden Reference and Anchors (MUSHRA) paradigm [64] except no hidden reference or anchors were utilized. The experiment started with the HI participant seated in a sound-treated chamber in front of a computer monitor. The speech stimuli that were recorded for that particular participant were extracted from the database. The participant was asked to navigate through a set of 20 screens, each one representing a noise condition. Within each screen, there were eight speaker icons which were randomly associated with the eight hearing aid recordings for that particular condition. The listener was asked to click on each speaker icon, listen to the ensuing stimulus, and rate the speech quality on a sliding scale ranging from poor quality on the low end to excellent quality on the high end. The software that was used to collect the ratings produced a score between 0 and 100 based on the chosen position on a sliding scale. The listeners were encouraged to listen to these eight stimuli multiple times and readjust the slider positions if needed. They were asked to move on to the next screen once they were satisfied with the relative and absolute speech quality ratings of the eight stimuli. The speech quality ratings were stored in a text file which was later loaded into SPSS software version 16.0 for statistical analysis. It is pertinent to note here that 10 of the 20 participants came back at a later date to redo the rating task, providing data for test-retest reliability analyses.

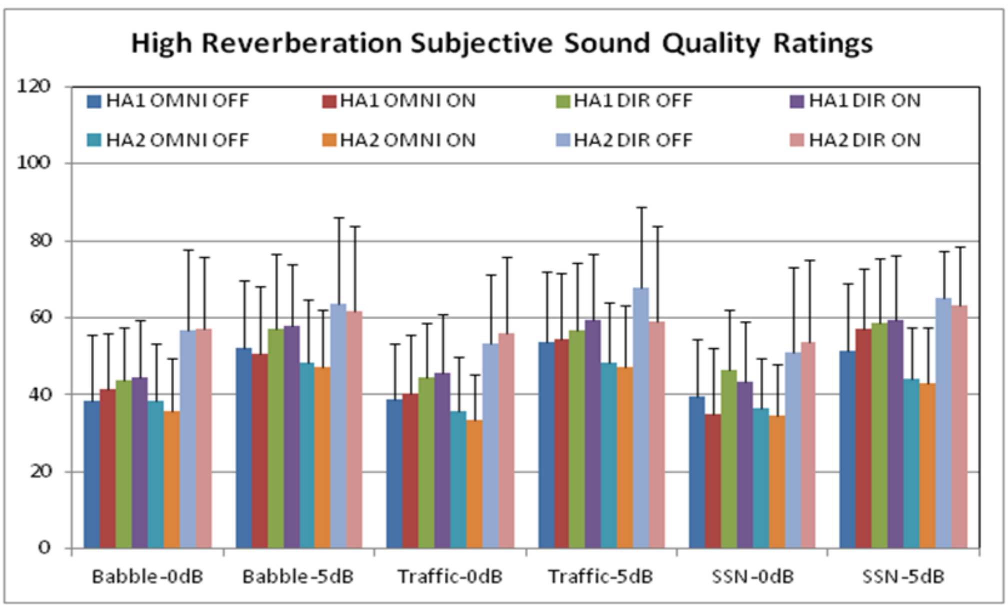
## 2.4 Results

### 2.4.1 Subjective Data

In order to measure the reliability of the subjective ratings, Cronbach's alpha was calculated. This is a measure of the inter-rating similarity between the ratings provided by each subject, where values of zero or less are indicative of random data and values approaching the maximum of one are indicative of highly reliable data. For the ratings provided in this study, the Cronbach's alpha was 0.887, which provides strong support to the notion that this is a reliable set of data. Similarly, correlation coefficients between the test – retest data ranged between 0.7 to 0.9, further attesting to the reliability of the quality ratings.




a)




b)

**Figure 2-5: Mean subjective sound quality ratings in a) the low reverberation environment and b) the high reverberation environment. HA1/HA2 refers to the hearing aid, OMNI/DIR indicates the directionality setting and ON/OFF refers to the state of the wireless link.**

Figure 2-5 shows the averaged subjective speech quality scores for stimuli recorded in the two environments.

The data in these graphs were grouped according to the noise condition, and the individual bars within each group represent one of the eight hearing aid conditions. The data in these graphs lend themselves to a few interesting observations. Beginning with the hearing aid model, it is clear that HA1 produced higher quality scores in the omnidirectional mode, while HA2 produced higher quality scores in the directional mode. In addition, the directional mode was preferred for both DHA models in both environments which indicates that the directionality algorithms were successful in improving the sound quality under the studied conditions. With respect to the wireless link, in some cases a slight improvement is observed while in other cases a slight degradation is observed. Based on this data, it appears that the wireless link does not offer any improvement in regards to sound quality.

Repeated measures ANOVA was conducted using SPSS 16.0 software to measure statistical significance of the differences among the speech quality ratings. Table 2-1 reports the significant main, two-way, and three-way interactions among the different variables. The main effects of chamber (low vs. high reverberation), SNR (0 dB vs. 5 dB), and microphone mode (omnidirectional vs. adaptive directional) were not surprising. It is interesting that noise type was not a main factor and the wireless variable is

**Table 2-1: Results of the repeated measures ANOVA.**

| | Variable(s) | F | Hypothesis dF | Error dF | *p* |
|---|---|---|---|---|---|
| Main effects | Chamber | 43.135 | 1 | 19 | 0.000 |
| | SNR | 44.851 | 1 | 19 | 0.000 |
| | DHA | 4.481 | 1 | 19 | 0.048 |
| | Microphone mode | 88.101 | 1 | 19 | 0.000 |
| Two-way | Chamber x Microphone mode | 6.957 | 1 | 19 | 0.016 |
| | Chamber x SNR | 10.445 | 1 | 19 | 0.004 |
| | SNR x DHA | 6.689 | 1 | 19 | 0.018 |
| | DHA x Microphone mode | 79.749 | 1 | 19 | 0.000 |
| Three-way | Chamber x Noise x DHA | 6.590 | 2 | 18 | 0.007 |
| | Chamber x DHA x Microphone mode | 9.897 | 1 | 19 | 0.005 |

conspicuous in its absence among main effects. There was a significant main effect of DHA, indicating the performance differences between the two brands. Furthermore, there was a significant interaction between the DHA model and microphone mode, due to the aforementioned pattern of HA1 and HA2 scores when in omnidirectional and adaptive directional processing modes. The magnitude of difference between HA1 and HA2 scores in omnidirectional and directional modes depended on the environment and hence the three-way interaction between chamber, DHA, and microphone mode variables. The SNR x DHA interaction was significant as the scores between the DHAs, when collapsed across the microphone modes, were similar at 5 dB SNR and different at 0 dB SNR. In addition, while the speech quality scores were lower in the high reverberant environment for both SNRs, the drop relative to the ratings in the low reverberant environment was steeper for the 0 dB SNR (Chamber x SNR interaction). This result is not surprising, as there is evidence that noise and reverberation synergistically degrade speech perception [65], which explains the steeper drop in speech quality in the presence of both higher reverberation and background noise. The final three way interaction between chamber, noise, and DHA stemmed from the substantial drop in HA2 speech quality scores for the asymmetric noise condition between low and high reverberant environments.

### 2.4.2 Objective Data

Spectrographic analyses were conducted on the DHA recordings to gain further insight on DHA processing. Figure 2-6 depicts a comparison of sample spectrograms computed from a set of stimuli recorded in the low reverberant chamber in the presence of asymmetric noise at 0 dB SNR. The top panel shows the spectrogram of the clean speech stimulus at the input of the DHA. The bottom three spectrograms display the time-frequency content of the corresponding outputs from HA1 in omnidirectional mode, HA1 in adaptive directional mode, and HA2 in adaptive directional mode, respectively. The increased clarity of the speech features (harmonic structure, formant tracks and transitions) in the HA2 directional output is evident in Figure 2-6, which reflects the higher subjective speech quality ratings for this condition.

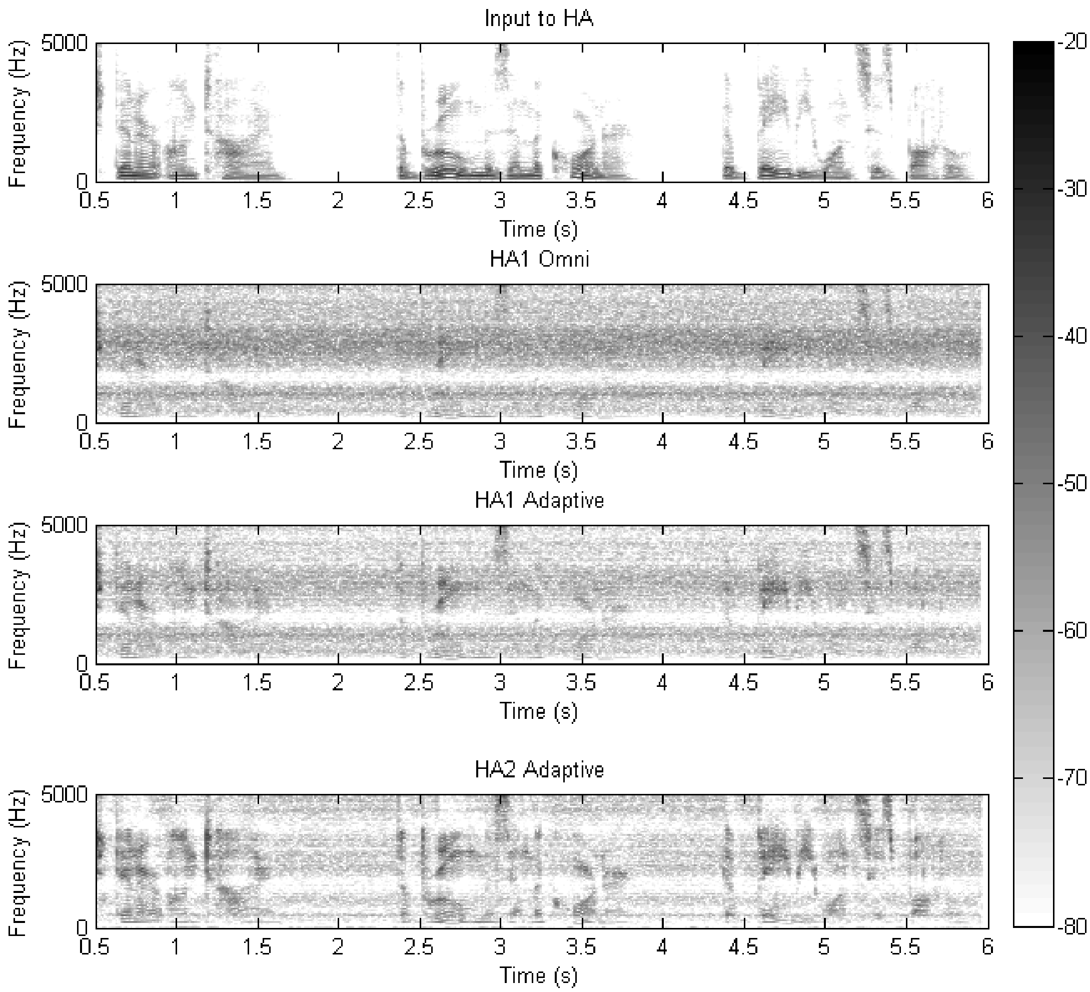



**Figure 2-6: Spectrograms showing the clean speech, an omnidirectional recording and an adaptive recording from each DHA.**

Both LLR and HASQI values were computed for the 3200 stimuli in the database. As discussed earlier, both these metrics require a clean reference speech sample for comparative purposes. This clean reference was generated in two different ways:

- by a separate recording through the DHA with all the noise sources turned off and every other variable (environment, DHA microphone mode, and talker) remaining the same. This quiet recording served as the reference for that particular DHA condition; and

- by applying a static Finite Impulse Response (FIR) filter to the clean speech sample. This FIR filter was designed separately for each HI participant to match the targets specified by the DSL 5.0 [63] algorithm for a 65 dB SPL input. This approach follows a similar procedure undertaken by Arehart and colleagues [57], [58] in their studies investigating the behaviour of HASQI.

The computation of LLR or HASQI metrics started with temporal alignment of the reference and test signals using the cross-correlation procedure. For the LLR metric, the reference and test signals were divided into 30 ms frames with 25% overlap between successive frames. An $18^{th}$ order LPC filter was utilized, and a frame-wise LLR metric was calculated using Equation 2.1. The final LLR metric was the average of these frame-wise LLR values. Within the HASQI implementation, a 32-channel gammatone filterbank was used, with the centre frequencies spanning between 150 Hz and 8000 Hz. Given that both left and right ear recordings were captured, the average of the individual left and right ear ratings generated by both the HASQI and LLR metrics were taken as the overall quality estimates. For each of the 3200 stimuli, the absolute SPL of the bilateral DHA outputs was noted during the recording stage and passed on to the HASQI computational algorithm along with the appropriate audiogram. In addition to the overall HASQI value, the linear, nonlinear, and CC values were also retrieved and investigated through correlational analysis.

Table 2-2 displays the result from this analysis. The first two rows show the correlation coefficients for the LLR metric, and the last six rows for different HASQI versions. It can be noted that the LLR correlation is poor when a static FIR filter is used for frequency-shaping the clean reference. Due to the WDRC operation, the frame-to-frame DHA output spectra are different from the average DSL 5.0 targets which the static filter emulates. This issue is mitigated by utilizing the appropriate quiet recording as a reference, so that the frame-to-frame dynamics are taken into account. With the quiet recording as the reference, an increase in the correlation coefficients can be seen in Table 2-2.

**Table 2-2: Correlation coefficients of different DHA speech quality metrics with subjective ratings. All reference signals were generated following the FIR filter approach unless otherwise specified.**

| Electroacoustic Measure | Low Reverberation Correlation | High Reverberation Correlation |
|---|---|---|
| LLR | -0.243* | -0.277* |
| LLR Quiet Ref | -0.729* | -0.606* |
| HASQI | 0.847 | 0.887 |
| HASQI Linear | 0.330 | 0.074 |
| HASQI Non Linear | 0.818 | 0.905 |
| HASQI CC | 0.877 | 0.898 |
| HASQI Quiet Ref | 0.873 | 0.870 |
| HASQI No HL | 0.781 | 0.762 |

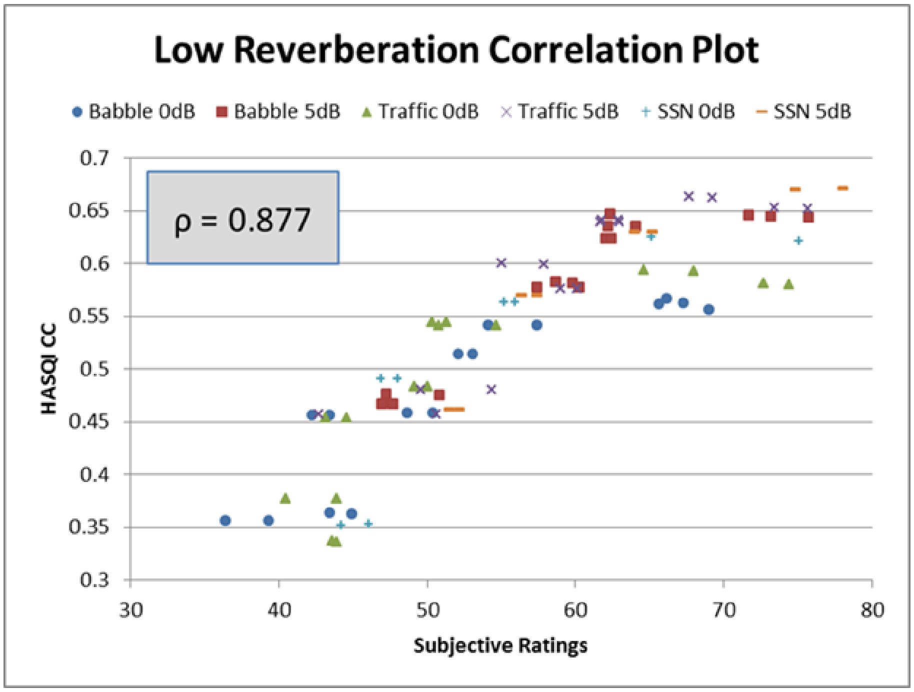
*For LLR, a more negative score is indicative of better performance, with the best possible performance indicated by a score of -1.

With the exception of the linear term, it can also be seen that the HASQI correlation values are greater than those resulting from the LLR. Two other salient points are of interest from Table 2-2: (a) the HASQI CC, which is the average of the cross-correlation of the processed and clean cepstral bases functions, performed just as well as the overall HASQI, and (b) there was a significant reduction in the correlations when the HASQI computational scheme simulating normal audition, i.e., no cochlear hearing loss (termed HASQI No HL in the table) was used.

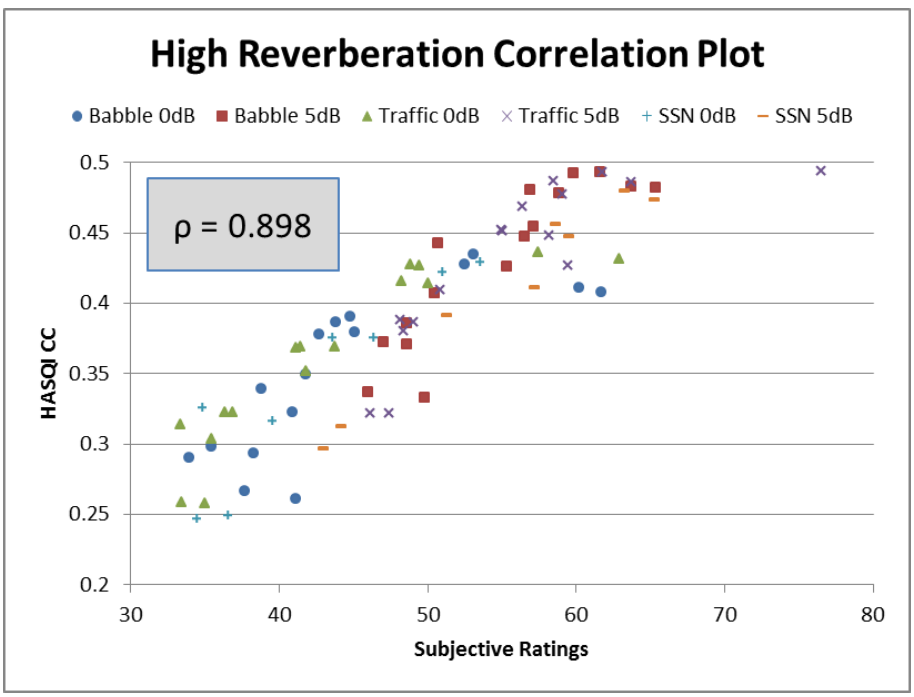
Finally, Figure 2-7 displays the scatter plots between the subjective ratings and the HASQI CC ratings across different noisy and reverberant conditions. These are included since HASQI CC exhibited the best overall performance and the high correlation coefficients in both the high and low reverberant environments are evident in this figure.

## 2.5 Discussion

This study evaluated the speech quality performance of two modern DHAs in a variety of environmental conditions, both objectively and subjectively. Several interesting results were observed and the more salient ones are discussed below.




a)




b)

**Figure 2-7: Correlation plots for a) the low reverberation and b) the high reverberation environments.**

The procedure for collecting subjective speech quality data was more rigorous in this study, than those found in the literature. A custom database of DHA-processed speech stimuli, individualized for each of the HI participants, was created for this study. Furthermore, the speech quality ratings themselves were obtained using the MUSHRA technique, which - although popular in telecommunications and audio engineering fields- is rarely used in DHA speech quality evaluation. The MUSHRA approach allows for multiple DHA stimuli to be heard and compared – it not only allows for rank ordering different DHA settings, but also allows for quantifying the relative differences between them. The inter- and intra-subject reliability with MUSHRA data is high, as evidenced by the Cronbach's $\alpha$ of 0.887. Nunnally [66] states that an instrument or measure with a Cronbach's $\alpha$ of 0.7 or above can be considered acceptable as a rule of thumb.

The speech quality ratings were obtained from two different models of bilateral DHAs – Oticon Agil and Siemens Motion. Each bilateral pair was further programmed to operate with four different combinations of the microphone mode (omnidirectional and adaptive directional) and wireless communication (activated or deactivated). Analysis of the subjective data revealed an interesting pattern – listeners preferred the quality of Agil in omnidirectional mode, while Motion was preferred in the adaptive directional mode. Spectrographic analyses revealed that the adaptive directional system in Motion reduced background noise more and preserved speech components better. The reason for better performance with Agil in omnidirectional mode is less clear. A probable cause is the difference in WDRC strategies – while Agil uses the "Speech Guard" system which strives to preserve speech dynamics as much as possible, Motion employs multichannel compression with syllabic time constants.

Currently there is a paucity of studies investigating the impact of bilateral wireless communication. In contrast to the results presented in [13] and [14], where sound quality ratings were found to be improved with wireless synchronization of bilateral DHAs, this study was not able to demonstrate an improvement for the conditions studied. This is not entirely surprising, as another study [67] showed that there was no significant improvement in speech intelligibility, and there was a significant improvement in sound localization for only one of the conditions tested. Taken together, these results support the notion that the effect of wireless coordination in current bilateral hearing aids on sound quality is constrained to providing improvements under a limited number of specific conditions and it remains unclear if this would be noticeable to users in a real-world environment.

This study applied HASQI in predicting the DHA speech quality ratings when operating in real-world environments. Arehart et al. [57] reported correlations between HASQI ratings and subjective ratings for both NH and HI listeners. For the HI group, a simulated hearing aid was used, which differs from the real hearing aids used in this study. Arehart reported correlations of 0.957 for conditions that included noise and nonlinear hearing aid processing, 0.938 for conditions that included linear filtering and 0.963 for a set of signals that combined noise, nonlinear processing and linear filtering. For the normal hearing group, the correlations were 0.895, 0.785 and 0.877 respectively. Recently, Kressner et al. [59] conducted a robustness study of HASQI by computing predicted sound quality scores for a large set of speech signals processed by noise suppression algorithms. The predicted scores were compared to subjective ratings provided by normal hearing (NH) listeners and the reported correlation was 0.85. Based on this study, Kressner [59] concluded that HASQI “generalizes very well for NH listeners and achieves performance comparable to other commonly used metrics”.

This study further validated the robustness of HASQI though the application to a novel set of HI ratings, through the utilization of commercially available hearing aids rather than simulated hearing aids and by considering a high-reverberation environment. As shown in the results section, the correlation results of 0.877 for the low-reverberation environment and 0.898 for the high-reverberation environment indicate that HASQI maintains a high level of performance under these new conditions. It was interesting to note that by reducing the complexity of the HASQI measure to only include the previously described HASQI CC, the greatest overall performance was achieved. As can be seen in Table 2-2, the HASQI linear model did not generalize well to the signals used for this study. In addition, the fitting of features developed in the original HASQI model did not generalize to this study. Nevertheless, the HASQI CC did generalize well for this study which differed from previous studies in the use of real hearing aids rather than simulated hearing aids. HASQI CC significantly outperformed the traditional LLR measure.

## 2.6 Conclusions

In closing, this study described a procedure for collecting reliable speech quality data from HI listeners. This data was used to differentiate the performance of two different bilateral DHA models and their varied features. The study also served to further validate the robustness of HASQI for predicting DHA speech quality ratings collected from HI listeners. It must be noted here that for predicting the quality of a particular DHA-processed signal, HASQI requires a second signal, which is the cleaner (no-noise, no-distortion) version of the test signal. A better alternative is a metric that estimates speech quality based on the DHA output alone, and this class of “Reference-Free” metrics forms the focus of the next chapter.

# Chapter 3

# 3 Reference-Free Speech Quality Measure

This chapter introduces the use of a reference free speech quality measure developed for hearing aid signals. As will be discussed, this approach has advantages over previously developed hearing aid speech quality measures, while sacrificing a small amount of accuracy in the prediction of subjective ratings.

## 3.1 Motivation

The previously described HASQI model is an example of so-called "intrusive" speech quality estimation procedures, where the features are derived from two separate signals - the DHA output and the corresponding clean reference input. This procedure necessitates additional considerations prior to the computation of the quality metric, which include proper time alignment between the reference and processed signals and appropriate frequency shaping of the reference signal based on the hearing loss profile that was used to fit the DHA under test. In contrast, a reference-free speech quality measure will obviate the need for a proper comparative reference signal as the computation is based solely on the DHA recording. Furthermore, such a "non-intrusive" index has the potential for 'on-the-fly' adjustments to the DHA signal processing parameters such that the estimated quality of the processed signal is maximized[1]. A similar need for non-intrusive speech quality estimation techniques exists in the telecommunication industry. Without a non-intrusive method, it is necessary to inject a known signal into the portion of the network under test which can be quite costly and time consuming. A non-intrusive approach allows for the speech quality estimation to occur by simply capturing the transmitted signal at the points of interest within the network. Based on this advantage, a few reference-free speech quality metrics have been proposed [69]–[71] and standardized

[1] A similar strategy is used in premium digital hearing aids from Widex, where the hearing aid DSP parameters are fine-tuned to maximize the Speech Intelligibility Index (SII) [68]

by the ITU and American National Standards Institute (ANSI) for telecommunication applications [72], [73].

The ITU has adopted P.563 as the recommended method for non-intrusive speech quality estimation [72]. A more recent approach, the Auditory Non-Intrusive Quality Estimation Plus (ANIQUE+) metric has demonstrated improved performance in comparison to P.563.

The ANIQUE+ metric proposed in [71] for telecommunication applications is outlined in Figure 3-1. After normalizing the level of the input signal and filtering to account for the effect of the particular handset under study, the signal is applied to three separate distortion models. The outputs of these three distortion models is assimilated in the feature mapping block and a final estimated speech quality score is generated. The non-speech detection block and mute detection blocks seek to account for the effects of packet loss and bit errors that can occur in telecommunication networks and are not directly applicable to the current study. Conversely, the cochlear and modulation filterbank modelling and analysis are based on properties of the human auditory and speech production systems and therefore do have relevance to hearing aid applications.

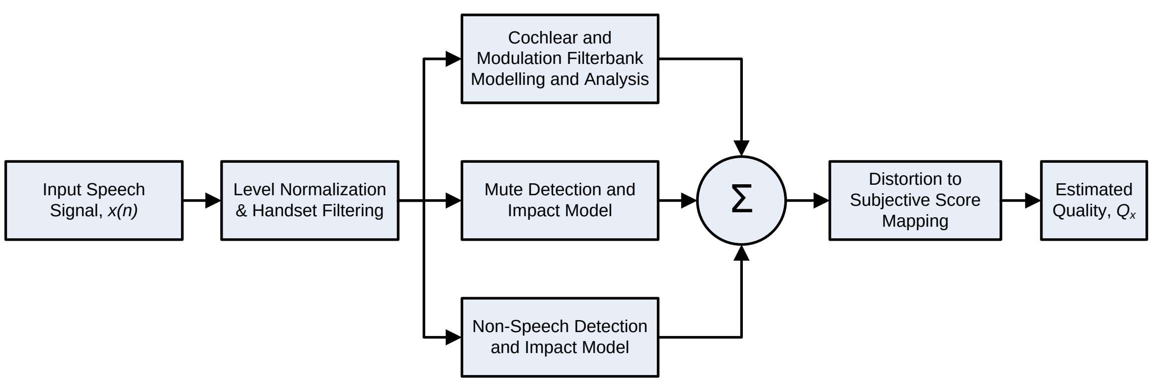



**Figure 3-1: Block diagram of the ANIQUE+ speech quality estimation model for telecommunication applications.**

Though recent speech quality estimation techniques for telecommunications such as ANIQUE+ have demonstrated impressive performance, studies applying reference-free speech quality indices to DHA applications are currently lacking. To this end, the study presented in this chapter proposes and investigates a novel reference- free speech quality metric, termed the Speech-to-Reverberation Modulation Ratio - Hearing Aid (SRMR-HA). The performance of SRMR-HA in predicting the speech quality ratings of DHA output signals obtained in a variety of noisy and reverberant environments is evaluated and compared with the performance of HASQI.

## 3.2 Development

As introduced above, a speech quality estimator that does not require a proper reference signal is attractive. Figure 3-2 shows the block diagram of one such estimator developed for DHA applications. The SRMR-HA is a modified and extended version of the Speech to Reverberation Modulation energy Ratio (SRMR) [70], which was originally developed for assessing the performance of dereverberation algorithms and validated with subjective data collected from NH listeners.

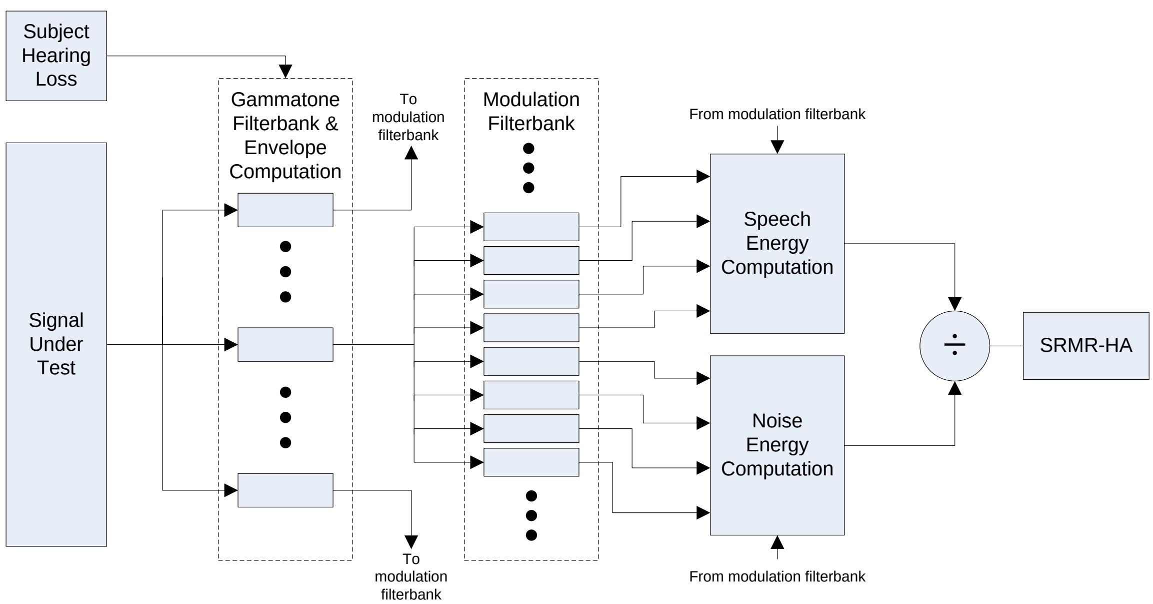



**Figure 3-2: A reference free speech quality estimator for hearing aid applications.**

Being a reference-free technique, the SRMR-HA method does not require any prior temporal alignment. Similar to the HASQI computational procedure, the processed signal is first passed through a gammatone filterbank which is implemented based on the work of Cooke described in [61]. The gammatone function is derived based on experimental studies of frequency selectivity in the human auditory system and is given by:

$$g(t) = t^{n-1}e^{-bt}\cos(\omega t)\,u(t) \tag{3.1}$$

where $g(t)$ is the gammatone impulse response, $n$ is the filter order, $b$ is related to the filter bandwidth, $\omega$ is the radian frequency and $u(t)$ is the unit-step function. For analysis and evaluation purposes, it was necessary to develop a digital domain filter approximation that fits this model as closely as possible. Cooke investigated various methods to achieve this and found that the application of an Impulse Invariant Transform (IIT) yielded the most accurate results. The impulse invariant transform approximates a continuous time filter by finding a digital domain transfer function that results from a sampled version of the continuous time impulse response. This can be expressed as follows:

$$H(z) = Z[h(m)] = Z[h_a(t)|_{t=mT}] \tag{3.2}$$

where $h_a(t)$ is the continuous time impulse response of the filter to be approximated, $H(z)$ is the transfer function of the discreet-time filter and $T$ is the sampling period. The gammatone filter of order $n$ can then be defined as follows:

$$IIT_n(z) = Z[(mT)^{n-1}e^{-bmT}u(mT)] \tag{3.3}$$

Based on the well-known properties of the Z transform, transfer function representations of the digital approximation to the gammatone filter for orders 1 through 4 were found to be:

$$IIT_1(z) = \frac{1}{1 - az^{-1}} \tag{3.4}$$

$$IIT_2(z) = \frac{Taz^{-1}}{1 - 2az^{-1} + a^2z^{-2}} \tag{3.5}$$

$$IIT_3(z) = \frac{T^2(az^{-1})(a + z^{-1})}{1 - 3az^{-1} + 3a^2z^{-2} - a^3z^{-3}} \tag{3.6}$$

$$IIT_4(z) = \frac{T^3(az^{-1})(a^2z^{-2} + 4az^{-1} + 1)}{1 - 4az^{-1} + 6a^2z^{-2} - 4a^3z^{-3} + a^4z^{-4}} \tag{3.7}$$

where $a = e^{-bt}$. For this study, $IIT_4(z)$ was used to implement the gammatone filter bank. In order to account for the effects of SNHL, the Q factor of each filter is adjusted based on the $OHC_{Loss}$ parameter derived from the HL data in line with the description provided in section 2.2.2 and equation (2.2).

After the gammatone filterbank portion of the model is complete, the next step is to apply the extracted envelope in each channel to an 8-channel modulation filterbank, which has centre frequencies of 4.00 Hz, 6.60 Hz, 10.8 Hz, 17.7 Hz, 29.0 Hz, 47.6 Hz, 78.0 Hz and 128 Hz. Each filter within the filterbank was implemented as a second order bandpass filter with a Q value of 2. The lower four channels of the modulation filterbank are assumed to contain mostly speech-related components, while the upper four channels are occupied by predominantly noise- or distortion-related components [70], [74]. As such, the SRMR-HA is calculated as the ratio of modulation energies in the lower and upper four channels. The rationale for quantifying the modulation energies in the above-described fashion can be explained from the modulation-domain spectrograms.

Figure 3-3 displays modulation spectrograms computed from a set of speech stimuli from the bilateral DHA database described in Chapter 2. In these plots, the abscissa represents the centre frequency of the modulation filterbank, the ordinate represents the centre frequency of the gammatone filterbank, and the colors represent the relative modulation energy. The top-left panel displays the modulation spectrogram of a clean speech sample. It is important to point out that much of the modulation energy in this figure occupies the 4 Hz – 10.8 Hz range.

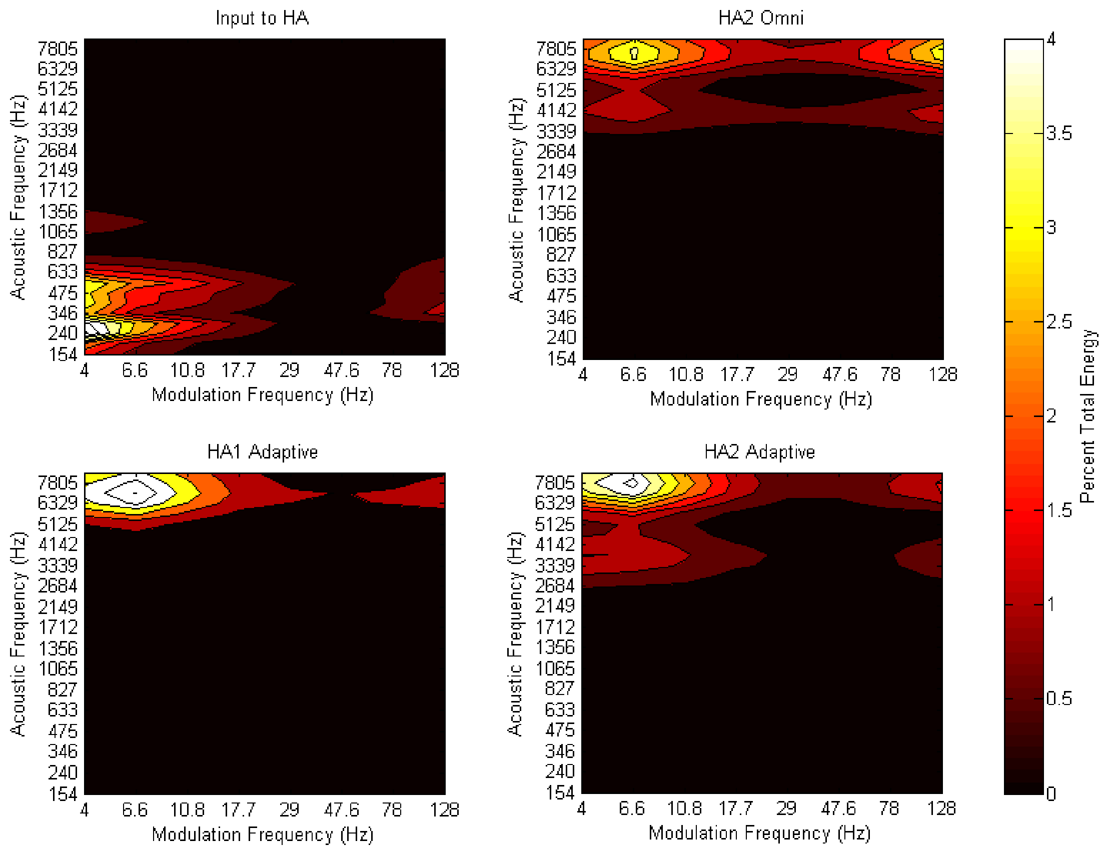



**Figure 3-3: Modulation spectrograms derived from a set of speech stimuli from the bilateral DHA database created in Chapter 2.**

The top-right panel shows the modulation spectrogram of the HA2 output, when it is programmed to be in omnidirectional mode and when the clean input speech sample was played back along with speech-shaped noise at 0 dB SNR in the low reverberant environment (the asymmetric noise condition described in Chapter 2). Two phenomena can be noticed in this plot: (a) there is a shift in modulation energy towards high frequencies along the y-axis. This is due to the high frequency gain imparted by the DHA to compensate for the high frequency hearing loss; and (b) the modulation energy is no longer concentrated in the lower frequencies, as presence of background noise led to the spread of modulation energy across the 4 – 128 Hz region. Activation of adaptive directionality counteracts against this, by reducing the background noise. The two modulation spectrograms in the bottom row of Figure 3-3 attest to this fact, where the spread of energy towards higher modulation frequencies is mitigated. It is also useful to highlight the differences between “HA2 adaptive” and “HA1 adaptive”. A greater proportion of the lower frequency modulation energy is preserved by HA2 adaptive. As such, it will have a greater SRMR-HA value. This relates to the subjective data, as results from the previous chapter showed that HI listeners preferred the quality of HA2 in directional mode and in the presence of background noise.

## 3.3 Performance Evaluation

The SRMR-HA was computed for all 3200 stimuli in the bilateral DHA database described in the previous chapter. Similar to HASQI, a 32-channel gammatone filter with centre frequencies ranging from 250 Hz to 8000 Hz was used in SRMR-HA computation. Figure 3-4 displays the scatter plot between the SRMR-HA measure and the subjective speech quality scores for the low- and high-reverberant environments respectively. Although the correlation coefficients are statistically significant, their absolute values (0.631 and 0.588) are low, especially when compared to high correlations reported by the HASQI CC in Chapter 2 for the same database.

Further investigations into these relatively poor correlations were undertaken by breaking the correlations down according to the background noise condition. Table 3-1 displays the correlation coefficients calculated from sub-classes of stimuli belonging to a particular noise and reverberation group.

**Table 3-1: Correlation coefficients between SRMR-HA and subjective speech quality scores for each noise and reverberation condition.**

| | Low-reverberation | High-reverberation |
|---|---|---|
| Multi-talker Babble | 0.610 | 0.511 |
| Traffic | 0.648 | 0.623 |
| Speech-shaped | 0.753 | 0.676 |
| Overall | 0.631 | 0.588 |

The highest correlations were noted for speech-shaped noise in the low reverberation environment, while the poorest correlations were noted with multi-talker babble in the high reverberation environment. Since the SRMR-HA is solely dependent on the relative distribution of modulation energy, its performance is affected in conditions where background noise has a “speech-like” modulation pattern or vice versa. Since multi-talker babble has modulation characteristics approaching that of speech and reverberation corrupts the speech modulation patterns, SRMR-HA performs poorly in these conditions. Conversely, speech-shaped noise has a modulation pattern unlike speech, and therefore SRMR-HA performs better, especially in low-reverberation environment.

To find a remedy to the poor correlations by SRMR-HA alone, feature augmentation was considered. It is very rare that non-intrusive or reference-free speech quality metrics are derived from a single feature alone. For example, the ITU standard P.563 [72] utilizes eight different features in deriving its speech quality estimate. The aforementioned ANIQUE+ method [71] uses three different feature sets in its speech quality model. As such, a modified SRMR-HA was derived as a linear combination of a set of features. Following the work of Petkov et al. [75], the chosen feature set included the mean and variance of the modulation filterbank output energies. The feature set was calculated for all the stimuli in the database, and the optimal combination of these features was decided through multiple linear regression analysis to match the subjective speech quality ratings, which was done separately for the low-reverberation and high-reverberation environmental data. The regression weight set (in the order of constant, speech portion mean, noise portion mean, speech portion variance, noise portion variance) was [254.60, 28.86, -43.09, -9.24, 9.81] and [373.18, -0.83, -43.65, -10.92, 24. 96] for the low and high reverberation data set respectively.

Figure 3-5 depicts the scatter plots generated after the multiple regression analysis, where the predicted speech quality scores using the linear combination of the features are plotted against the actual speech quality. It is evident that the correlation coefficients improved significantly in comparison to those shown in Figure 3-4 with the assimilation of additional features, with values of 0.857 and 0.792 for the low- and high-reverberation environments respectively. This is due to the fact that additional relevant features have been included and fit to the subjective ratings.

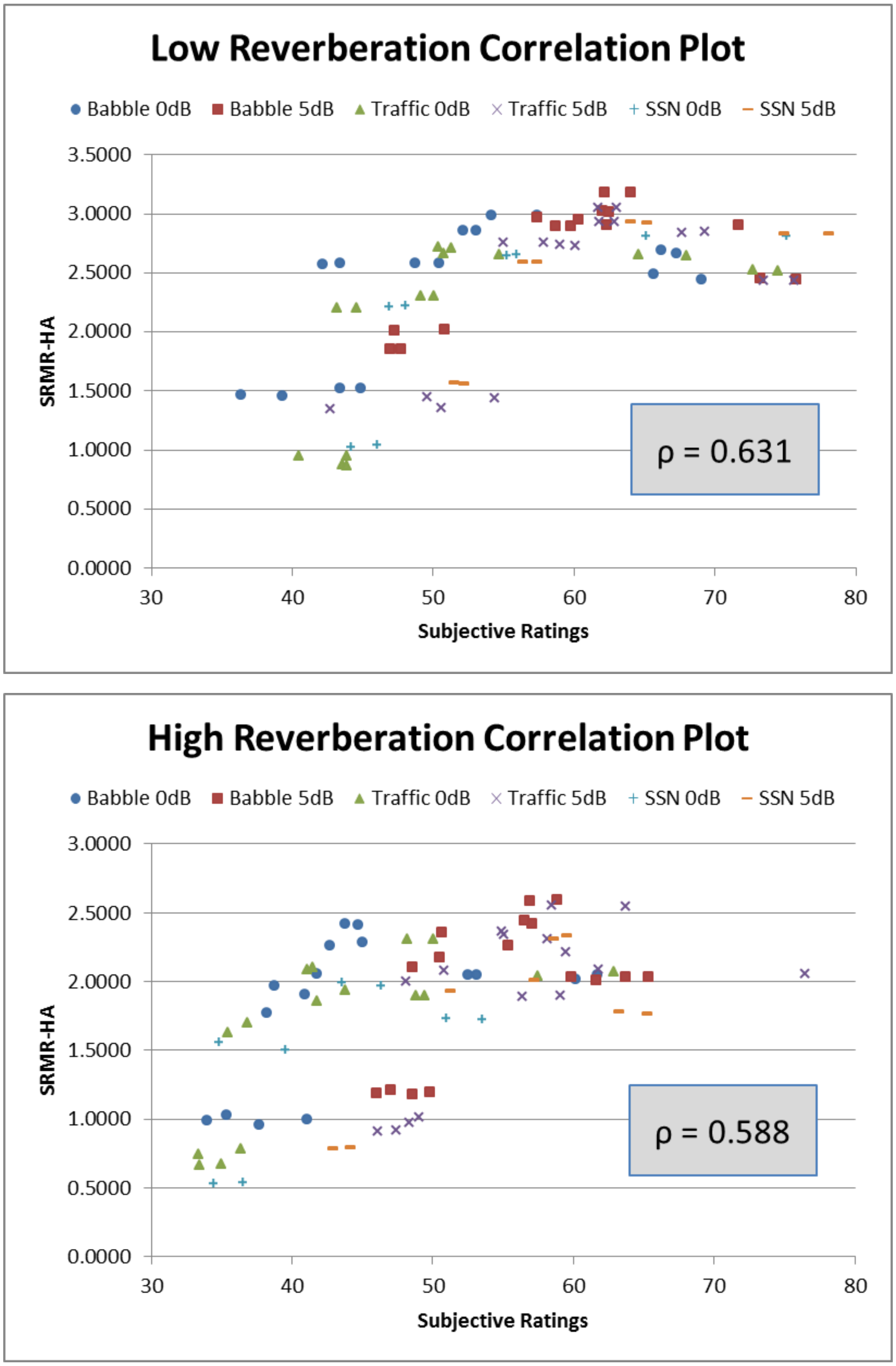
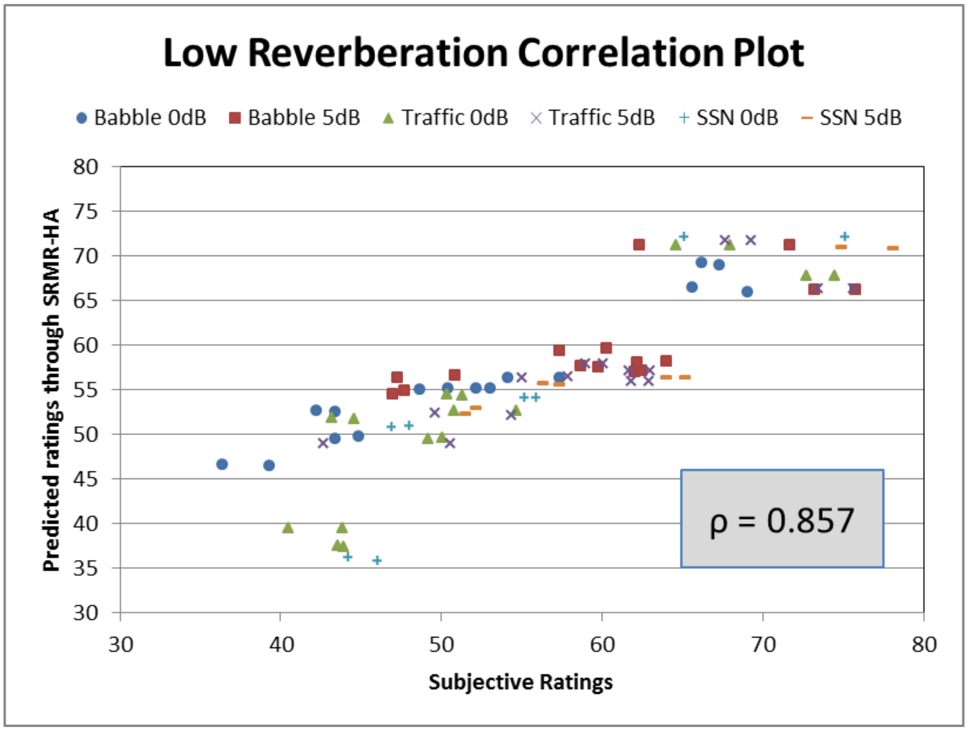



**Figure 3-4: Scatter plots for the SRMR-HA metric computed from the speech stimuli in bilateral DHA database.**



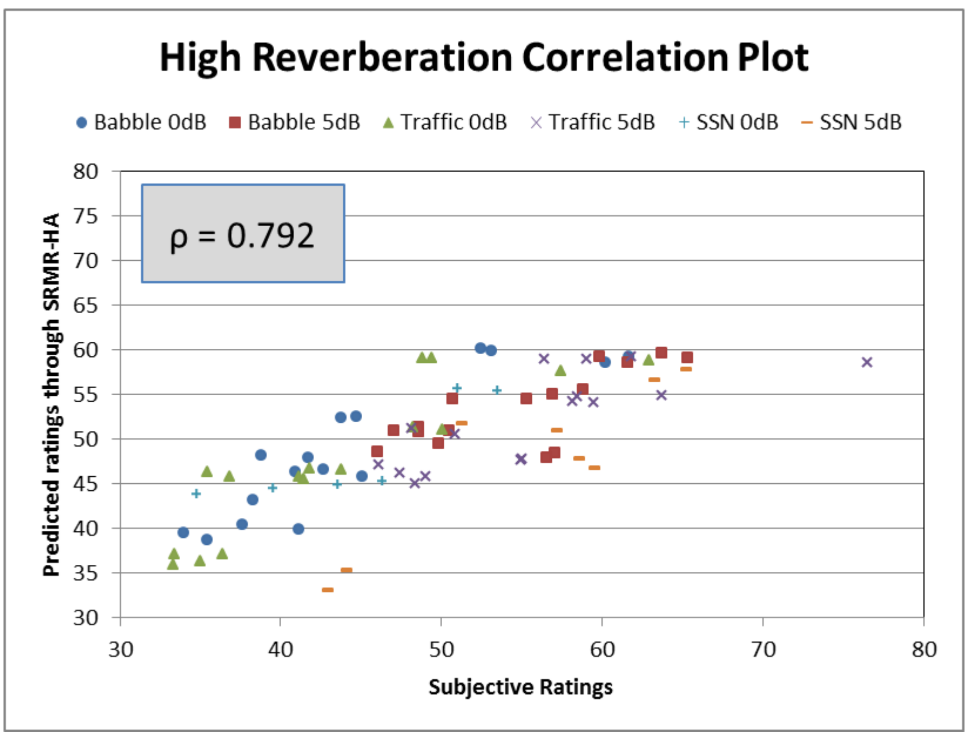




**Figure 3-5: Scatter plots between actual and predicted speech quality ratings for the bilateral DHA database. Predicted ratings were computed from multiple linear regression between SRMR-HA feature set and subjective ratings.**

## 3.4 Further Validation of SRMR-HA

In order to further evaluate the performance of the proposed reference-free SRMR-HA metric, a second speech quality rating database obtained with a different set of DHAs and recording equipment was utilized. This database was collected at the National Centre for Audiology as part of a separate research project [76], and a brief description of it is given below.

For this database, speech quality data was collected from 22 HI listeners, whose mean audiometric data are shown in Figure 3-6.

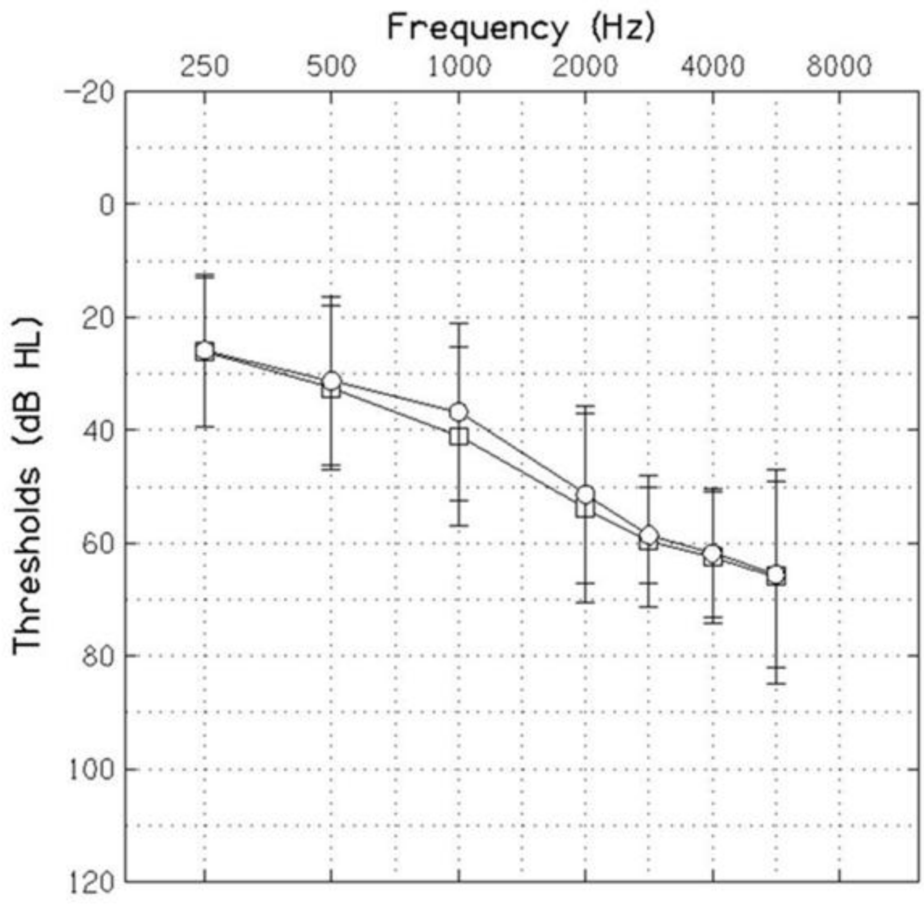



**Figure 3-6: Average pure-tone thresholds (with one standard deviation bars) for the right and left ears for the HI participants in the second database.**

Each of the HI listeners were fitted bilaterally with the Unitron experimental (modified Passport) behind-the-ear (BTE) DHAs using the DSL 5 adult prescriptive algorithm [63]. The subjects were then seated in the middle of a speaker array, either in a low reverberant (sound booth, $RT_{60}$ = 0.1s) or a highly reverberant environment (reverberation chamber, $RT_{60}$ = 0.9s). In both of these environments, three consecutive IEEE Harvard speech sentences [19] were played from the speaker at $0^o$ azimuth, while speech-shaped stationary noise or multi-talker babble was played from speakers positioned at $0^o$, $90^o$, $180^o$, and $270^o$ azimuths at 0 dB, or 5 dB Signal-to-Noise Ratios (SNRs). For each of these environmental conditions, HI subjects were asked to switch between four different DHA settings: omnidirectional, adaptive directional, partial strength DSP (where the directionality, digital noise reduction, and speech enhancement algorithms are operating at less than their maximum strengths), and Full Strength DSP (where all the DSP features were set to operate at their maximum strength). The subjects were then asked to rate the perceived quality of the speech stimulus for each of the DHA settings in each of the environmental conditions using a MUSHRA-like rating interface similar to Figure 2-4. The average subjective speech quality scores, shown in Figure 3-7, were later used to benchmark the performance of the quality metrics in each environmental condition as described below. The experimental DHAs were placed on a Knowles Electronic Manikin for Acoustic Research (KEMAR), which in turn was placed in the middle of a speaker array. Figure 3-8 displays the experimental setup for DHA recordings in the reverberation chamber.

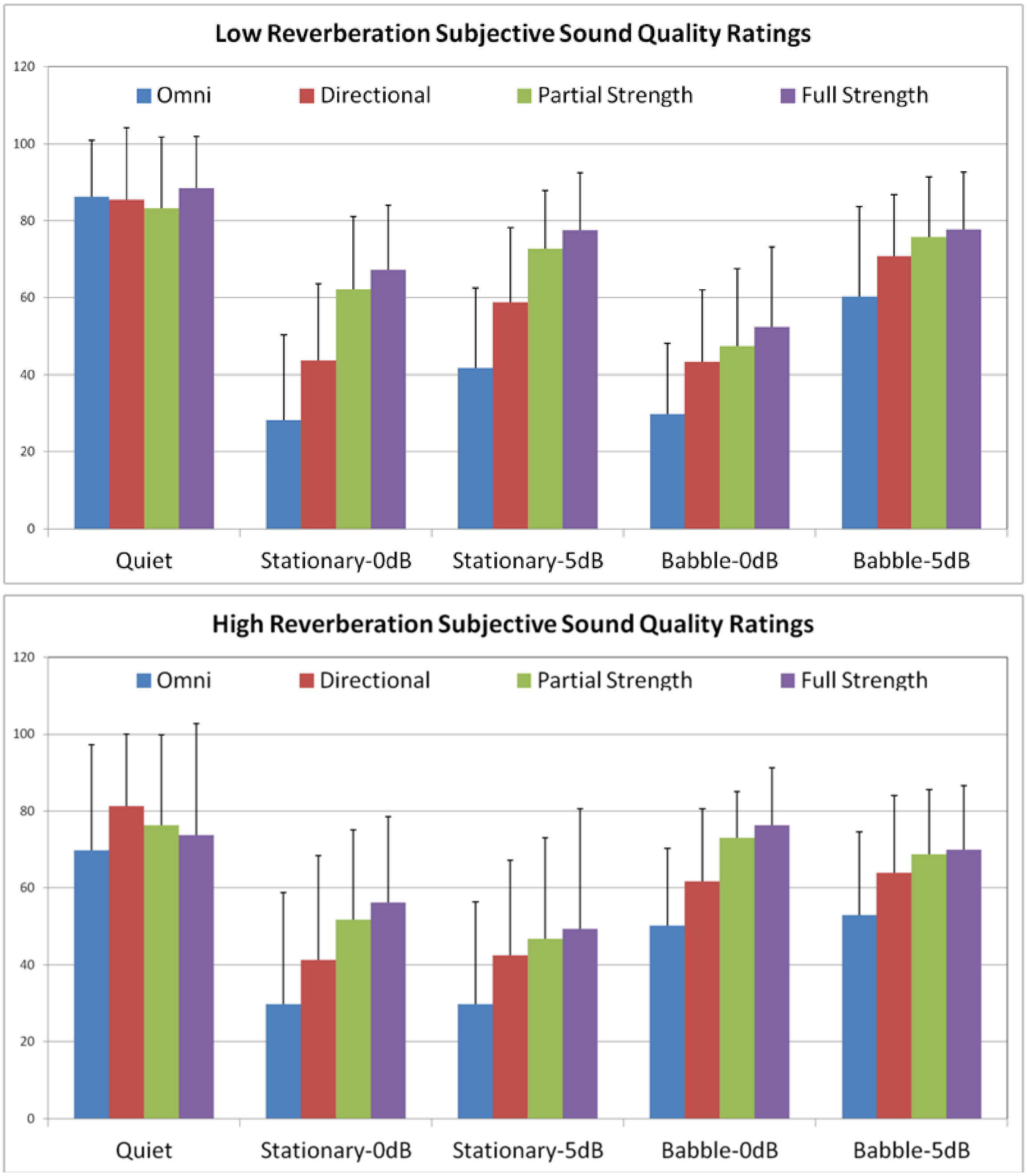



**Figure 3-7: Subjective speech quality ratings for different DHA settings across different noise and reverberation conditions. In general, an improvement in speech quality can be observed with DSP in noisy environments.**

**Figure 3-8: Hearing aid output in response to speech-in-noise stimuli was recorded using the KEMAR. Shown here is the setup in the reverberation chamber.**

The DHAs were programmed to each HL and the same speech and noise stimuli as presented in the subjective data collection procedure were played back and recorded through the DHAs. For each HI subject, a total of 4 (DHA settings) x 2 (noise types) x 2 (SNRs - only 0 and 5 dB were considered) + 4 (DHA settings) in quiet = 20 recordings were collected in each reverberant environment. Figure 3-9 depicts the spectrograms computed from a sample set of DHA recordings for visual inspection of DHA processing. In this figure, panel (a) shows the spectrogram of the first three sentences in quiet, panel (b) shows the spectrogram of the DHA output at 5 dB SNR and omnidirectional setting, panel (c) displays the spectrogram of DHA output when adaptive directionality is enabled for the same noisy condition, and panel (d) shows the spectrogram when all DSP features were operating at their maximum strength. It is evident that the clarity of the time-frequency components belonging to the input speech (harmonicity, formant tracks etc.) have improved substantially between panels (b) and (d).

Both the HASQI and SRMR-HA parameters were computed from the left and right DHA recordings and averaged. For each DHA recording, a proper reference signal was created to facilitate the HASQI computation. This reference signal was generated by applying a FIR filter to the original clean speech signal. The digital filter was designed to match the DSL targets specific to that particular DHA recording (i.e., hearing loss and presentation level).

Once again an insight into DHA processing can be obtained through observation of modulation spectral distributions, as shown in Figure 3-10.

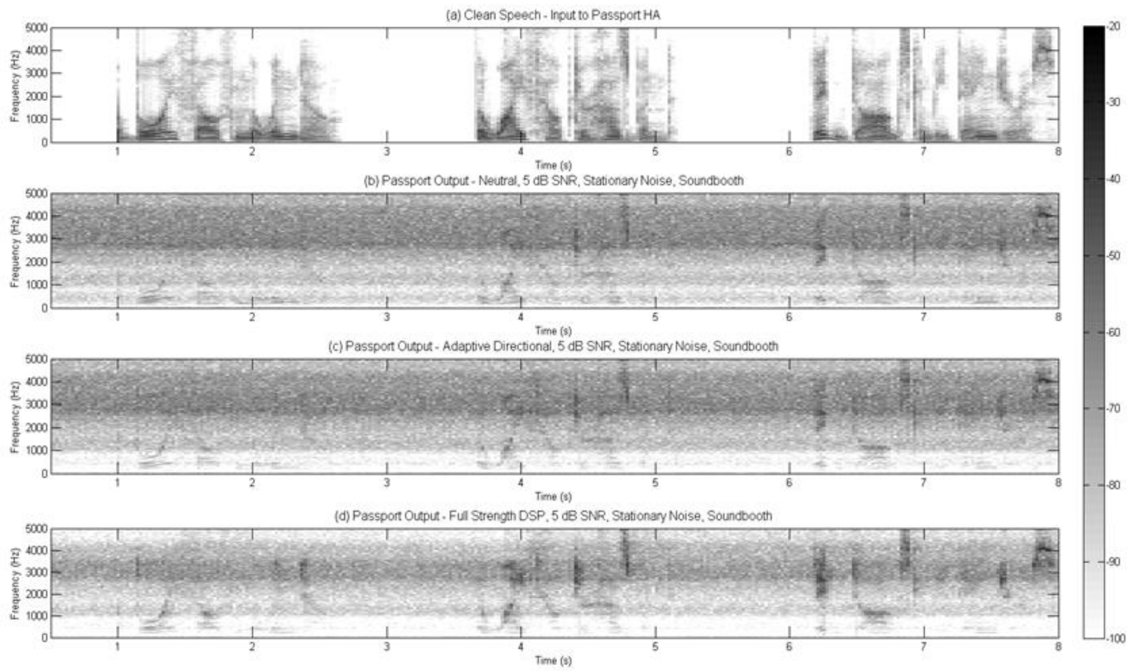



**Figure 3-9: Spectrograms of the DHA recordings in the sound booth with the DHA programmed to the four different settings. Data are from the second database.**

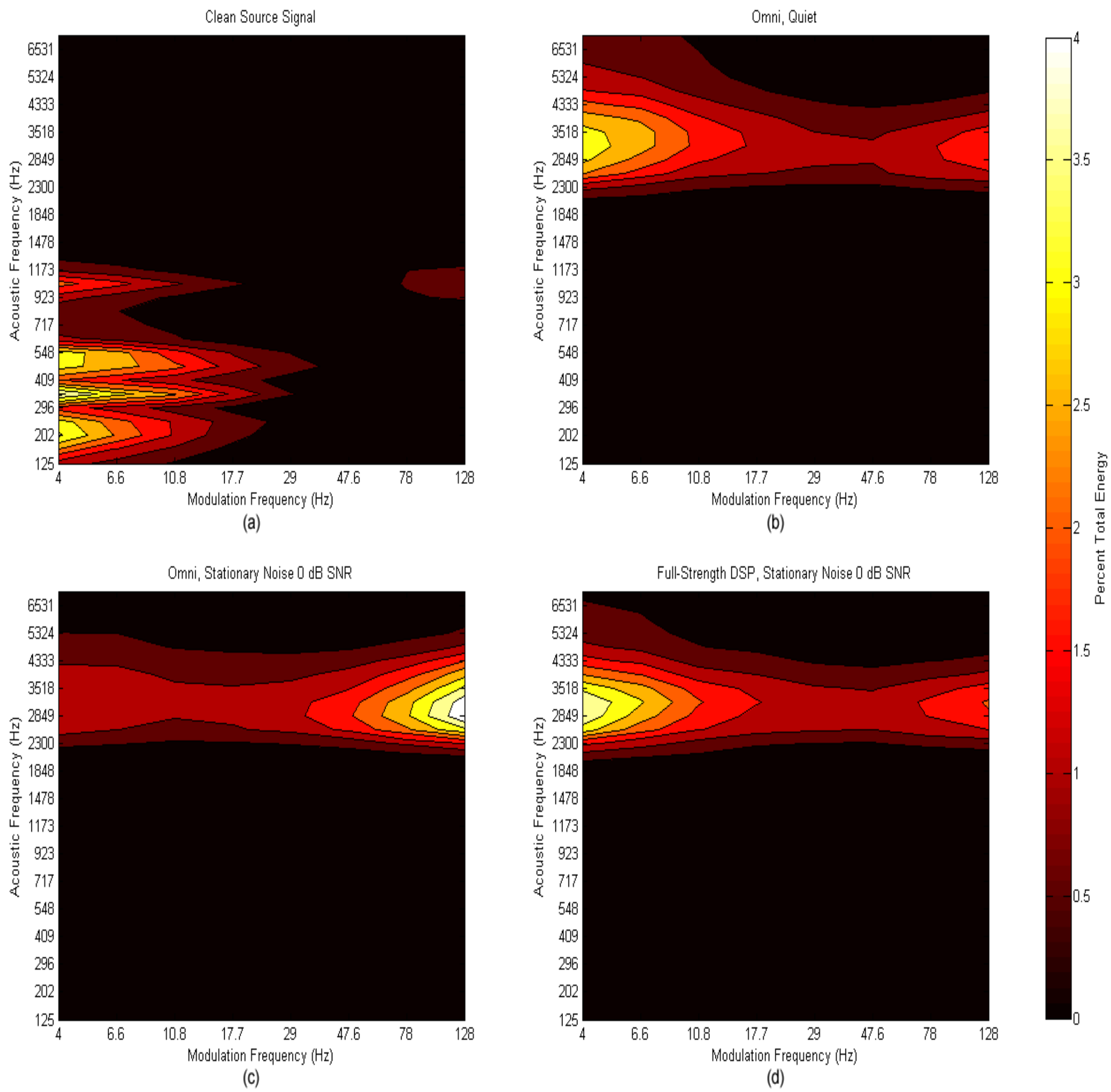



**Figure 3-10: Energy distribution across modulation and acoustic frequencies in the sound booth for a) the clean source signal, b) the DHA in omnidirectional mode with no noise, c) the DHA in omnidirectional mode with stationary noise at 0 dB SNR, d) the DHA in the full strength DSP setting.**

Here, the relative level of different modulation frequency components across the gammatone filterbank (represented by the centre frequency and labeled as “acoustic frequency”) is depicted. Figure 3-10a displays the “modulation spectrogram” of the DHA input, with predominant modulation energy below 10 Hz, as expected for a clean speech sample.  Figure 3-10b shows the modulation energy distribution of DHA output in omnidirectional mode when there is no background noise. The shift in the modulation components to a higher acoustic frequency is due to the DHA frequency shaping, but the dominant modulation components remain below 10 Hz, indicating that the speech components are preserved. The addition of noise, however, shifts the modulation energy towards higher frequencies. Additional signal processing combats this shift and the resultant modulation spectrogram shown in Figure 3-10d has a closer resemblance to the quiet version shown in Figure 3-10b.

Figure 3-11 displays the scatter plots between the objective and subjective metrics in the two reverberation environments.

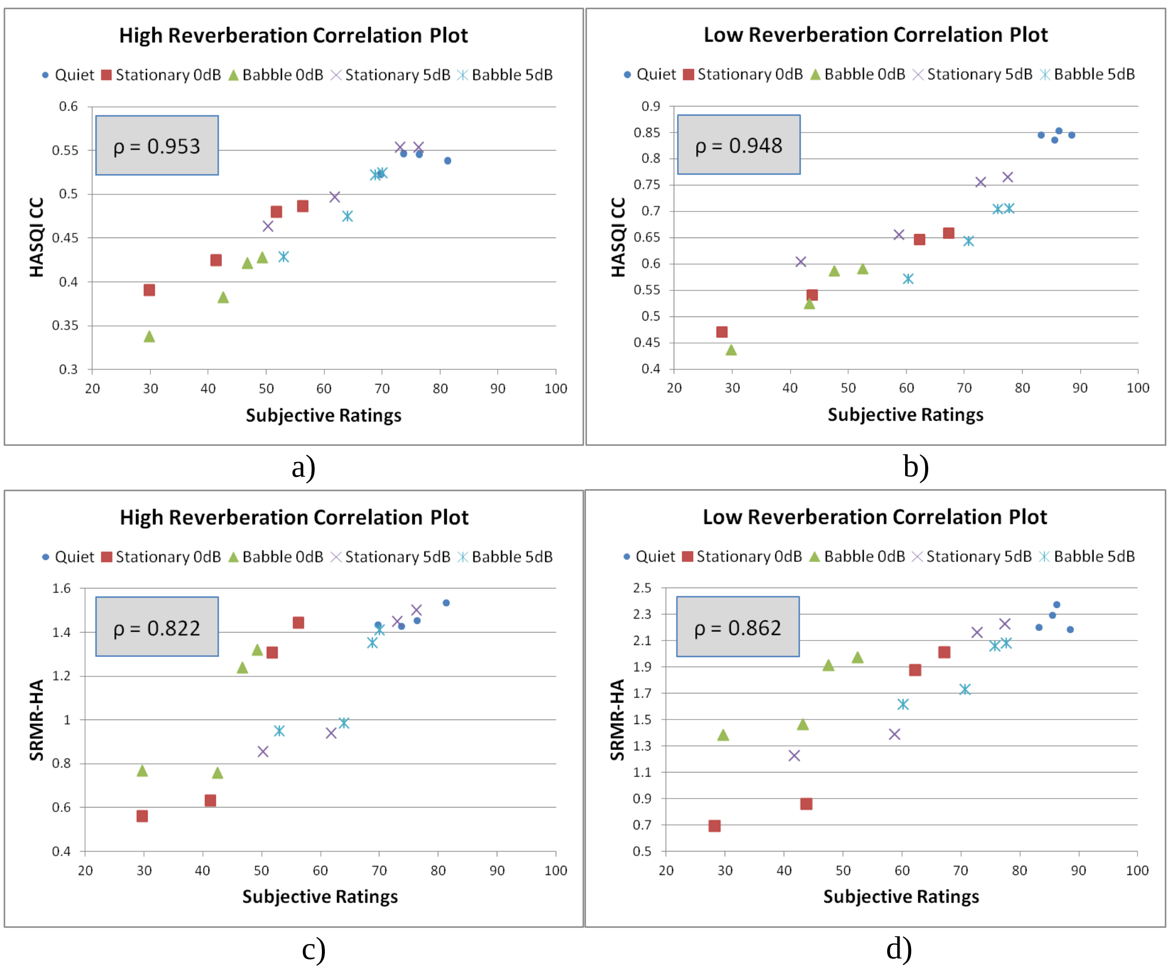



**Figure 3-11: Scatter plots displaying the relation between the objective metrics and subjective speech quality scores across the two reverberation environments.**

Figure 3-11a and Figure 3-11b show the results for the HASQI, while Figure 3-11c and Figure 3-11d display results for the SRMR-HA parameter. In both cases, a linear relationship between the objective and subjective metrics can be seen, with a higher value denoting better perceived quality. The degree of correlation between the objective and subjective metrics was assessed through correlation coefficients. HASQI performed well, explaining roughly 90% of variance in the speech quality ratings across both environments. In contrast SRMR-HA performed modestly with an average of 70% variance explained.

## 3.5 Discussion & Conclusions

This chapter addressed a topic that has not received much attention within the hearing aid research field, viz. objective estimation of DHA speech quality based only on the DHA output. Such an estimate has several advantages: (a) it precludes the need for a separate reference signal that is properly formatted in the temporal and spectral domains, and (b) it allows for real-time fine-tuning of DHA processing parameters through online monitoring of the quality of the DHA output.

The proposed reference-free metric was SRMR-HA, which was a modification and extension of the SRMR metric [70]. The implementation of the gammatone filterbank and the envelope extraction in SRMR-HA are different from SRMR. Furthermore, SRMR-HA incorporates a model for cochlear hearing loss. In order to see whether these enhancements led to an improvement in prediction performance, the correlations of the original SRMR and SRMR-HA with the speech quality ratings were compared. For the bilateral DHA database, the correlations improved from 0.56 to 0.63 and 0.50 to 0.59 for the low- and high-reverberant environments respectively. Similarly, for the second database the coefficients increased from 0.75 to 0.86 for the sound booth data, and 0.7 to 0.81 for the reverberation chamber data. Thus the proposed modifications enabled better prediction of speech quality ratings obtained from HI listeners.

Even with the improvements, the correlation coefficients for the SRMR-HA were inferior to those reported by HASQI. An investigation into the correlation data revealed that the performance of SRMR-HA was poor in situations where either the background noise had modulation patterns mimicking those of speech, or speech modulation patterns themselves were compromised. In such situations, an alternative approach to boost the performance of SRMR-HA is to enrich the feature set extracted from the DHA output. By combining multiple features, each potentially tapping into different perceptual attributes that make up the overall speech quality, a better performance can be obtained. A preliminary investigation along this line of thought was conducted. Results showed that by linearly combining the mean and variance of modulation filterbank output energies, a significantly better performance was obtained.

In summary, a reference-free speech quality metric, SRMR-HA, was applied for the first time to DHA recordings. The correlations with subjective speech quality ratings were modest, with HASQI performing the best. Nonetheless, these initial results hold promise for further enhancement of the performance of SRMR-HA through feature set augmentation and better feature mapping techniques.

# Chapter 4

# 4 Electroacoustic Evaluation of Hearing Aid DNR Algorithms

The previous two chapters have investigated two objective metrics of DHA speech quality and their relationship to perceptual ratings from HI listeners. This chapter exploits the good correlation between the objective and subjective data presented in the previous two chapters to compare and contrast different DHA models. In particular, a framework is developed wherein objective metrics of speech quality and speech intelligibility are used to verify and benchmark DNR performance in a hearing aid test box.

## 4.1 Background

DNR is a feature of many modern digital hearing aids. The aim of DNR is to minimize the amount of noise present in the DHA output signal, as it is well known that noise commonly causes discomfort and reduced intelligibility for HI individuals. Attempts to incorporate noise reduction into hearing aids have been ongoing for many years. Certain analog models from as far back as the 1970s included a switch that would activate a high pass filter with the goal of removing unwanted noise [11]. Unfortunately, the degree of benefit provided by DNR remains unclear. A number of studies have examined the effects of DNR under specific conditions and in some cases a benefit was identified. Specifically, Ricketts and Hornsby [33] conducted a subjective experiment that identified a significant sound quality preference for when the DNR feature of a specific DHA was enabled versus disabled. Bentler et al. [54] found that DNR caused a significant improvement in ease of listening. Sarampalis et al. [77] studied the ability of normal hearing individuals to perform simultaneous tasks while identifying words in noisy signals and concluded that DNR improved the simultaneous task performance. Oliveira et al. [78] found that a specific noise reduction algorithm caused a significant improvement in speech intelligibility. Pittman [79] found that HI children, ages 11-12, experienced significant improvement in their ability to learn words with DNR enabled in a noisy environment. In another study, Pittman [80] found that children gained an improvement in their ability to categorize words while subjected to auditory and visual distractions when using a DNR enabled DHA. Chung [81] found that a modulation based DNR algorithm was effective at reducing wind noise. In contrast, a number of studies, some the same as those mentioned above, have concluded that DNR is not beneficial in certain situations where it might have been expected that a benefit would be seen. Bentler [54] et al. found that DNR did not offer a significant improvement in listening comfort. Sarampalis et al. [77] examined the effect of DNR on speech intelligibility and found no significant improvement. Quintino et al. [82] found no significant benefit offered by a DNR algorithm used by subjects for speech in noise signals. Pittman [80] found no benefit for children ages 9-10 to learn words with DNR enabled in a noisy environment. Stelmachowicz et al. [83] studied speech perception of children with DNR and found no significant improvement. McCreery et al. [84] conducted a review of the literature that included the benefit seen by children from DNR and concluded that no significant benefit was provided.

As can be seen from the brief review presented above, there is a lack of generality in the reported benefits of DNR. Some of these studies have proposed potential reasons for this including variability in hearing aid performance (time constants, number of channels, sensitivity to modulation, gain applied as a function of frequency) [11], [25], the preferences of individual study participants [33] and the nature of the signals presented [11].

Furthermore, there is currently no validated or standardized procedure for Audiologists to assess the DNR algorithm performance [84]. This lack of standardized measure prevents clinical audiologists from assessing the relative benefits of various devices that offer similar, but not identical, noise reduction algorithms.

Very few studies have undertaken cross-brand comparison of DNR performance. Hoetink et al. [85] investigated the performance of DNR algorithms in twelve different DHA models. Noise reduction performance was assessed using simulated speech and speech-like noise stimuli. Results revealed performance differences among different DHA models in terms of the magnitude of noise reduction, frequency range over which noise reduction was active, the input level threshold for DNR activation, and the time taken to engage the noise reduction algorithm. Moreover, the audiogram used to program the DHAs interacted with the performance of the DNR algorithm. While this study highlighted the differences in DNR algorithm performance among different DHA brands, it did not provide a perceptually valid method of DNR performance assessment, as the measurements were based on simulated speech and noise stimuli and the performance was measured only in terms of the amount of noise reduction. More recently, Houben et al. [86] compared the DNR algorithms in five different DHAs. The response of the DHAs to a composite input stimulus containing speech and multi-talker babble at an SNR of 10 dB, was recorded and its speech quality was estimated using the HASQI. While the authors showed a difference in HASQI scores among the five different DHAs, it was not clear whether these differences will generalize for different noise types and SNRs as well as other audiograms. Moreover, the impact of DNR on speech intelligibility was not measured.

The goal of this study is to develop a novel framework to test the DNR performance of a given DHA in a manner that further exposes the underlying signal properties when compared to previous studies, and provides perceptually valid metrics of DNR performance. The proposed procedure makes use of a signal cancellation technique that allows the output noise and speech signals to be analyzed independently despite the fact that they are presented simultaneously to the DHA. This provides great flexibility in analyzing the DNR performance as will be explained later in this chapter. In addition to a detailed description of the proposed evaluation technique, this study also presents the results of applying the technique to seven commercially available DHAs. Statistical analyses of speech intelligibility and speech quality data are presented to describe the DNR performance in relation to different noise types, SNRs, and audiometric configurations. By developing a more detailed understanding of how particular DHAs affect speech-plus-noise signals, it is expected that the process of fitting a DHA running DNR to the specific needs of a HI individual could be significantly improved.

## 4.2 Test Suite Development

In order to apply the evaluation techniques used for this study, it was first necessary to make speech-plus-noise recordings under various conditions. This section will describe the procedure that was followed.

### 4.2.1 Apparatus

The equipment and interconnections used to make the speech-plus-noise recordings are shown in Figure 4-1. As can be seen, the configuration provides for one playback channel and two recording channels. Beginning at the left side of the diagram, custom Matlab software was written on the Personal Computer (PC) for the playback and recording of the digital signals. The PC is connected via Universal Serial Bus (USB) to a Sound Devices USBpre 2 sound card which handles the digital-to-analog conversion in the playback path and the analog-to-digital conversion in the recording path. The sound card output is connected to a Tucker Davis Technologies PA5 programmable attenuator which is used to control the level of the playback signal via a USB connection to the PC. Finally, the attenuated playback signal is connected to an output speaker found within an Interacoustics Dedicated Test Chamber (DTC) TBS25 M/P. The recording path begins within the DTC where the hearing aid under test is connected to an IEC 126 2CC coupler which in turn is connected to one of two G.R.A.S. 40AG pressure microphones. The second pressure microphone is used to capture a reference version of the signal. The two recorded signals next pass through a pre-amplifier, before returning to the USBpre 2.

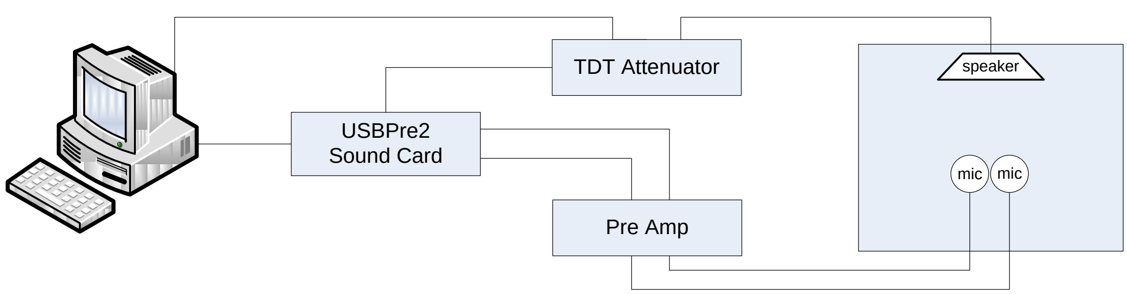



**Figure 4-1: Recording setup.**

### 4.2.2 Stimuli

The stimuli used for this study consisted of speech combined with various types of noise. The International Speech Test Signal (ISTS) [87] was chosen as the standard speech signal. This was combined with each of the following three types of noise: speech shaped noise (SSN), multi-talker babble and traffic noise. Stimuli were created at both 0 dB and 5 dB SNR and in addition a recording was made with clean speech. This resulted in a total of seven distinct playback signals.

### 4.2.3 Hearing Aids

Recordings were performed with the following hearing aids: Siemens Motion 700 P (Motion) Oticon Agil P (Agil), Starkey S Series iQ (SiQ), Phonak Ambra Micro P (Ambra), Unitron Passport Serial (Passport), Widex M440-9 (M440-9) and Sonic Innovations Velocity (Velocity). For each of the seven playback stimuli listed in section 4.2.2, recordings were created with the hearing aid fit to each of three standard audiograms; a moderately sloping mild loss (labeled as N2), a steeply sloping moderate/severe loss (labelled as S3) and a moderately slopping moderate/severe loss (labelled as N4) as defined in Bisgaard et al. [88]. In addition, all recordings were made both with the DNR enabled and with the DNR disabled. This resulted in a total of forty-two recordings per hearing aid.

### 4.2.4 Recordings

In order to evaluate the intelligibility and quality of the speech portion of the recorded signal, it was necessary to extract the speech from the combined speech-plus-noise signal. This was accomplished using an approach described in Wu and Bentler [89]. Each desired speech-plus-noise condition is recorded twice, but for the second recording the noise signal is inverted. The two resulting recordings are then aligned and added together which results in the cancellation of the noise portion and a doubling of the speech portion. Dividing the result by two yields a very close representation of the speech portion of the signal, where a small error will exist due to imperfect alignment, distortion due to system nonlinearities (if any) and system noise. In order to evaluate the attack time of the DNR and for speech intelligibility calculations, it was necessary to extract the noise portion of the speech-plus-noise signal. This was accomplished by following the same procedure as outlined for the speech extraction, except that the inverted noise signal is subtracted from the primary signal rather than added. A block diagram of the procedure is shown in Figure 4-2. Ellaham et al. [90] validated this technique with nonlinear hearing aids in a recent publication.

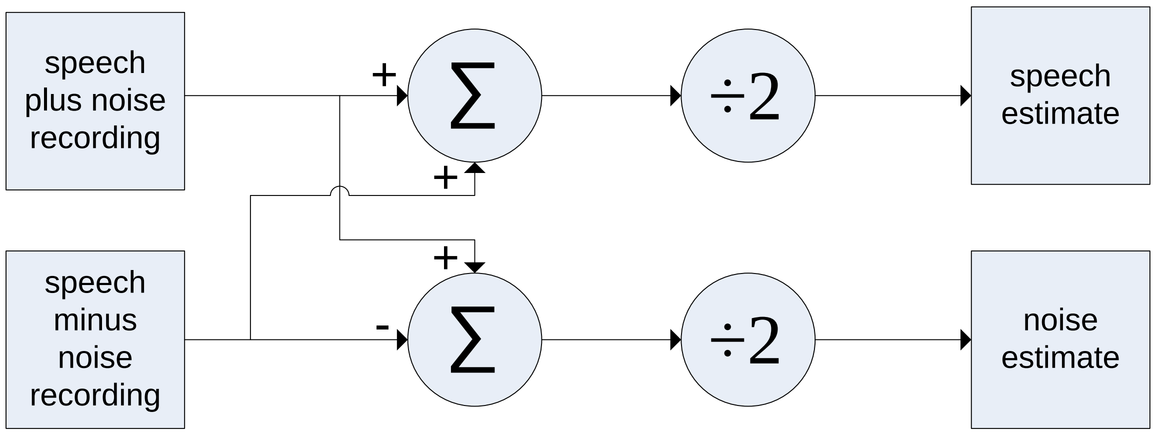



**Figure 4-2: Speech and noise extraction from speech plus noise signal.**

## 4.3 Test Methodology

As described above, one of the signal cancellation technique outputs is the speech only portion of the speech-plus-noise signal. For this study, three electroacoustic methods of evaluation were chosen to investigate the effect of DNR on the speech only signal.

The first was the Speech Intelligibility Index (SII) as defined in [40] for the evaluation of speech intelligibility. The approach taken by this metric is to calculate the predicted speech intelligibility according to the following equation:

$$S = \sum_{i=1}^{n} I_i A_i \tag{4.1}$$

where $S$ is the predicted speech intelligibility, $n$ is the number of computational bands, $I_i$ is the band importance function, $A_i$ is the band audibility function and $i$ indexes the frequency bands. The SII standard includes four different frequency band options, for this study the one-third octave band procedure was used. The band audibility function specifies for each frequency band the proportion of the speech dynamic range that adds to the intelligibility under less than ideal conditions. In order to calculate the band audibility function, it is necessary to determine a number of input vectors. The first is the equivalent speech spectrum level which specifies the level of the speech only portion of the signal. The second input, the equivalent noise spectrum level, is similarly defined except that it is based on the noise portion (if any) of the signal. The third is the equivalent hearing threshold level which consists of the hearing thresholds of the listener for whom the SII is being calculated. In order to calculate the SII for hearing aid speech-plus-noise output signals, it is clearly necessary to have isolated speech and noise signals. Since this is not naturally available from a hearing aid recording, an approach such as the signal cancellation technique described in section 4.2.4 must be employed. It must be noted here that the spectrum levels used in SII calculation in this thesis are referred to the 2 cc coupler.

Based on the input vectors, the SII accounts for a number of different factors that influence the audibility. These include the internal noise of the auditory system, masking effects of both the speech and noise and the level dependent speech distortion. This allows the SII to account for adverse conditions including noisy and reverberant environments, loud presentation levels and levels below hearing thresholds across the frequency range. The purpose of the band importance function is to weight the band audibility function in each frequency band according to the contribution that each band makes on average to speech intelligibility [40].

The second electroacoustic measure used for this study was the SRMR-HA as outlined in section 3.2. This approach relies on a cochlear model to extract relevant speech quality features, and computes an estimated quality score based on a ratio of modulation frequencies that can be attributed to speech and modulation frequencies that can be attributed to noise.

The third electroacoustic measure employed by the study was the HASQI CC as introduced in section 2.4, which is based on the HASQI introduced in section 2.2.2. Similar to the SRMR-HA, this approach relies on a cochlear model to extract relevant speech quality features. The HASQI-CC differs from the SRMR-HA in that it fits the extracted features to a set of cepstral basis functions for both the processed signal and a clean reference signal and then computes an average of their correlation values to predict the sound quality.

Aside from the speech only signal, the other output from the signal cancellation technique is the noise only signal which was used in this study to determine the DNR attack times. Attack time is defined as the amount of time necessary for the DNR algorithm to reduce the noise to a level that is within 3 dB of the steady state level. Figure 4-3 below shows the noise only signal extracted from two hearing aids with significantly different attack times. As can be seen, the top panel shows an attack time on the order of 2-3 seconds, while the bottom panel demonstrates an attack time that is closer to 20 seconds. To determine the attack time of the DNR employed by each of the hearing aids, the overall level of the noise only signal was determined in blocks of length 125 ms. Starting with the beginning of the signal, the first block that was found to have a level reduced to within 3 dB of the steady state level was considered to be the end of the attack period.

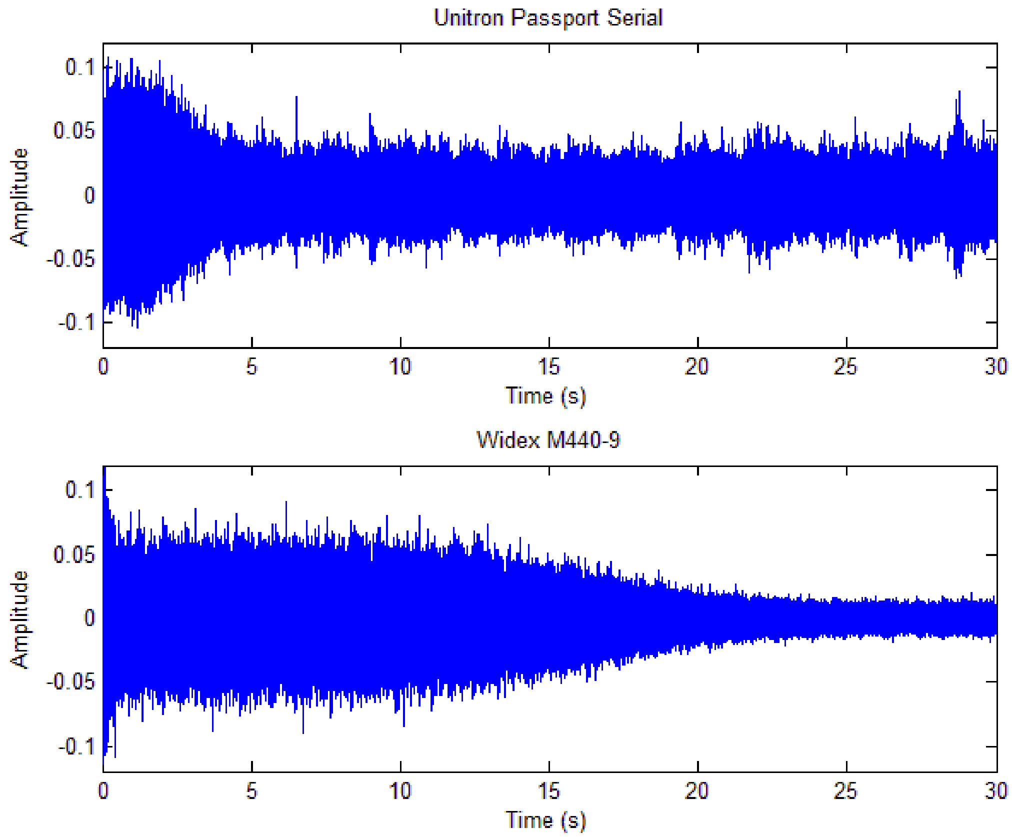



**Figure 4-3: Noise only signal for two hearing aids with different attack times.**

## 4.4 Results

This section includes a limited set of representative results. The remainder of the results are available in Appendix A.

Figure 4-4 presents two spectrograms that compare the DNR OFF with the DNR ON condition. It is evident from this comparison that the DNR has reduced the noise content from approximately the 2-3 second mark and onward.

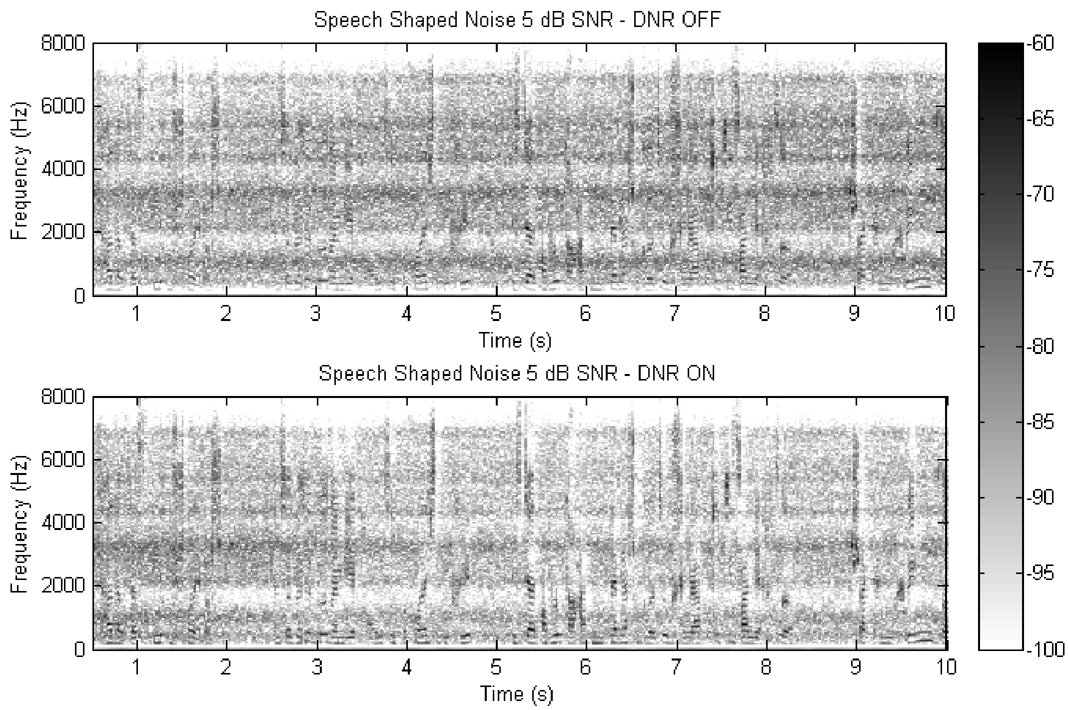
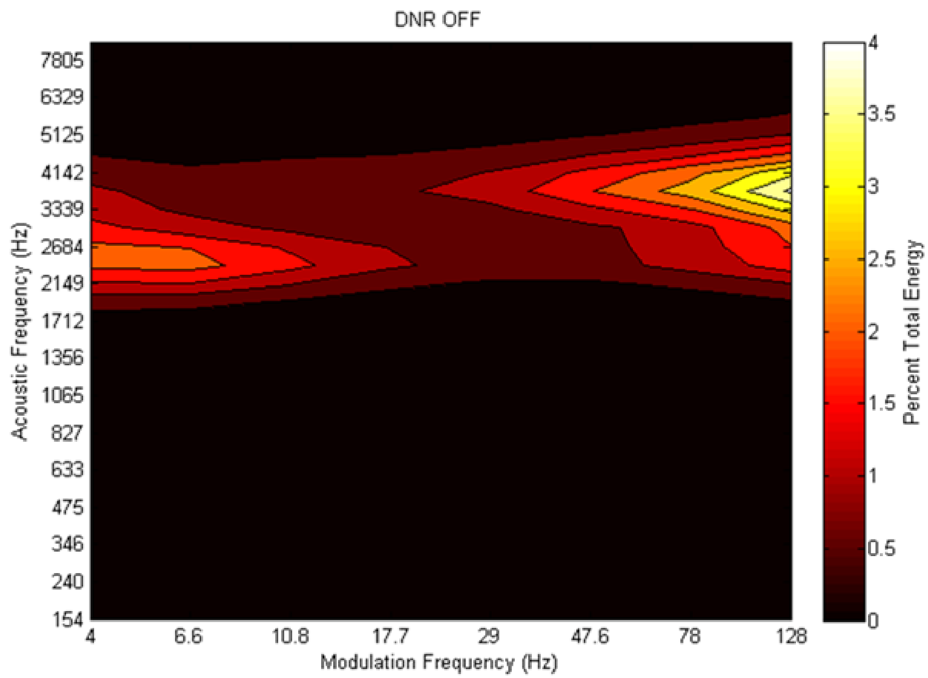
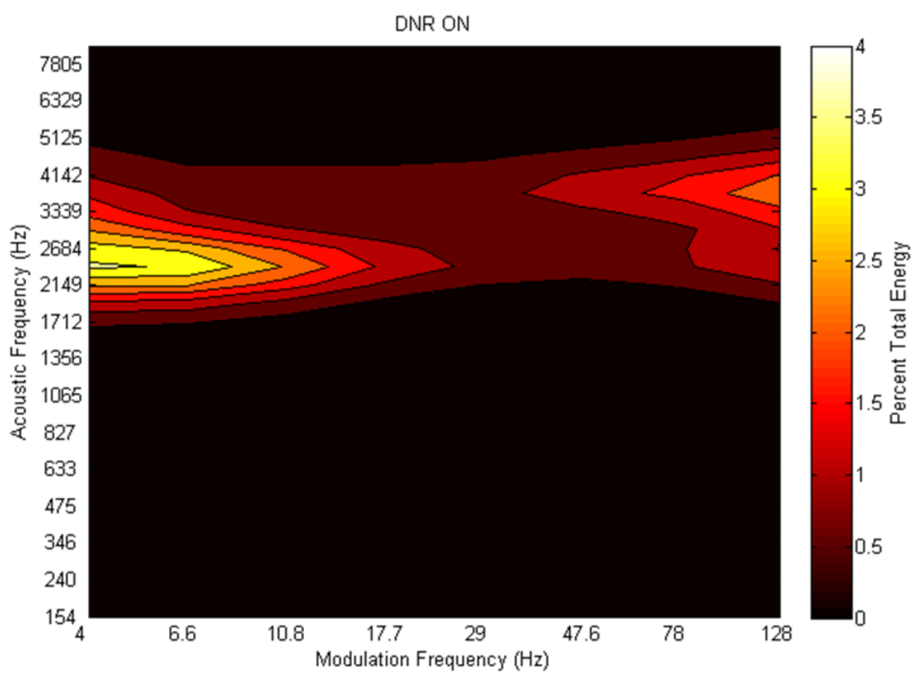



**Figure 4-4: Spectrograms for the DNR Off and DNR On settings - Siemens Motion DHA.**

Figure 4-5 presents a comparison between the modulation energy plots for the DNR OFF and DNR ON conditions. As can be seen, with DNR OFF, the modulation energy is more focused in the upper four modulation frequency bands, whereas with the DNR ON, it is clear that the energy predominately resides in the lower four modulation bands.




a)




b)

**Figure 4-5: A comparison of the modulation energy plots for a) the DNR OFF and b) DNR ON conditions.**

In order to visualize the degree of improvement offered by the DNR algorithms of the hearing aids tested, scatter plots for two noise conditions and two electroacoustic metrics are included below. All four plots were generated from N2 audiogram data.

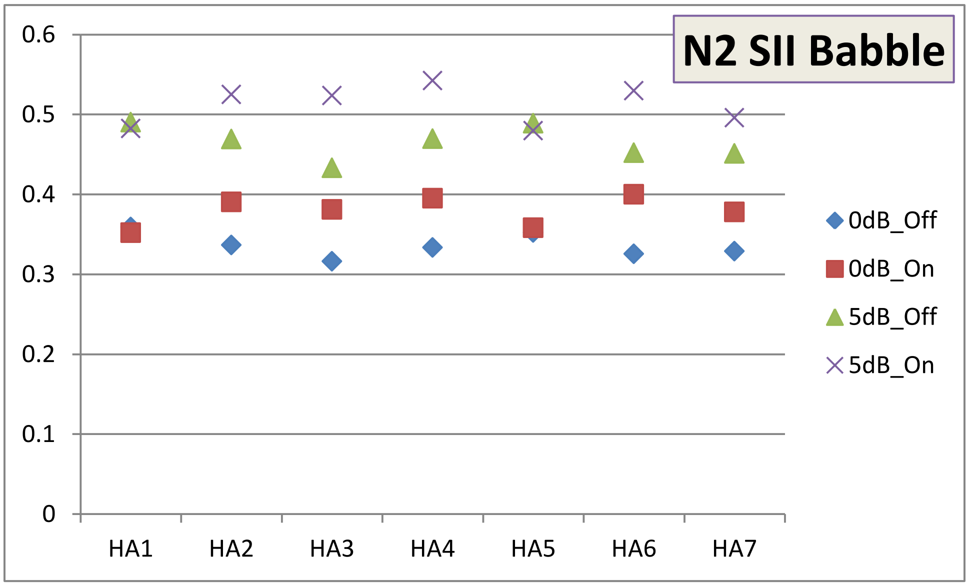



**Figure 4-6: Comparison of SII values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

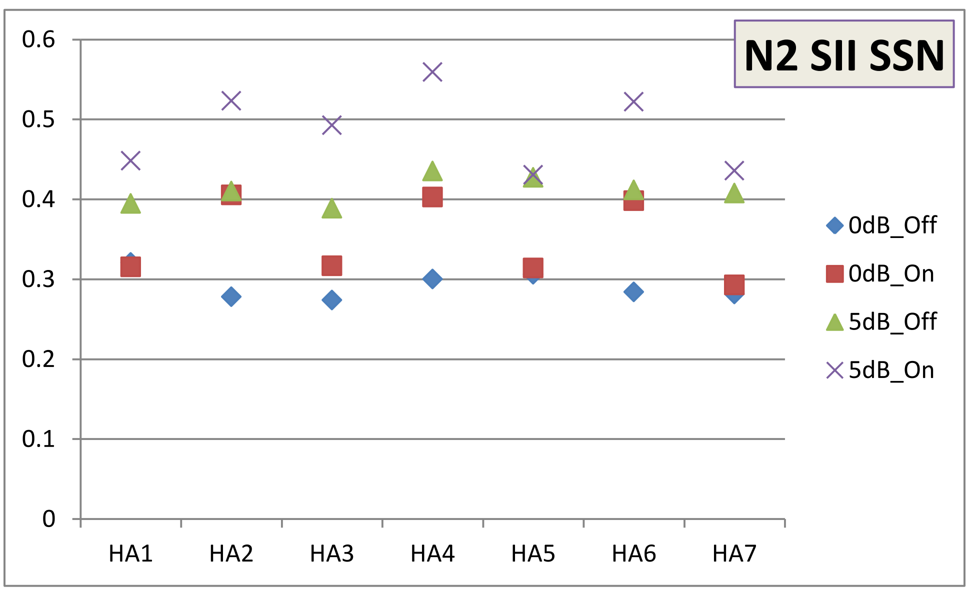



**Figure 4-7: Comparison of SII values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

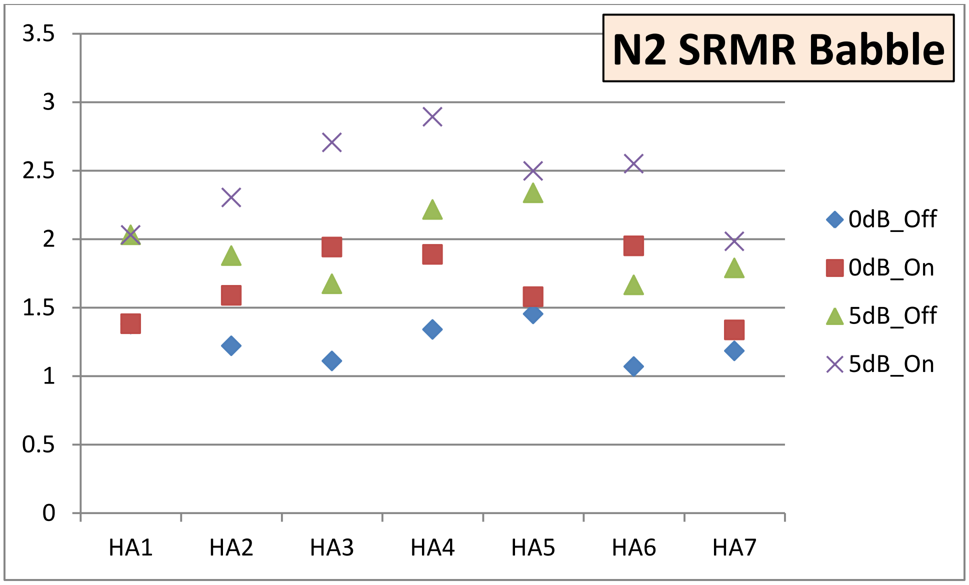



**Figure 4-8: Comparison of SRMR values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

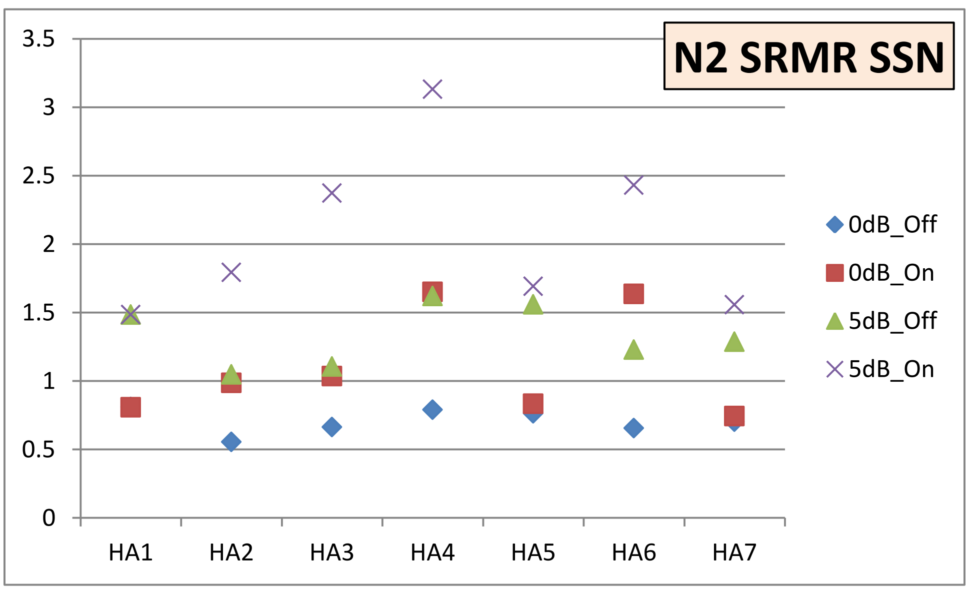



**Figure 4-9: Comparison of SRMR values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

Table 4-1 presents all of the electroacoustic ratings for the N2 audiogram. Similar tables for the S3 and N4 audiograms are included in Appendix A.

**Table 4-1: Electroacoustic measures of DNR performance for the N2 audiogram.**

| Noise Condition | HA1 | | HA2 | | HA3 | | HA4 | | HA5 | | HA6 | | HA7 | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON |
| SII | | | | | | | | | | | | | | |
| Babble 0dB | 0.36 | 0.35 | 0.34 | 0.39 | 0.32 | 0.38 | 0.33 | 0.40 | 0.35 | 0.36 | 0.33 | 0.40 | 0.33 | 0.38 |
| Babble 5dB | 0.49 | 0.48 | 0.47 | 0.53 | 0.43 | 0.52 | 0.47 | 0.54 | 0.49 | 0.48 | 0.45 | 0.53 | 0.45 | 0.50 |
| SSN 0dB | 0.32 | 0.32 | 0.28 | 0.41 | 0.27 | 0.32 | 0.30 | 0.40 | 0.31 | 0.31 | 0.28 | 0.40 | 0.28 | 0.29 |
| SSN 5dB | 0.39 | 0.45 | 0.41 | 0.52 | 0.39 | 0.49 | 0.44 | 0.56 | 0.43 | 0.43 | 0.41 | 0.52 | 0.41 | 0.44 |
| Traffic 0dB | 0.33 | 0.32 | 0.30 | 0.39 | 0.28 | 0.34 | 0.30 | 0.38 | 0.34 | 0.32 | 0.29 | 0.38 | 0.29 | 0.30 |
| Traffic 5dB | 0.46 | 0.45 | 0.44 | 0.51 | 0.40 | 0.50 | 0.43 | 0.53 | 0.45 | 0.44 | 0.42 | 0.51 | 0.41 | 0.44 |
| SRMR | | | | | | | | | | | | | | |
| Babble 0dB | 1.38 | 1.38 | 1.22 | 1.59 | 1.11 | 1.94 | 1.34 | 1.89 | 1.45 | 1.58 | 1.07 | 1.95 | 1.18 | 1.34 |
| Babble 5dB | 2.03 | 2.03 | 1.88 | 2.30 | 1.67 | 2.71 | 2.22 | 2.89 | 2.34 | 2.50 | 1.67 | 2.55 | 1.79 | 1.98 |
| SSN 0dB | 0.81 | 0.81 | 0.56 | 0.99 | 0.67 | 1.04 | 0.79 | 1.65 | 0.77 | 0.84 | 0.66 | 1.64 | 0.70 | 0.75 |
| SSN 5dB | 1.49 | 1.49 | 1.05 | 1.79 | 1.11 | 2.37 | 1.62 | 3.13 | 1.56 | 1.69 | 1.23 | 2.43 | 1.29 | 1.56 |
| Traffic 0dB | 0.77 | 0.77 | 0.65 | 1.05 | 0.62 | 0.90 | 0.69 | 1.28 | 0.75 | 0.79 | 0.60 | 1.27 | 0.67 | 0.67 |
| Traffic 5dB | 1.41 | 1.41 | 1.22 | 1.87 | 1.07 | 2.08 | 1.45 | 2.49 | 1.55 | 1.67 | 1.12 | 2.07 | 1.22 | 1.45 |
| HASQI | | | | | | | | | | | | | | |
| Babble 0dB | 0.51 | 0.51 | 0.49 | 0.52 | 0.44 | 0.47 | 0.51 | 0.57 | 0.50 | 0.51 | 0.45 | 0.54 | 0.45 | 0.46 |
| Babble 5dB | 0.67 | 0.67 | 0.66 | 0.66 | 0.59 | 0.62 | 0.68 | 0.71 | 0.68 | 0.69 | 0.62 | 0.68 | 0.62 | 0.65 |
| SSN 0dB | 0.46 | 0.46 | 0.45 | 0.51 | 0.41 | 0.45 | 0.44 | 0.56 | 0.45 | 0.46 | 0.43 | 0.52 | 0.41 | 0.32 |
| SSN 5dB | 0.61 | 0.61 | 0.60 | 0.64 | 0.55 | 0.59 | 0.61 | 0.71 | 0.62 | 0.63 | 0.57 | 0.65 | 0.57 | 0.55 |
| Traffic 0dB | 0.46 | 0.46 | 0.47 | 0.53 | 0.42 | 0.44 | 0.44 | 0.54 | 0.47 | 0.47 | 0.43 | 0.52 | 0.41 | 0.32 |
| Traffic 5dB | 0.61 | 0.61 | 0.62 | 0.64 | 0.56 | 0.60 | 0.61 | 0.68 | 0.64 | 0.64 | 0.57 | 0.65 | 0.56 | 0.52 |

Table 4-2 lists the attack times found for each of the seven hearing aids tested while programmed with each of the three standard audiograms and for both 0 dB SNR and 5 dB SNR.

**Table 4-2: DNR attack times listed in seconds.**

| Hearing Aid | Attack Time (s) | | | | | |
|---|---|---|---|---|---|---|
| | N2 Audiogram | | S3 Audiogram | | N4 Audiogram | |
| | 0 dB | 5 dB | 0 dB | 5 dB | 0 dB | 5 dB |
| HA1 | 0.00* | 0.00* | 0.00* | 0.00* | 3.94 | 0.00* |
| HA2 | 2.75 | 2.69 | 3.44 | 3.44 | 3.44 | 3.44 |
| HA3 | 2.75 | 2.75 | 3.44 | 3.44 | 2.50 | 2.44 |
| HA4 | 2.31 | 3.37 | 3.37 | 3.44 | 3.37 | 11.4 |
| HA5 | 0.00* | 0.00* | 1.50 | 2.69 | 0.81 | 0.44 |
| HA6 | 2.75 | 2.69 | 2.75 | 2.06 | 2.87 | 2.69 |
| HA7 | 21.6 | 18.2 | 19.5 | 17.2 | 21.3 | 18.7 |

*zero values indicate that the hearing aid did not reduce the noise level by at least 3 dB for the length of the recording

## 4.5 Statistical Analyses of DNR performance

While the ISTS signal has the benefit of being a standardized speech stimulus, it does not lend itself for statistical characterization of DNR performance, as single values of SII, SRMR-HA, and HASQI are calculated for each condition (noise type, SNR, and audiogram). In order to apply repeated-measures ANOVA, multiple values are required for the same condition. To accomplish this, a separate set of recordings were obtained from the same group of DHAs. The ISTS signal was replaced by ten different IEEE Harvard speech sentences [19] spoken by five female and five male talkers. The noise type, SNR, and audiometric configuration parameters were the same as the ISTS recordings. The metrics were then calculated from the DHA recordings for each of the ten sentences, and entered into Statistica 10.0 software for repeated measures ANOVA computation.

Table 4-3 and Table 4-4 display the salient main effect and multi-way interactions for the SII and HASQI data respectively. Similar results were obtained with the SRMR-HA data.

**Table 4-3: Results of the repeated measures ANOVA with SII data.**

| | Variable(s) | F | Hypothesis dF | Error dF | *p* |
|---|---|---|---|---|---|
| Main effects | DHA | 345.24 | 6 | 54 | 0.000 |
| | Audiogram | 2745.72 | 2 | 18 | 0.000 |
| | Noise Type | 5.10 | 2 | 18 | 0.018 |
| | SNR | 7251.39 | 1 | 9 | 0.000 |
| Two-way | Audiogram * Noise Type | 32.86 | 4 | 36 | 0.000 |
| | Audiogram * DHA | 560.63 | 12 | 108 | 0.000 |
| | Noise Type * DHA | 34.58 | 12 | 108 | 0.000 |
| | SNR * DHA | 149.12 | 6 | 54 | 0.000 |
| Three-way | Audiogram * Noise Type * DHA | 20.98 | 24 | 216 | 0.000 |
| | Audiogram * SNR * DHA | 47.89 | 12 | 108 | 0.000 |
| | Noise Type * SNR * DHA | 15.83 | 12 | 108 | 0.000 |

**Table 4-4: Results of the repeated measures ANOVA with HASQI data.**

| | Variable(s) | F | Hypothesis dF | Error dF | *p* |
|---|---|---|---|---|---|
| Main effects | DHA | 62.65 | 6 | 54 | 0.000 |
| | Audiogram | 89.61 | 2 | 18 | 0.000 |
| | SNR | 1856.54 | 1 | 9 | 0.000 |
| Two-way | Audiogram * SNR | 33.88 | 2 | 18 | 0.000 |
| | Audiogram * DHA | 67.68 | 12 | 108 | 0.000 |
| | Noise Type * DHA | 12.38 | 12 | 108 | 0.000 |
| | SNR * DHA | 23.86 | 6 | 54 | 0.000 |
| Three-way | Audiogram * Noise Type * DHA | 8.46 | 24 | 216 | 0.000 |
| | Audiogram * SNR * DHA | 14.51 | 12 | 108 | 0.000 |
| | Noise Type * SNR * DHA | 10.28 | 12 | 108 | 0.000 |

## 4.6 Discussion and Conclusions

The present chapter introduced a framework for verifying and benchmarking DNR algorithms in DHAs. Key features of this framework include a metric to measure the impact of DNR on speech intelligibility, two metrics to assess the impact of DNR on speech quality, and a measure of the DNR activation time. By combining these measurements, an Audiologist can quickly gauge the performance of a DNR algorithm. It must be noted here that this framework, while not applied to DNR assessment before, has been applied to assess the functioning of other DHA DSP features. For example, Kates [91] combined measures of speech intelligibility and speech quality to characterize the behaviour of multichannel WDRC algorithms in DHAs.

Results using the ISTS signal as the speech stimulus showed the differences in DNR performance across DHA models. As can be seen in Figures 4-6 to 4-9 and in Table 4-2, there are performance differences across the DHAs. For example, HA1 neither significantly enhances nor degrades SII or SRMR-HA values across different conditions. However, HA4 improves both the SII and SRMR-HA scores across the same conditions. Similarly, there are a group of DHAs that perform similar to HA4 in the SII domain, but are at lower rung compared to HA4 in the SRMR-HA metric. These results highlight the differences among DNR implementations in DHAs, and the presented framework facilitates the Audiologist to compare and contrast different devices.

In addition, the following general trends can be noted in these results: (a) the SII generally increased with DNR ON, (b) noise level generally decreased with DNR ON, and (c) sound quality of the overall output signal generally increased with DNR ON. The magnitude of these changes were dependent on the noise type, SNR, DHA model, and the audiogram, which is consistent with the noise reduction data presented by Hoetink et al. [85]. Further statistical analyses confirmed the significance of these changes – Table 4-3 and Table 4-4 show several of the two-way and three-way interactions were statistically significant, indicating the multi-dimensional nature of DNR evaluation.

A correlational analysis among the three metrics across different conditions resulted in a Pearson correlation coefficient of 0.67 between SII and HASQI, and 0.80 between HASQI and SRMR-HA. This is to be expected as both HASQI and SRMR-HA are quantifying speech quality degradation and therefore have a higher degree of correlation.

As can be seen from the presented results, the proposed DNR evaluation technique provides an in depth view of the affect that various DNR algorithms have on an array of speech-plus-noise signals. It is clear from these results that when comparing DNR algorithms between different DHA models, there is a high degree of variability in the observed effects on specific aspects of the output signal. By increasing the awareness of the relative performance of DNR offered by state-of-the art DHAs, it should be possible to make an informed fitting decision to best meet the specific needs of the HI individual. Past studies have shown that when comparing sound quality ratings amongst HI individuals, opinions vary significantly [33]. The fact that DNR varies significantly between DHAs may be viewed as advantageous since it affords an opportunity to select a DNR approach that is tailored to the preferences of the individual. As an example, it is clear that certain DHAs offer increased noise reduction at the expense of reduced sound quality in the underlying speech. It would be to the benefit of the user to choose this hearing aid if they had a high aversion to noise and a low aversion to a reduction in speech quality. Conversely, some users may prefer that the sound quality is preserved to the highest extent possible at the expense of less noise reduction. Based on the data provided by the techniques outlined in this study, it should be possible to recommend an appropriate DHA.

# Chapter 5

# 5 Electroacoustic Evaluation of Directional and Bilateral Wireless Hearing Aids

In the previous chapter, an electroacoustic measurement procedure was described for comprehensive assessment of the DNR feature in DHAs. As briefly discussed in Chapter 1, DNR is one of the two features that modern DHAs employ to mitigate the presence of background noise. The other feature is multiband adaptive directionality, wherein DHAs attempt to exploit potential spatial and spectral differences between the desired speech signal and the unwanted background noise. This chapter describes the need for measuring adaptive directionality performance and details the development of a flexible electroacoustic system for the assessment of the adaptive directionality feature.

## 5.1 Background

As mentioned in Chapter 1, adaptive directionality is a DHA feature where the polar pattern is adjusted such that the noise originating from the rear azimuths is reduced the most. The multiband adaptive directionality technique goes one step further, by optimizing polar plots in multiple frequency regions independently and simultaneously. Differences exist among different models of DHAs in terms of the number of simultaneous polar plots, the rules for selecting the appropriate polar plot in different frequency regions, and the speed of activation and adaptation of directionality. For example, Wu and Bentler [92] reported the adaptation times for different DHAs as shown in Table 5-1. The time constants are shown for two different situations: (a) when the DHA switches from omnidirectional to directional mode in response to the start of a noise source emanating from 90°, and (b) when the DHA switches its polar plots when the noise source at 90° is turned off and a new noise source is activated at 180°. The wide range of adaptation time constants, both within and across DHA models, is apparent in this data. Furthermore, Wu and Bentler [92] presented data which showed disparate directional performance from different DHAs, which was both frequency- and level-dependent. A similar report of varied directional performance, not only across, but within different hearing aid brands is presented by Ricketts [8].

**Table 5-1: This table shows the adaptation times for 5 different hearing aids under two conditions. The first is when the environment changes from silence to having a noise at 90° and the second is when the environment changes from having a noise at 90° to having a noise at 180°. Reproduced from [92].**

| | DHA1 (ms) | DHA2 (ms) | DHA3 (ms) | DHA4 (ms) | DHA5 (ms) |
|---|---|---|---|---|---|
| From silence to 90° | < 10 | 75 | 18000 | 8500 | 3000 |
| From 90° to 180° | <10 | 40 | <10 ms | 4500 | 3000 |

As discussed in Chapter 1, current standards are limited for measuring directional DHA performance, with no standard available to benchmark multiband adaptive directionality. For example, directional DHA performance measurement is out of the scope of the ANSI 3.22 [37] standard, and ANSI S3.35 [39] only specifies procedures for fixed directionality. It is therefore not surprising that commercially available hearing aid test systems do not facilitate measurement of adaptive directionality. For example, the Audioscan Verifit system utilizes two speakers within the test chamber – one for speech and the other for noise. The DHA is placed in the chamber with the front microphone facing the speech speaker. During the playback of speech and noise, secondary short noise bursts are randomly interspersed with either the speech or noise source, so the measurement software can isolate the response of the DHA to signals coming from either the front or the back. This Front-to-Back Ratio (FBR) across different frequencies is utilized as a measure of DHA directivity. While this provides some information on the directional processing abilities of the DHA, it does not quantify its actual directional performance. Moreover, as the speakers and the DHA are fixed, this system cannot measure adaptive directional performance. Similarly, the hearing aid test system from Frye Electronics, Fonix 8120, uses a single speaker for signal presentation and a turntable upon which the DHA is mounted. By rotating the turntable and analyzing the DHA output, the measurement system creates the polar plot. However, this measurement system requires that the DHA be programmed in linear mode (i.e. no WDRC). In addition, it is not feasible to test adaptive directionality.

Wu and Bentler [89], [92] have developed techniques to assess the directionality performance of DHAs in a more rigorous manner. The test apparatus the authors used is shown in Figure 5-1. As can be seen, the jammer speaker and a stand for the DHA were mounted on a turntable while the probe speaker was mounted on a stationary stand. As the turntable rotates, the angle of arrival of the probe varies, while the jammer remains fixed with respect to the DHA. The purpose of the jammer speaker is to freeze the polar pattern of the DHA by delivering a high level noise signal. The probe speaker emits a lower level signal and the probe signal power in the hearing aid output is calculated as the turntable (and the DHA + jammer speaker combination) rotates. The probe signal power is determined using the signal cancellation technique described in Chapter 4. For every angular position of the turntable, two recordings are made: one with the probe and jammer, and the other with the probe and inverted jammer signals. If the procedure was perfectly repeatable and free of external interference, then the addition of the two recordings would result in a complete cancellation of the jammer signal and a doubling of the probe signal. Adaptive directionality can be assessed in this system by adjusting initial angular orientation of the jammer speaker relative to the DHA. Wu and Bentler [92] later enhanced this method by utilizing impulse sounds as probe signals, which allow "snapshot" measurements of DHA directivity. While the techniques proposed by Wu and Bentler are more robust and rigorous, they are still imbued with the following limitations: (a) the DHA must be programmed to operate in linear mode; (b) it is not feasible to test multiband adaptive directionality; and (c) the impact of bilateral adaptive directionality on sound localization cues is not measured. The last point is elaborated on in the next few paragraphs.

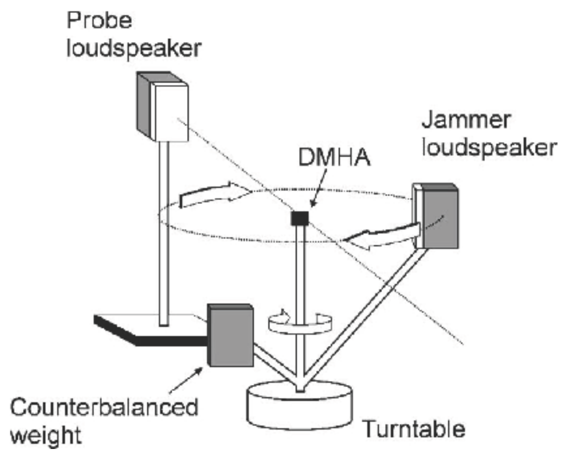



**Figure 5-1: The apparatus used to implement the signal cancellation technique in [89]. Here "DMHA" stands for Directional Microphone Hearing Aid.**

### 5.1.1 Sound Localization

The ability to accurately determine the direction of arrival of an incoming sound is important for many reasons. Commonly referred to as sound localization, this ability allows listeners to focus their attention on a sound of interest, which improves speech intelligibility and allows for proper communication through the use of visual cues such as facial expressions and body gestures. In a busy environment, proper sound localization ability can allow listeners to avoid dangerous events by turning their attention in the correct direction in time to avoid any potential harm. Sound localization is mediated by the timing and level differences between the signals received at each ear, as well as the spectral shaping provided by the pinna. This difference can be attributed to two factors, the ITD and the ILD [93]. As will be explained, hearing aids can produce side effects that adversely impact the ITD and ILD.

For sound frequencies below approximately 1500 Hz, it is the ITD that is predominant in the sound localization process. Since sound propagates at a fixed speed (340.29 m/s at sea level), there is usually a small time difference between when a given sound arrives at each ear. The only exception could occur when the sound originates from a location that is equidistant from each ear which for humans would be directly in front or directly behind the listener. By determining the delay between the two received signals and comparing it to learned values that are part of early development, the auditory processing area of the brain is able to determine the source direction [93].

For frequencies above 1500 Hz, it becomes difficult to resolve the exact time delays due to the fact that more than one wavelength occurs within the distance separating the two ears. This is when the ILD becomes important for sound localization. To understand this functionality, it is first important to realize that with increasing frequencies, the attenuation of sounds caused by obstructions in their path increases. Since the head and upper torso act as an obstruction to incoming auditory signals, there is a significant difference in the level of the sound received at each ear at higher frequencies. In a similar manner to the case of the ITD, the auditory processing area of the brain can use the level differences and learned values to determine the source direction. Also, as previously stated for the case of the ITD, for sounds that originate near the front or back, the difference will be close to zero. This fact can sometimes lead to a front-back confusion in the sound localization process, but spectral cues from the pinna and concha can be used in this case as well as for vertical localization [94].

It is important to note that bilateral DHAs operating independently can impact the sound localization cues. Keidser et al. [36] summarized the potential impact of different DHA features:

- Independent WDRC processing in left and right hearing aids can upset the ILDs
- Unrestricted adaptive directionality in the left and right hearing aids can affect both the ITDs and ILDs
- Independent DNR strategies can distort the ILDs

There is evidence that independent adaptive directional systems can degrade sound localization performance by HI listeners. Van Den Bogaert et al. [95] conducted sound localization experiments with 10 HI listeners wearing bilateral DHAs operating independently in omnidirectional or adaptive directional modes. Results showed a significant decrement in sound localization performance when the DHAs are in independent adaptive directional mode.

To summarize, multiband adaptive directionality is a staple feature in higher end DHAs. Currently, there are no systems available to measure multiband adaptive directionality performance. Moreover, a measurement system that can additionally provide information on the effect of DHA processing on sound localization cues is also desirable. This chapter describes a proof-of-concept system developed for these purposes, and provides preliminary evaluation data.

## 5.2 Turntable-based System Development

In order to create polar plots, and thereby study the directional impact on DHA output signals caused by DHA signal processing algorithms, a new measurement system was built based on the approach originally proposed in [89]. A picture of the apparatus for free-field recordings as setup in a sound-treated chamber is shown in Figure 5-2. As can be seen, a turntable forms the base of the structure. The computer-controlled Brüel & Kjær Type 9640 was used as it provides continuous, relative, or absolute rotation. In addition, the turntable provides for programmable acceleration rate and flexible marking of the 0° azimuth. Mounted on top of the turntable was a stand holding a 2cc coupler for the DHA under test. In addition, three speaker holding systems were designed and attached to the turntable. The speakers were 5” spherical A’Diva speakers from Anthony Gallo Acoustics, and are identified by the yellow numbers in Figure 5-2. Beginning at the left side of the figure, speakers 1 and 2 were used to present the high level jammer signals. These speakers can be rotated to any angle between 90° and 270° (the rear half of the azimuth). This allowed for a high degree of freedom when designing both wideband and multiband measurements. Towards the right of the picture, speaker 3 is also attached to the turntable. The purpose of this speaker is to present a front jammer signal which is a new feature that was not included in the setup by Wu and Bentler [89]. The purpose of the front jammer will be explained later in this section. Finally, speaker 4 was mounted on a fixed stand and was used to present the probe signal. As the turntable rotated, it remained at the same location, which resulted in a change of the angle of arrival of the probe signal at the hearing aid. The vertical offset that can be seen in the image between speaker 4 and speaker 3 was necessary so that the jammer speakers do not block the probe signal. In addition to the free-field setup shown in Figure 5-2, a HATS setup was also used as shown in Figure 5-3. The HATS includes couplers for each ear that connect to Behind-the-Ear (BTE) DHAs. This particular setup was utilized to measure in-situ polar plots as well as the sound localization data.

Custom software was developed to control the turntable, the playback of multiple jammer signals, the sound level of different jammer signals, and the ensuing DHA recordings. The turntable rotation is controlled by a personal computer, which is also used for playback and recording. Some of the typical parameters used for experimental data collection include:

- Typical interval sizes (angular difference between adjacent recordings) of 10°, 6° and 4°.
- Playback of the probe and jammer signals at each recording angle for a total of 35 seconds.
- Recording of the DHA responses to the final 10 seconds of the playback, with the first 25 seconds designated to allow the DHAs to fully adapt prior to recording.
- Signal levels of 75 dB SPL for the rear jammers, 65 dB SPL for the front jammer and 55 dB SPL for the probe.
- Bandwidth spanning 250 Hz to 8 kHz for the wideband jammer and probe signals

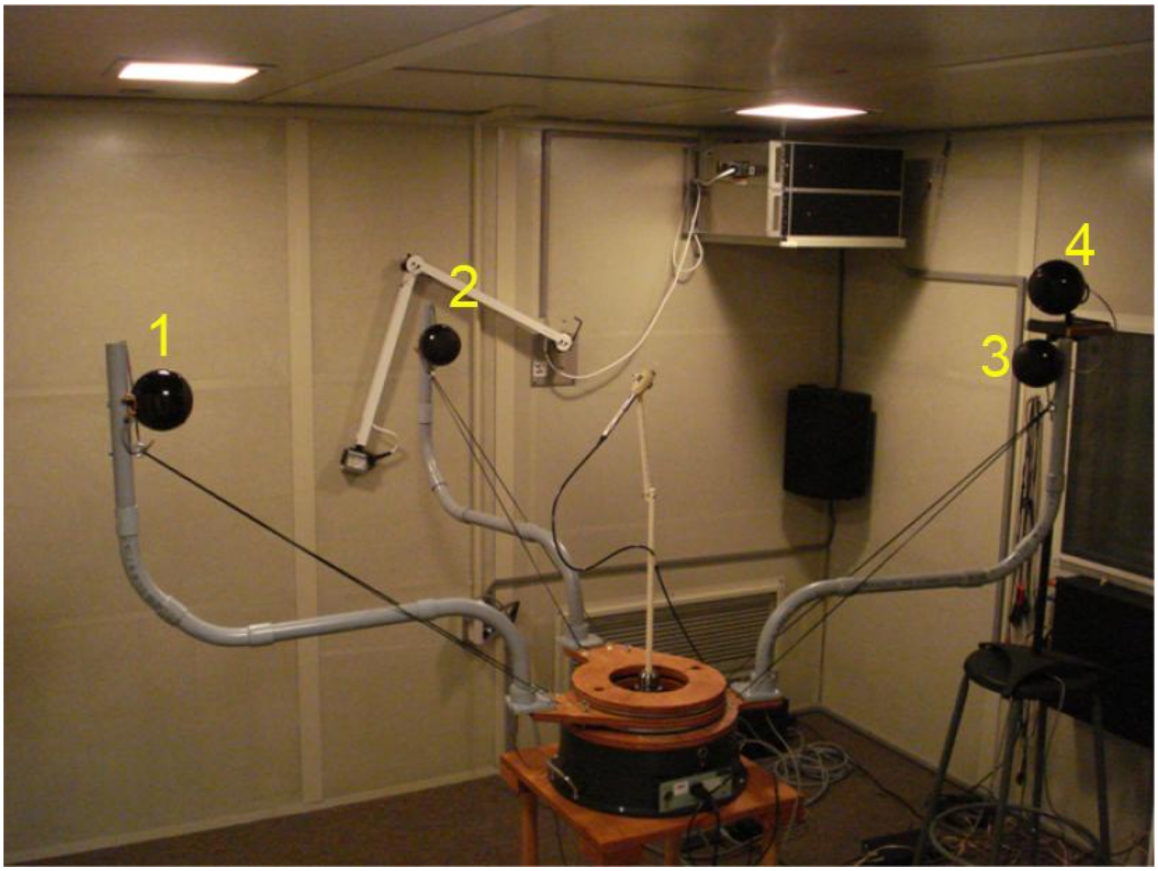



**Figure 5-2: The experiment setup for free-field recordings. Speakers 1 and 2 are used for the high level jammer signals, speaker 3 is used for the front jammer signal and speaker 4 is used for the probe signal.**

**Figure 5-3: The HATS turntable experiment setup.**

The justification for the 25 seconds adaptation time stems from Table 5-1, where it can be seen that the maximum adaptation time for any of the five DHAs tested was 18 seconds, while most adaptation times were under 10 seconds. Also, as previously mentioned, the front jammer was a new addition to the experiment setup described by Wu and Bentler [56]. This allowed for the testing of DHAs with WDRC in contrast to the experiments performed in [56] which were limited to DHAs programmed in a linear gain mode. Since the DHA uses the level of a signal originating from the front as an input to the WDRC algorithm, the constant level front jammer causes the gain set by the WDRC to remain fixed across all recordings. In addition, the front speaker also allows for the presentation of speech stimulus, with different noises played from the other two speakers in the rear azimuth. Such a setup will facilitate recording of DHA-processed speech stimuli and the application of speech intelligibility and speech quality indices discussed in the previous chapters.

## 5.3 Techniques to Measure Directionality

Two techniques to measure DHA directionality have been investigated for this study. The first was proposed in [89] and involves a signal cancellation approach which was also utilized in evaluating the DNR performance in Chapter 4. The second involves the combination of orthogonal signals to form a composite signal with similar characteristics to white noise.

### 5.3.1 Signal Cancellation Technique

As discussed in more detail in Chapter 4, the signal cancellation technique relies on two sets of recordings. For the measurement of adaptive directionality, the first recording is made with simultaneous presentation of the probe and jammers, while the second one is obtained with the probe and the inverted versions of all the jammers. By temporally aligning these two recordings and summing them together, the response of the DHAs to the probe is extracted. By repeating this procedure at multiple angular positions of the turntable, the complete polar plot can be constructed.

### 5.3.2 Orthogonal Signals Method

As described in [96], this method involves the synthesis of a complex sound constituting sinusoidal signals of varying frequency and random phase. The principal idea is that if the signals that need to be separated do not overlap in the frequency domain, then they can be easily separated from each other at the output of the system under test. If it is desirable to have a signal that is similar to white noise, sinusoids can be assigned to each source signal in an alternating manner such that each signal is wide band. To understand this method further, it is useful to consider a simple example as follows:

1. The bandwidth of interest is 60 Hz
2. The desired frequency separation is 10 Hz
3. One jammer and one probe signal are required

With these requirements, the jammer, probe and recorded signal spectra are shown in Figure 5-4. Extraction of the probe from the recorded signal is accomplished with relative

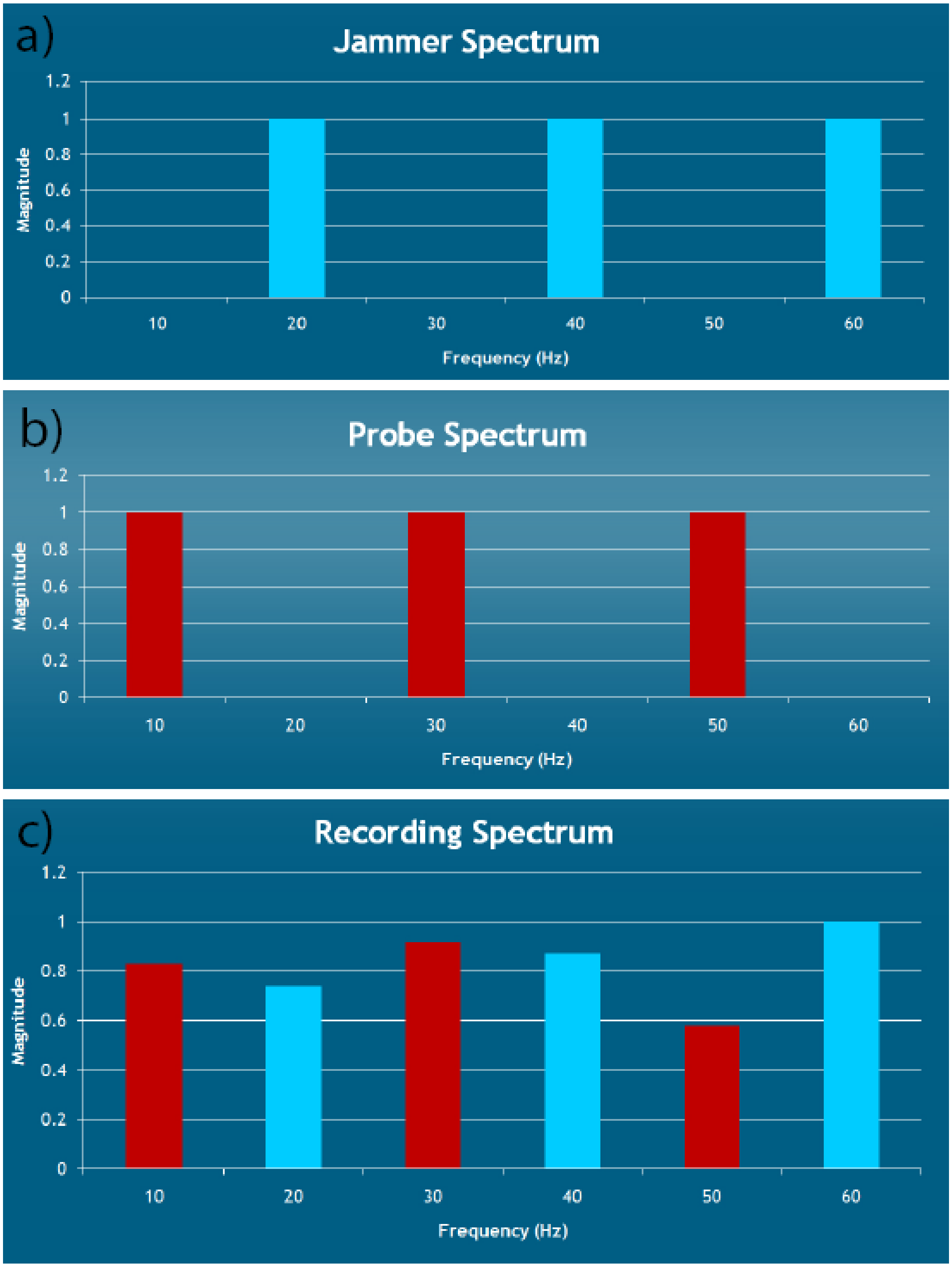



**Figure 5-4: Example orthogonal signal spectra for a) the jammer signal, b) the probe signal and c) the recorded signal.**

ease by considering only the frequency bins corresponding to the probe which are shown in red.

## 5.4 HATS Measurements

In-situ adaptive directionality measurements with the HATS followed the procedures described in the previous section. Bilateral DHAs were placed on the HATS and coupled to the built-in microphones in the left and right ears through an ear mold simulator and a sound tube. For the HATS measurements, both the left and right outputs are recorded, not only to generate bilateral polar plots, but also to compute the sound localization parameters. As mentioned earlier in this chapter, many of the DHA features impact ILDs more so than the ITDs. As such, the frequency-specific ILD data was estimated using the left and right DHA spectra.

## 5.5 Results

This section will present a number of results in the form of polar plots obtained from DHAs listed in Table 5-2. Due to space limitations, it is not possible to include all polar plots generated from this study, but those chosen for inclusion form an adequate representation of the various conditions and techniques that were employed.

**Table 5-2: Hearing aids tested.**

| Hearing Aid Model | Directivity | Binaural Wireless Communication |
|---|---|---|
| Siemens Motion | Adaptive | Yes |
| Oticon Epoq | Adaptive | Yes |
| Unitron Yuu | Adaptive | No |
| Starkey Destiny | Fixed | No |

### 5.5.1 Free Field Recordings

Figure 5-5a and b show the polar plots for the Siemens DHA when a single noise jammer is presented from 240° and 180° respectively. In this figure, the adaptive nature of the Siemens directivity feature is clearly demonstrated by the fact that in a) the polar pattern is hypercardioid and in b) the polar pattern is cardioid. In this example the orthogonal signals technique was used to extract the probe from the recordings. For comparison purposes, Figure 5-6 shows the 240° noise condition where the signal cancellation technique was used to extract the probe signal. The high degree of similarity between these results serves to validate the accuracy of the results obtained through the use of both the orthogonal and signal cancellation techniques. For the remainder of the section, the presented plots were generated using only the orthogonal signals technique to avoid repetition.

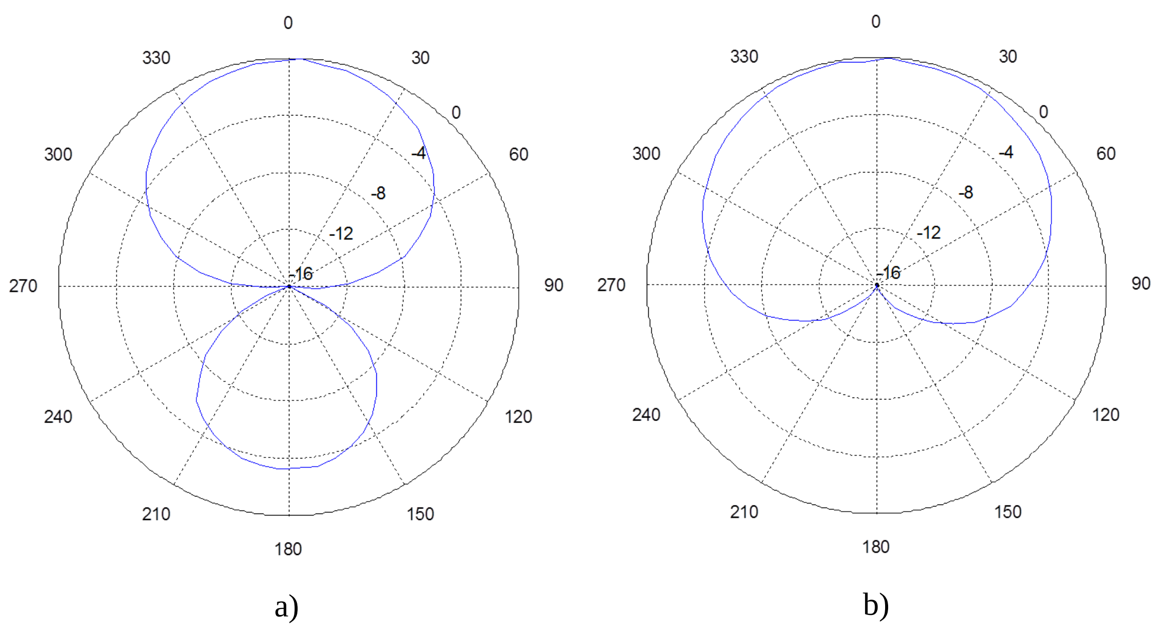



**Figure 5-5: The free field polar plots obtained from the Siemens Motion using the orthogonal signals method for a) the jammer at 240° and b) the jammer at 180°.**

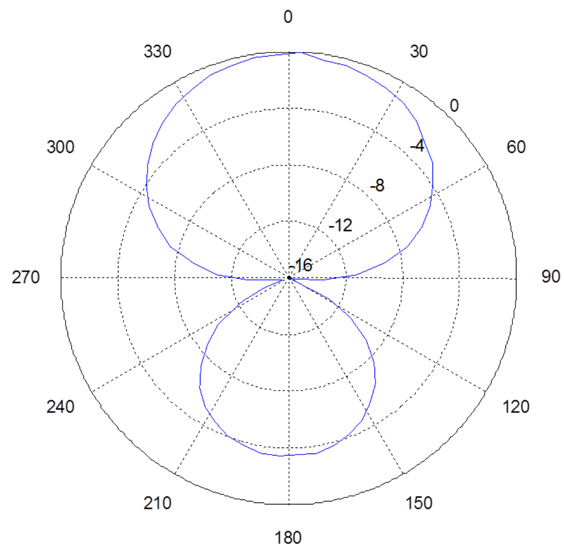
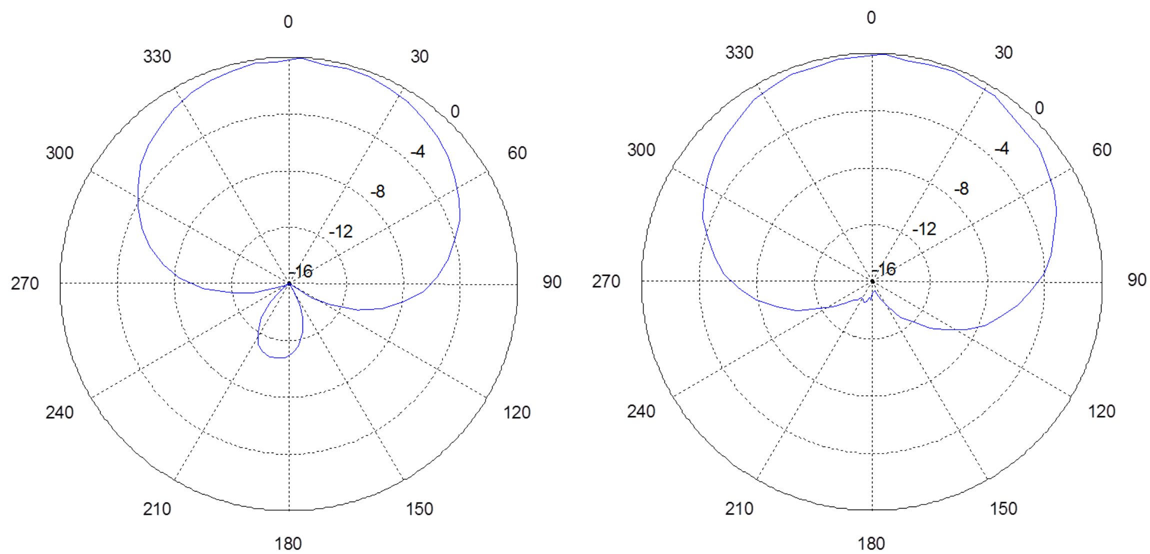



**Figure 5-6: The free field polar plots obtained from the Siemens Motion using the signal cancellation method for the jammer at 240°.**




a) b)

**Figure 5-7: The free field polar plots obtained from the Oticon Epoq using the orthogonal signals method for a) the jammer at 240° and b) the jammer at 180°.**

Figure 5-7 shows the Oticon polar plot for the condition where a single noise jammer is presented from 240° in a) and 180° in b). Once again, it is clear that the directivity is adaptive as the polar plot changes based on the source angle of the noise. It is interesting to note that the rear lobe of the plot in a) is much smaller than the rear lobe for the Siemens aid in the same condition.

Figure 5-8 shows the Starkey polar plot for the condition where a single noise jammer is presented from 240° in a) and 180° in b). Starkey DHAs employs acoustic directional microphones which are designed to provide hypercardioid spatial response. As such, the polar plots look almost identical for both noise conditions.

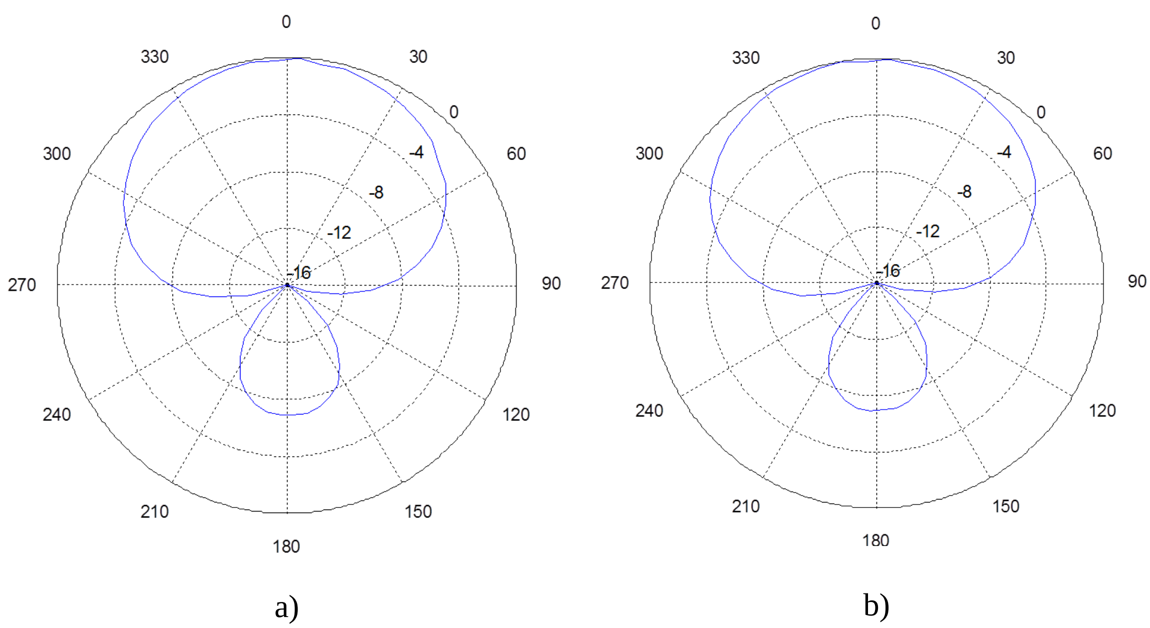
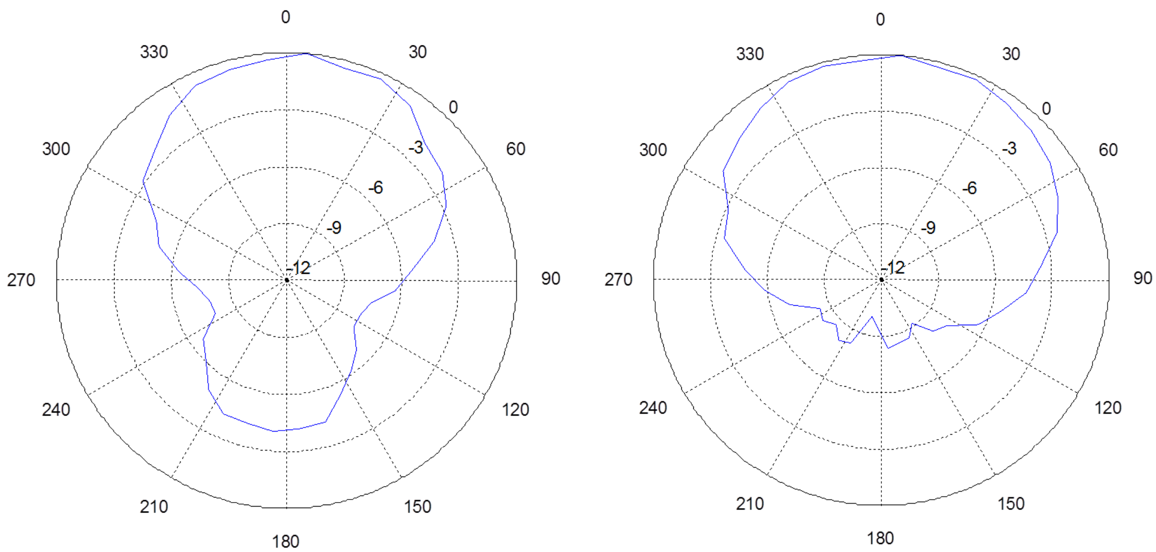



**Figure 5-8: The free field polar plots obtained from the Starkey Destiny using the orthogonal signals method for a) the jammer at 240° and b) the jammer at 180°.**

Figure 5-9 shows the results of a multi-band recording for the Unitron Yuu. For this recording a jammer signal confined to the 2 kHz octave band was presented from 240° and at the same time a second jammer signal confined to the 4 kHz octave band was presented from 180°. For the plot in a), the probe signal was filtered to include only the 2 kHz octave band and similarly the plot in b) includes only the 4 kHz octave band. As can be seen, the polar plots are similar to the expected forms, but are not as smooth or symmetric as the wide band plots. One of the likely causes of this reduction in quality is the fact that the band cutoff frequencies that the hearing aid uses for defining the boundaries of multiband adaptive directionality are proprietary information and are therefore unknown. By arbitrarily choosing the octave band cutoff frequencies for these plots, it is likely that the results are an average across a number of bands internal to the hearing aid. Nevertheless, the results shown are clear enough to demonstrate the assessment of multiband adaptive directionality.

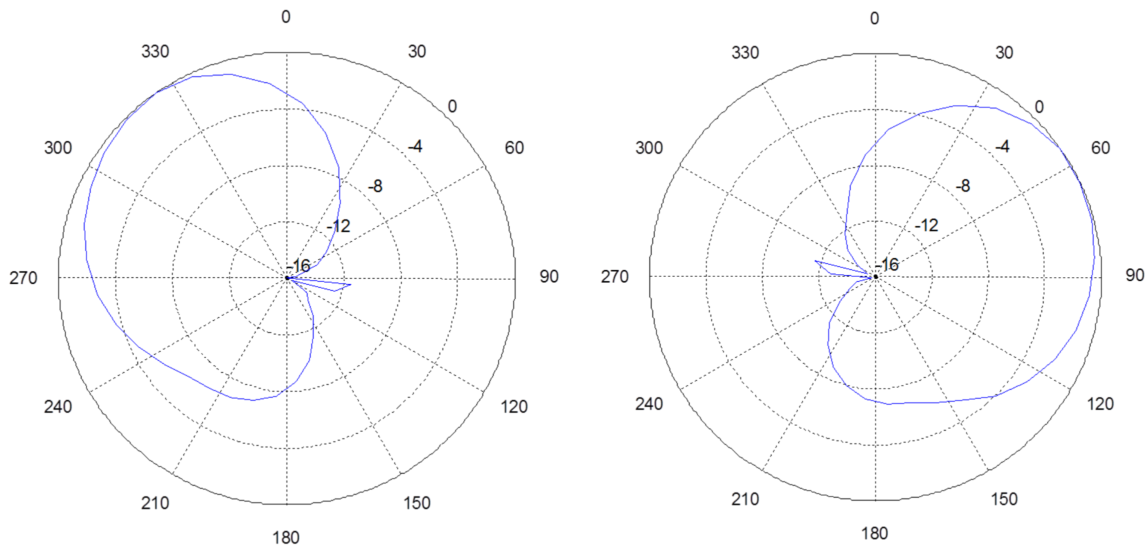



a) b)

**Figure 5-9: The free field multi-band polar plots obtained from the Unitron Yuu using the orthogonal signals method for a) the jammer at 240° and confined to the 2 kHz octave band and b) the jammer at 180° and confined to the 4 kHz octave band.**

### 5.5.2 Head and Torso Simulator Recordings

As previously described, two channel recordings were made with bilateral DHAs fitted to the HATS. This section will present some of the results from this experiment. Figure 5-10 shows the left and right ear recordings for the HATS without any attached DHAs (“open ear” response). This is included for reference purposes as all subsequent polar plots will include this head and torso shadowing effect.

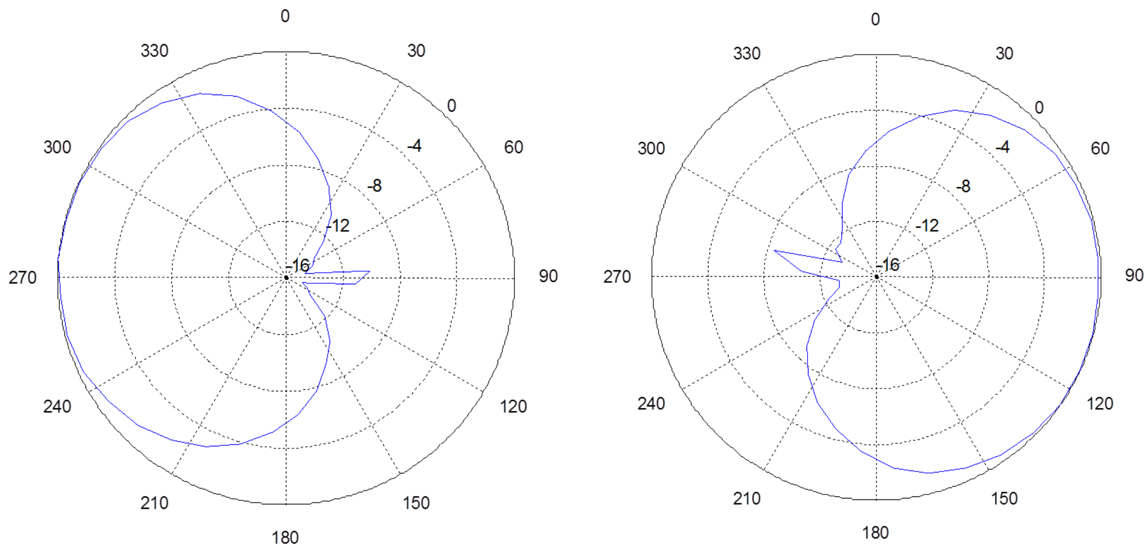



a) b)

**Figure 5-10: The HATS only polar plots for the a) left ear and b) right ear.**

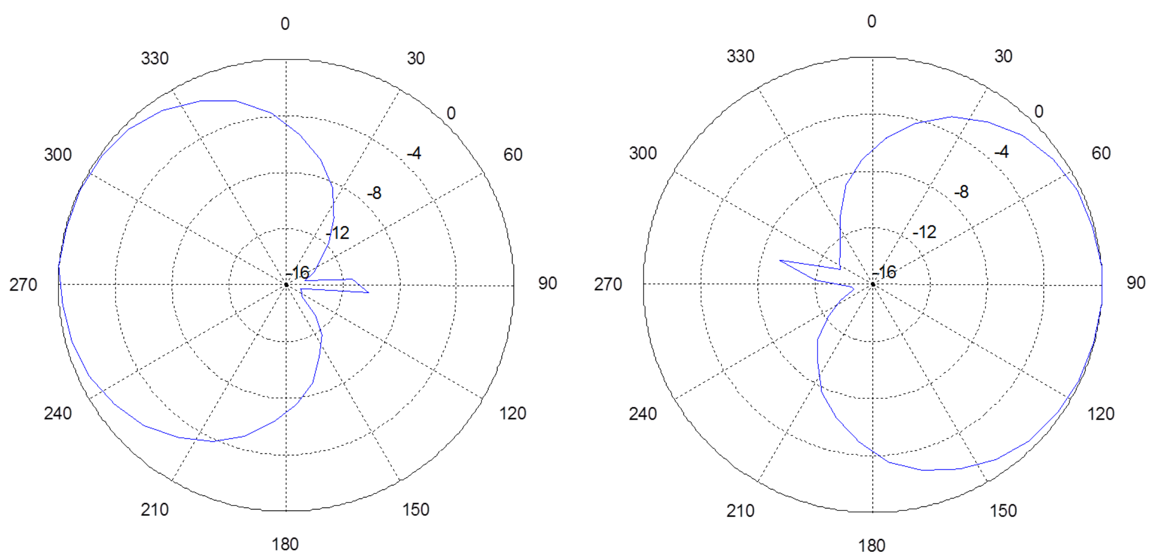
Figure 5-11 shows the polar plots for the left and right ears when the HATS was fitted with a pair of Siemens DHAs in the omnidirectional mode with the wireless communication off. Comparing this to the HATS only plots, it appears that some gain has been applied in the rear hemisphere.

Figure 5-12 shows the same condition as Figure 5-11 except that the wireless communication has been turned on. As can be seen, the plots are very similar.




a) b)

**Figure 5-11: The polar plots obtained with the HATS fitted with Siemens Motion hearing aids in the omni mode with wireless communication OFF for a) the left ear and b) the right ear.**




a) b)

**Figure 5-12: The polar plots obtained with the HATS fitted with Siemens Motion hearing aids in the omni mode with wireless communication ON for a) the left ear and b) the right ear.**

In Figure 5-13, the wireless communication is again off, the directivity has been changed to the adaptive mode and a jammer was presented from 240°. As can be seen in a), the left ear polar plot appears to be a combination of a hypercardiod pattern and the head and torso shadowing effect which is what would be expected. In contrast the right ear plot shown in b) appears to be a combination of a cardioid pattern and the head and torso shadowing effect. This may not be what is initially expected but after considering that the noise must bend around the head to reach the right hearing aid, it makes sense that the right hearing aid would adapt for a noise originating more from the rear rather than the side.

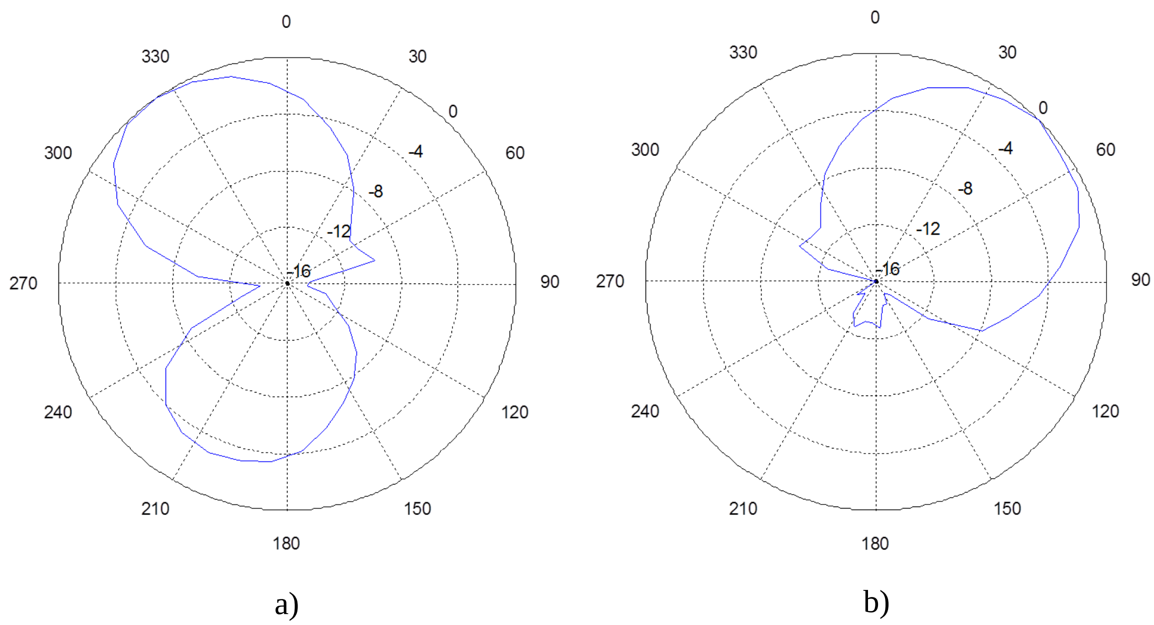
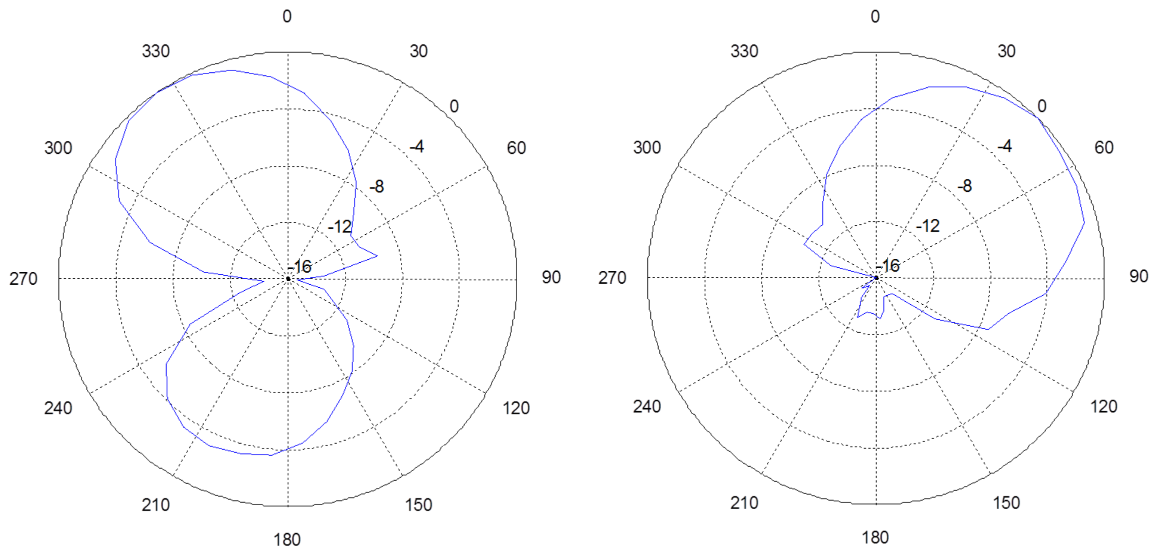



**Figure 5-13: The polar plots obtained with the HATS fitted with Siemens Motion hearing aids in the adaptive directional mode with wireless communication OFF for a) the left ear and b) the right ear. A noise jammer signal was presented from 240°.**

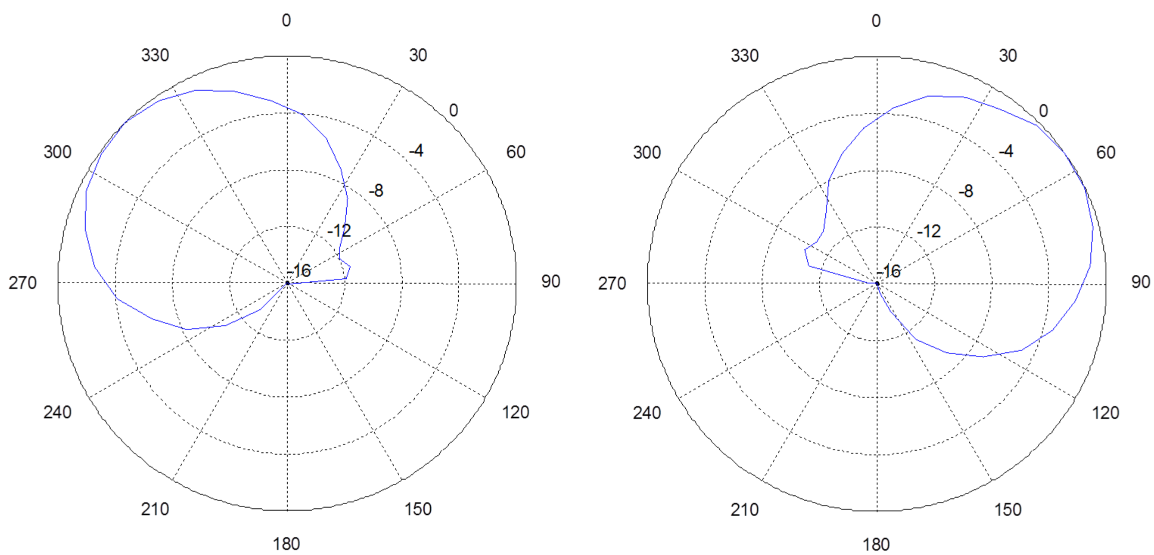
Figure 5-14 shows the same condition as Figure 5-13 except that the wireless communication has been turned on. No evident difference in polar plots was observed between the wireless activated and deactivated settings.

In Figure 5-15 the noise jammer has moved to 180°. As can be seen, the plots show a high degree of symmetry and are in line with the expected cardioid-like pattern.




a) b)

**Figure 5-14: The polar plots obtained with the HATS fitted with Siemens Motion hearing aids in the adaptive directional mode with wireless communication ON for a) the left ear and b) the right ear. A noise jammer signal was presented from 240°.**




a) b)

**Figure 5-15: The polar plots obtained with the HATS fitted with Siemens Motion hearing aids in the adaptive directional mode with wireless communication OFF for a) the left ear and b) the right ear. A noise jammer signal was presented from 180°.**

Figure 5-16 shows the same condition as Figure 5-15 for the Oticon Epoq hearing aids and the result is the expected cardioid-like pattern.

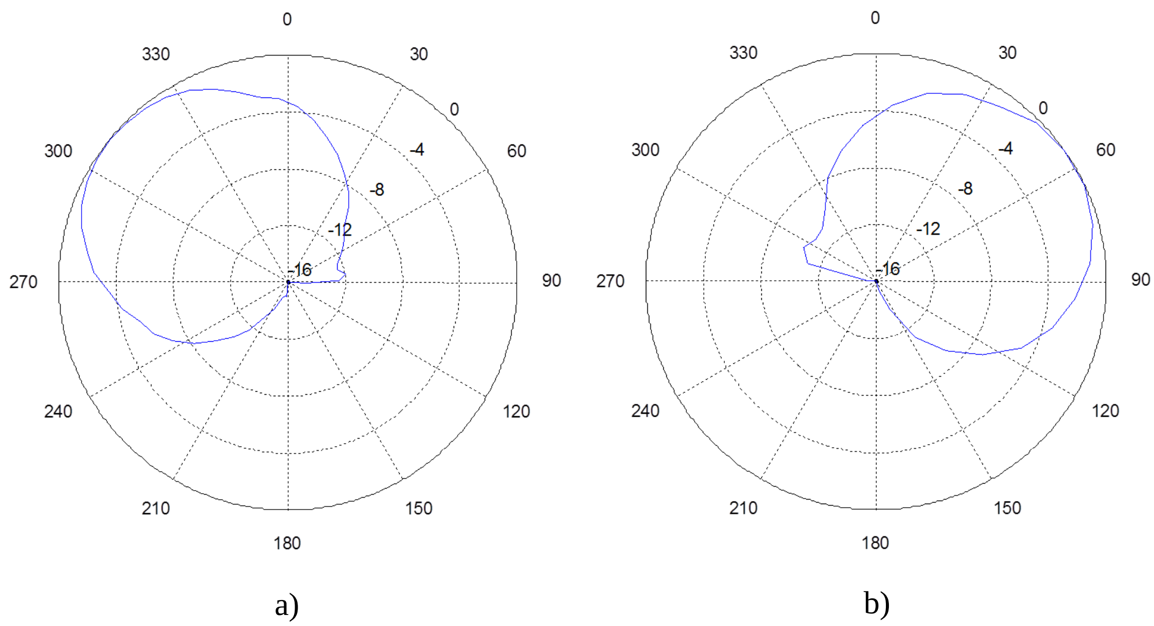
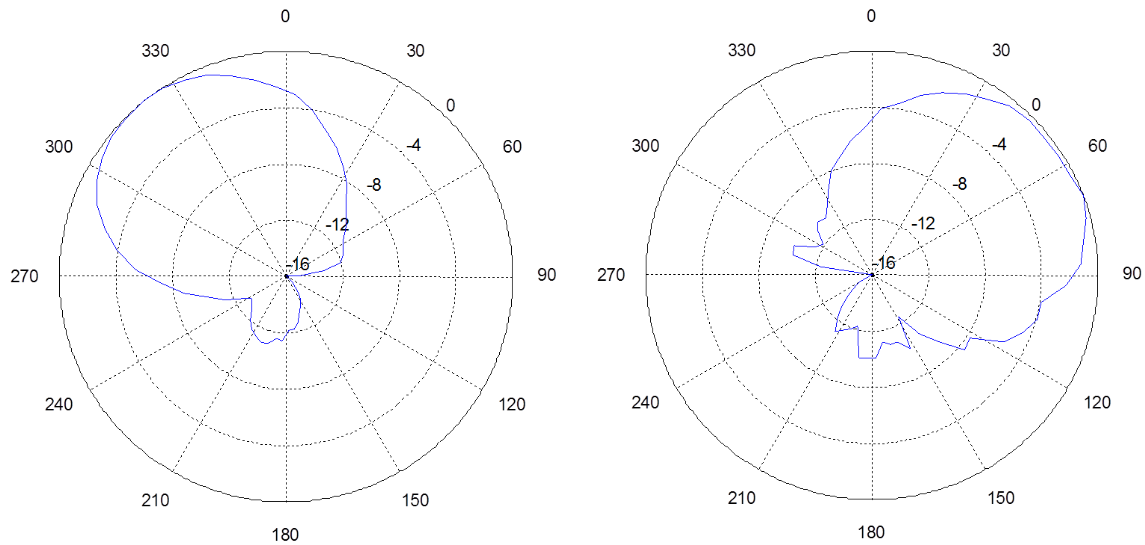



**Figure 5-16: The polar plots obtained with the HATS fitted with Oticon Epoq hearing aids in the adaptive directional mode with wireless communication OFF on a) the left ear and b) the right ear. A noise jammer signal was presented from 180°.**

Finally, Figure 5-17 shows the results with the Oticon Epoq when the noise is moved back to 240°. The left ear polar plot seems to show a pattern that is similar to the Oticon hypercardioid pattern obtained in the free field experiment. The right side polar plot appears to have an increased level of noise in the rear hemisphere which may be a result of some confusion from the multiple paths taken by the noise as it propagates around the head and torso to the right side hearing aid.




a) b)

**Figure 5-17: The polar plots obtained with the HATS fitted with Oticon Epoq hearing aids in the adaptive directional mode with wireless communication OFF on a) the left ear and b) the right ear. A noise jammer signal was presented from 240°.**

### 5.5.3 ILD Data

The frequency-specific ILDs were extracted for the bilateral pairs of Siemens Motion and Oticon Epoq DHAs, in four different settings: omnidirectional and wireless synchronization turned off, omnidirectional and wireless synchronization turned on, adaptive directional and wireless communication turned on, and adaptive directional and wireless communication turned on. Figure 5-18 and Figure 5-19 show the results emanating from this experiment, when there was a broadband noise source located at 240°.

Figure 5-18 displays the ILD data from bilateral Siemens DHAs at 500 Hz and 2000 Hz. The "open ear" ILD curves obtained with HATS wearing no DHAs is shown in these graphs for comparative purposes. The magnitude of ILDs is different between the two graphs – this is to be expected as ILDs are more prominent at high frequencies. The jammer at 240° does not affect the ILD data, when the DHAs are in omnidirectional mode. A significant change is apparent, when the DHAs switch into the adaptive directional mode. ILD distortion around the null (240°) is apparent at both 500 Hz and 2000 Hz.

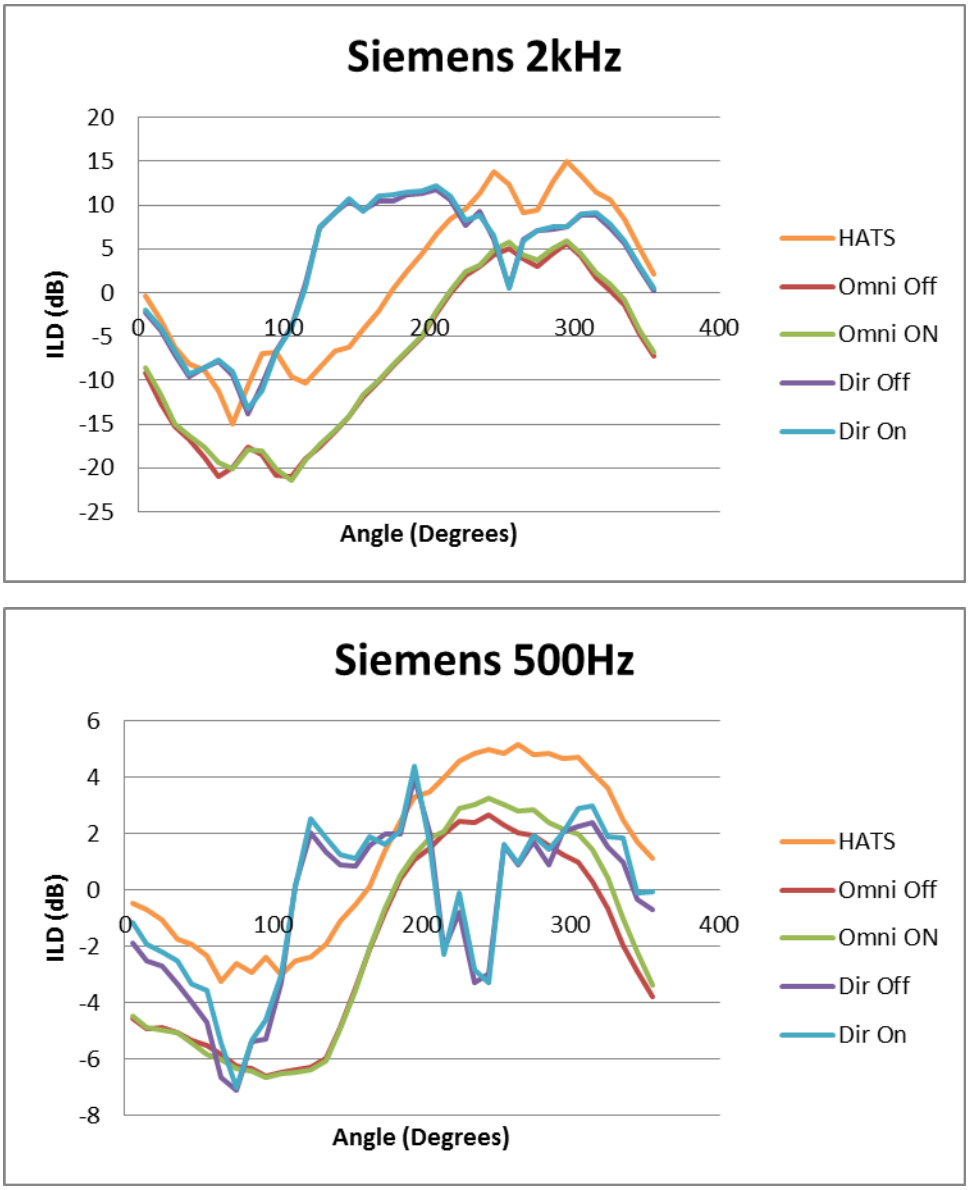



**Figure 5-18: Siemens ILDs. Here "Omni" and "Dir" refer to omnidirectional and adaptive directional modes, while "Off" and "On" refer to the state of wireless coordination.**

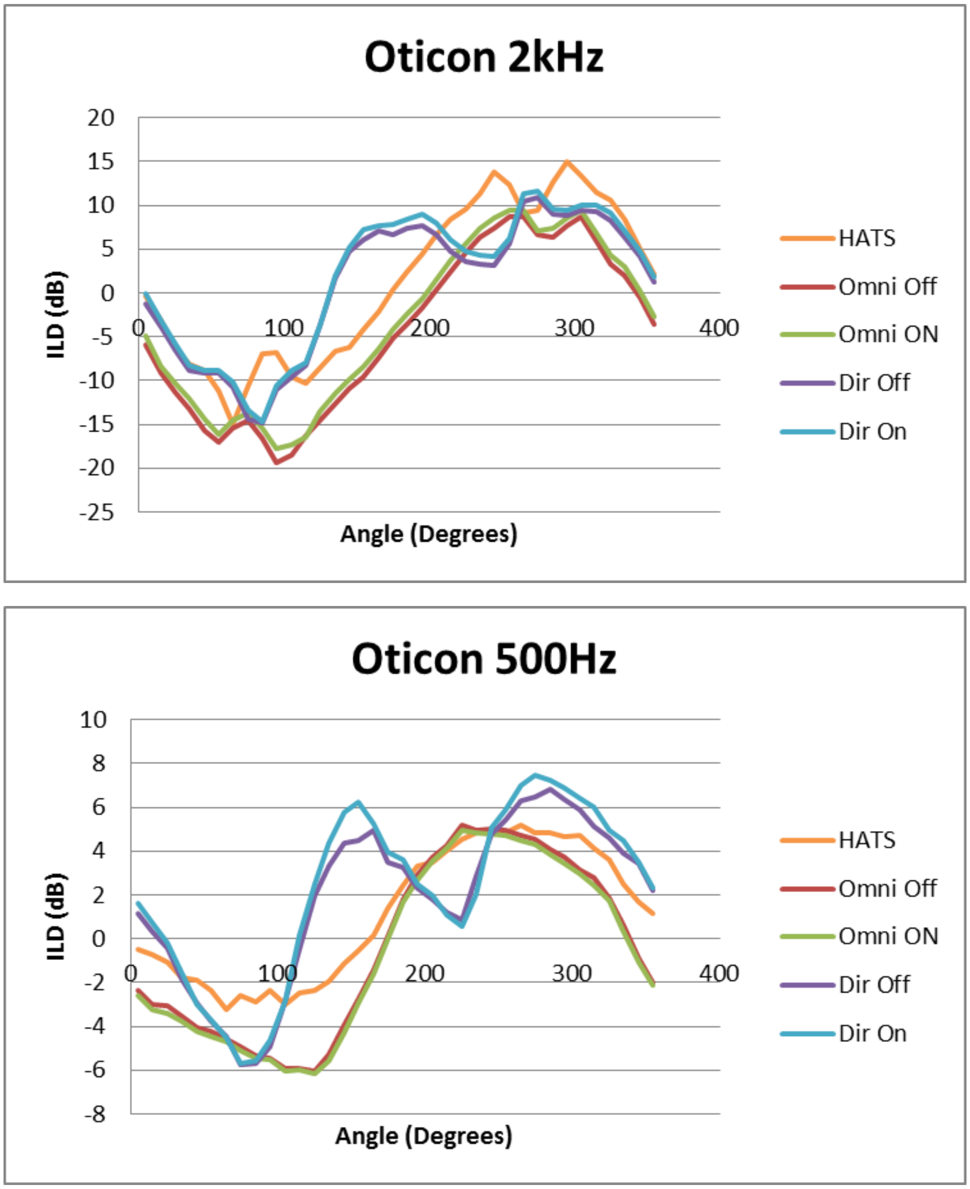



**Figure 5-19: Oticon ILDs. Here “Omni” and “Dir” refer to omnidirectional and adaptive directional modes, while “Off” and “On” refer to the state of wireless coordination.**

Similar results can be seen in Figure 5-19, which displays the data collected from bilateral Oticon DHAs. Another salient result from Figure 5-18 and Figure 5-19 is the similarity of ILD curves for wireless synchronization activation and deactivation states. No significant differences were observed, which is in line with the observations from Chapter 2.

## 5.6 Discussion and Conclusions

This chapter described a proof-of-concept flexible turntable-based electroacoustic measurement system that was designed and developed to test the functionality of bilateral DHAs. The system constitutes three independent loudspeakers that rotate with the turntable. One of these speakers is positioned at $0^{o}$ azimuth, while the remaining two are placed in the rear azimuths. There are two main advantages of this setup for assessing adaptive directionality: (a) the front speaker helps control the interaction between WDRC and directional processing features. As such, no special precautions need to be taken for programming the DHA; and (b) the two speakers in the rear can be positioned at different spatial locations, which aids in evaluating multiband adaptive directionality. Although not tested in this thesis, this setup also facilitates speech-in-noise DHA recordings through playback of speech stimuli from the front speaker and different types of noise stimuli from the rear speakers placed at different angles. Speech intelligibility and quality metrics described in Chapters 2 – 4 can then be applied for a comprehensive assessment of DHAs incorporating multiband directionality.

Two alternative methods of directionality assessment were investigated. The signal cancellation method utilizes two recordings: probe + jammers and probe + inverted jammers. The orthogonal method on the other hand utilizes a single recording, where the probe and jammer spectral components occupy different bins along the frequency axis. It was clear that the orthogonal and signal cancellation methods yielded very similar results. This was important to note as it serves to support the accuracy of both methods. The impact of the adaptive directivity feature employed by many modern DHAs was clearly demonstrated in this chapter. For both the free field and head and torso recordings it was clear that the hearing aids were adapting to the expected polar patterns to maximize the noise attenuation. For both of the hearing aids that include a bilateral wireless communication feature, there was no significant effect observed on the polar plots when the wireless link was changed from off to on. This was a surprise as it was expected that the wireless feature would impact the ILD to improve the sound localization ability of the user.

Currently, no standardized procedure exists for the evaluation of DHA directionality performance. It is expected that the work presented in this chapter could prove useful in the development of such standard.

# Chapter 6

# 6 Summary

This chapter will present an overview of the presented work with a particular focus on the key contributions and proposed future work.

## 6.1 Thesis Summary

Past studies have shown that the quality of hearing aid processed sound has a direct impact on the level of acceptance that a user has for a particular device. As the feature set of modern digital hearing aids continues to evolve, it becomes increasingly important to have access to methods of evaluating the effect that the advertised state-of-the-art features have on the speech quality produced by the device.

This thesis has focused on the performance evaluation of real hearing aids. A number of hearing aid features have been studied including bilateral wireless communication, digital noise reduction and adaptive directionality. A particular focus has been placed on the impact to speech quality imparted by hearing aid features.

In Chapter 2, the previously proposed Log-Likelihood Ratio (LLR) and Hearing Aid Sound Quality Index (HASQI) electroacoustic speech quality metrics are used to examine the effect that a variety of environmental conditions and hearing aid features have on the speech quality produced by two bilateral wireless hearing aid models. It is shown that the wireless feature did not produce a noticeable improvement in sound quality. In addition, a modified version of HASQI that is less computationally complex than the original metric is proposed and shown to produce estimates of speech quality that are more highly correlated with the subjective ratings.

In Chapter 3, a reference free electroacoustic speech quality algorithm, termed the Speech to Reverberation Modulation energy Ratio - Hearing Aid (SRMR-HA) is investigated. Commonly referred to as "non-intrusive", reference-free algorithms have been the focus of a variety of studies in the telecommunication field resulting in the publication of standard P.563 by the International Telecommunication Union (ITU) and the proposal of a metric that produces improved results, named ANIQUE+. The work presented in Chapter 3 proposes a non-intrusive speech quality algorithm designed specifically for hearing aid applications that accounts for the effects of hearing loss. The performance of the proposed metric is evaluated through the use of two subjective ratings databases and compared to the performance of the intrusive modified HASQI metric proposed in Chapter 2. Advantages of non-intrusive algorithms are discussed including the fact that it is not necessary to provide a time aligned, frequency shaped, clean reference signal as an input and the potential for use in real-time applications such as dynamically adjusting hearing aid parameters to maximize sound quality.

The focus of Chapter 4 is on the evaluation of hearing aid Digital Noise Reduction (DNR) performance. A novel approach to evaluate DNR performance is presented which includes a method to create recordings that allow for separation of the speech and noise portions of the signal. This allows for an analysis of the effect of DNR on the speech alone and on the speech-plus-noise signal and leads to accurate determination of the DNR attack time. Speech quality of the output signals is evaluated using the previously discussed modified HASQI and the SRMR-HA. In addition, speech intelligibility is measured using the well-known Speech Intelligibility Index (SII). Results are presented for seven state-of-the-art hearing aid models. By comparing the results, it is clear that the performance characteristics vary significantly between hearing aids which leads to the conclusion that the accurate performance characterization produced by the proposed approach offers relevant information when selecting a DHA model to fit the needs of particular users.

Chapter 5 presents a turn-table based method of evaluating digital hearing aid directionality performance. The approach taken was to lock the directionality pattern of the hearing aid and then capture recordings at a fixed azimuth interval over a complete 360°. In order to lock the directionality pattern, it was necessary to create recordings in such a way that a high level jammer signal and a lower level probe signal can be separated in the output recordings. For this study, two methods of separation were employed, one of which was similar to the separation technique used for the study presented in Chapter 4. The results showed strong agreement between the techniques which serves to support the accuracy of the approach. Once captured, the recordings were used to generate polar plots in order to visualize the performance of the directional noise cancellation. This was completed for four different hearing aid models, two of which included a bilateral wireless feature. A number of test conditions were utilized including varying the origin of the jammer signal, recording in the free-field with a single aid and with a bilaterally fitted head and torso simulator and enabling and disabling the bilateral wireless feature (when available). Based on this study, it is possible to conclude that the directionality pattern between hearing aids varies, that the adaptive directional hearing aids were successful in altering the polar plot based on the angle of noise arrival and that the wireless feature advertised by two of the hearing aids did not offer any adjustment to the level difference between the left and right side output signals.

## 6.2 Key Contributions

In addition to the thesis summary presented above, it is possible to highlight some key contributions that results from this work.

### 6.2.1 Chapter 2

- This study validated the HASQI speech quality metric by demonstrating the improved results obtained when compared to the more traditional LLR metric.

- It was shown that increased generality can be achieved through the use of a reduced, less computationally complex version of HASQI.

- It was shown that the bilateral wireless feature included with the two hearing aids under test did not offer any improvement to sound quality in both the subjective and electroacoustic portions of the study.

### 6.2.2 Chapter 3

- The development of a non-intrusive speech quality metric specifically for hearing aids applications was presented

- It was shown that the newly proposed metric performed quite well in predicting speech quality scores while eliminating the need for a time aligned, frequency shaped reference signal and allowing for real time speech quality prediction

### 6.2.3 Chapter 4

- Extensive DNR benchmarking that encompassed measures of not only speech intelligibility but also speech quality through the application of two electroacoustic metrics

- It was shown that DNR performance varied widely amongst state-of-the-art hearing aids based on unique methods to characterize their performance

### 6.2.4 Chapter 5

- An advanced bilateral hearing aid test system was presented

- It offers a number of novel features including bilateral testing with a head and torso simulator, the ability to determine the head-related transfer function with hearing aids and the ability to test multi-band adaptive directionality

## 6.3 Future Work

Based on the completed projects, a number of opportunities exist for future work:

- This study focused specifically on speech quality. In the future, the developed techniques should be extended to look at music and sound quality. Arehart et al. [58] investigated the performance of HASQI in predicting music quality, but this study was done using a simulated hearing aid. A study based on a real hearing aid that includes additional metrics such as SRMR-HA may yield results of interest.
- With respect to music sound quality, the HASQI and SRMR-HA neglect to consider the signal fine structure. An extension of these metrics to consider fine structure should be investigated, with the expectation of achieving improved music sound quality prediction.

- The reference-free SRMR-HA metric could be extended to account for additional speech quality features. One such example may be the addition of a neural model. A past study [97] has proposed neural models for speech intelligibility prediction that may serve as a useful starting point.
- This study has focused on three specific hearing aid features, namely DNR, directionality, and bilateral communication. The techniques developed could be extended to the study of additional hearing aid features such as feedback cancellation, Frequency Modulation (FM) systems, remote microphones, frequency compression etc.
- Looking back at the developed evaluation techniques, it seems possible to implement a system that could perform DNR, directionality and speech quality measurements in a single test environment. This would likely involve an apparatus similar to the turn-table setup presented in Chapter 5 combined with a speaker array similar to the setup presented in Chapters 2 and 3.
- In regards to DNR specifically, it should be possible to further refine the test technique by considering the use of a continuous, single signal that would allow for the measurement of attack time, release time and the effect of the DNR on the speech quality.
- Finally, it would be very interesting to investigate the possibility of fine tuning hearing aid DSP based on a non-intrusive speech quality estimation technique such as the SRMR-HA. It may be necessary to initially off-load some of the processing to a device external to the DHA and to investigate efficiency improvements to the algorithm, but if successful could lead to beneficial results.

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# Appendices

## Appendix A: DNR Study Additional Results

This appendix includes all of the results from the DNR study that were not presented in Chapter 4.

**Table A-1: S3 audiogram ratings where OFF and ON refer to the state of the DNR and the dB values listed for the noise conditions are the magnitude of the signal SNR.**

| Noise Condition | Agil | | Ambra | | Motion | | Velocity | | SiQ | | Passport | | M440-9 | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON |
| SII | | | | | | | | | | | | | | |
| Babble 0dB | 0.30 | 0.29 | 0.24 | 0.30 | 0.23 | 0.29 | 0.24 | 0.28 | 0.31 | 0.30 | 0.24 | 0.30 | 0.25 | 0.29 |
| Babble 5dB | 0.40 | 0.39 | 0.35 | 0.40 | 0.32 | 0.40 | 0.34 | 0.39 | 0.40 | 0.40 | 0.34 | 0.41 | 0.34 | 0.38 |
| SSN 0dB | 0.27 | 0.27 | 0.22 | 0.30 | 0.22 | 0.28 | 0.23 | 0.30 | 0.26 | 0.26 | 0.23 | 0.31 | 0.22 | 0.23 |
| SSN 5dB | 0.38 | 0.37 | 0.31 | 0.40 | 0.30 | 0.41 | 0.33 | 0.42 | 0.37 | 0.36 | 0.33 | 0.42 | 0.32 | 0.34 |
| Traffic 0dB | 0.27 | 0.26 | 0.22 | 0.29 | 0.21 | 0.27 | 0.22 | 0.27 | 0.30 | 0.28 | 0.22 | 0.30 | 0.23 | 0.23 |
| Traffic 5dB | 0.38 | 0.37 | 0.32 | 0.39 | 0.29 | 0.39 | 0.32 | 0.40 | 0.38 | 0.38 | 0.32 | 0.41 | 0.32 | 0.34 |
| SRMR | | | | | | | | | | | | | | |
| Babble 0dB | 1.12 | 1.11 | 0.86 | 1.09 | 0.78 | 1.44 | 0.75 | 1.20 | 1.61 | 1.60 | 0.85 | 1.56 | 0.82 | 0.81 |
| Babble 5dB | 1.56 | 1.56 | 1.38 | 1.61 | 1.19 | 2.03 | 1.25 | 1.92 | 1.87 | 1.84 | 1.37 | 2.14 | 1.21 | 1.20 |
| SSN 0dB | 0.61 | 0.59 | 0.42 | 0.71 | 0.47 | 0.81 | 0.43 | 1.19 | 0.61 | 0.61 | 0.57 | 1.44 | 0.48 | 0.46 |
| SSN 5dB | 1.10 | 1.08 | 0.85 | 1.41 | 0.89 | 2.04 | 0.88 | 2.24 | 1.19 | 1.17 | 1.11 | 2.16 | 0.88 | 0.91 |
| Traffic 0dB | 0.64 | 0.63 | 0.52 | 0.79 | 0.45 | 0.73 | 0.42 | 0.82 | 0.73 | 0.82 | 0.47 | 1.15 | 0.50 | 0.43 |
| Traffic 5dB | 1.09 | 1.08 | 0.96 | 1.43 | 0.79 | 1.56 | 0.82 | 1.68 | 1.51 | 1.37 | 0.93 | 1.87 | 0.85 | 0.76 |
| HASQI | | | | | | | | | | | | | | |
| Babble 0dB | 0.60 | 0.56 | 0.48 | 0.52 | 0.39 | 0.44 | 0.47 | 0.54 | 0.59 | 0.58 | 0.48 | 0.57 | 0.48 | 0.48 |
| Babble 5dB | 0.77 | 0.74 | 0.67 | 0.65 | 0.54 | 0.60 | 0.66 | 0.70 | 0.75 | 0.74 | 0.67 | 0.72 | 0.65 | 0.66 |
| SSN 0dB | 0.52 | 0.48 | 0.33 | 0.44 | 0.34 | 0.40 | 0.38 | 0.52 | 0.48 | 0.47 | 0.43 | 0.53 | 0.40 | 0.33 |
| SSN 5dB | 0.70 | 0.67 | 0.56 | 0.60 | 0.51 | 0.59 | 0.58 | 0.69 | 0.67 | 0.65 | 0.61 | 0.69 | 0.58 | 0.56 |
| Traffic 0dB | 0.52 | 0.49 | 0.39 | 0.44 | 0.31 | 0.38 | 0.39 | 0.50 | 0.51 | 0.51 | 0.42 | 0.53 | 0.40 | 0.32 |
| Traffic 5dB | 0.71 | 0.67 | 0.60 | 0.57 | 0.47 | 0.56 | 0.57 | 0.67 | 0.70 | 0.68 | 0.60 | 0.69 | 0.57 | 0.55 |

**Table A-2: N4 audiogram ratings where OFF and ON refer to the state of the DNR and the dB values listed for the noise conditions are the magnitude of the signal SNR.**

| Noise Condition | Agil | | Ambra | | Motion | | Velocity | | SiQ | | Passport | | M440-9 | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON | OFF | ON |
| SII | | | | | | | | | | | | | | |
| Babble 0dB | 0.28 | 0.28 | 0.23 | 0.27 | 0.28 | 0.32 | 0.25 | 0.29 | 0.28 | 0.29 | 0.24 | 0.30 | 0.26 | 0.31 |
| Babble 5dB | 0.39 | 0.38 | 0.33 | 0.38 | 0.37 | 0.43 | 0.35 | 0.41 | 0.40 | 0.39 | 0.34 | 0.40 | 0.36 | 0.40 |
| SSN 0dB | 0.25 | 0.26 | 0.19 | 0.27 | 0.24 | 0.28 | 0.23 | 0.30 | 0.23 | 0.25 | 0.21 | 0.30 | 0.22 | 0.23 |
| SSN 5dB | 0.36 | 0.36 | 0.29 | 0.37 | 0.33 | 0.41 | 0.33 | 0.43 | 0.34 | 0.35 | 0.31 | 0.40 | 0.32 | 0.35 |
| Traffic 0dB | 0.25 | 0.25 | 0.21 | 0.27 | 0.25 | 0.29 | 0.22 | 0.28 | 0.25 | 0.27 | 0.21 | 0.29 | 0.23 | 0.24 |
| Traffic 5dB | 0.36 | 0.36 | 0.31 | 0.37 | 0.34 | 0.41 | 0.33 | 0.41 | 0.36 | 0.37 | 0.31 | 0.39 | 0.33 | 0.36 |
| SRMR | | | | | | | | | | | | | | |
| Babble 0dB | 1.30 | 1.28 | 0.71 | 0.94 | 1.01 | 1.71 | 0.96 | 1.43 | 1.35 | 1.52 | 0.91 | 1.54 | 1.14 | 1.28 |
| Babble 5dB | 1.83 | 1.82 | 1.14 | 1.43 | 1.32 | 2.28 | 1.48 | 1.98 | 1.93 | 1.90 | 1.35 | 2.02 | 1.62 | 1.75 |
| SSN 0dB | 0.61 | 0.71 | 0.33 | 0.49 | 0.53 | 1.03 | 0.47 | 1.17 | 0.56 | 0.70 | 0.45 | 1.27 | 0.62 | 0.69 |
| SSN 5dB | 1.14 | 1.20 | 0.55 | 0.94 | 0.81 | 1.94 | 0.94 | 1.93 | 1.06 | 1.35 | 0.83 | 1.91 | 1.11 | 1.36 |
| Traffic 0dB | 0.74 | 0.74 | 0.42 | 0.63 | 0.50 | 0.89 | 0.55 | 1.07 | 0.75 | 0.90 | 0.51 | 1.10 | 0.70 | 0.67 |
| Traffic 5dB | 1.27 | 1.26 | 0.75 | 1.16 | 0.81 | 1.65 | 1.03 | 1.79 | 1.43 | 1.68 | 0.90 | 1.71 | 1.16 | 1.33 |
| HASQI | | | | | | | | | | | | | | |
| Babble 0dB | 0.58 | 0.58 | 0.45 | 0.56 | 0.62 | 0.65 | 0.51 | 0.61 | 0.60 | 0.59 | 0.43 | 0.54 | 0.50 | 0.52 |
| Babble 5dB | 0.75 | 0.75 | 0.65 | 0.72 | 0.80 | 0.80 | 0.69 | 0.77 | 0.77 | 0.75 | 0.62 | 0.70 | 0.67 | 0.72 |
| SSN 0dB | 0.50 | 0.53 | 0.33 | 0.50 | 0.55 | 0.61 | 0.40 | 0.57 | 0.46 | 0.47 | 0.36 | 0.52 | 0.41 | 0.37 |
| SSN 5dB | 0.69 | 0.69 | 0.52 | 0.63 | 0.73 | 0.78 | 0.59 | 0.75 | 0.66 | 0.66 | 0.52 | 0.67 | 0.59 | 0.60 |
| Traffic 0dB | 0.50 | 0.52 | 0.36 | 0.50 | 0.54 | 0.60 | 0.40 | 0.56 | 0.50 | 0.50 | 0.36 | 0.49 | 0.43 | 0.37 |
| Traffic 5dB | 0.68 | 0.68 | 0.56 | 0.66 | 0.71 | 0.77 | 0.59 | 0.73 | 0.69 | 0.68 | 0.53 | 0.65 | 0.58 | 0.59 |

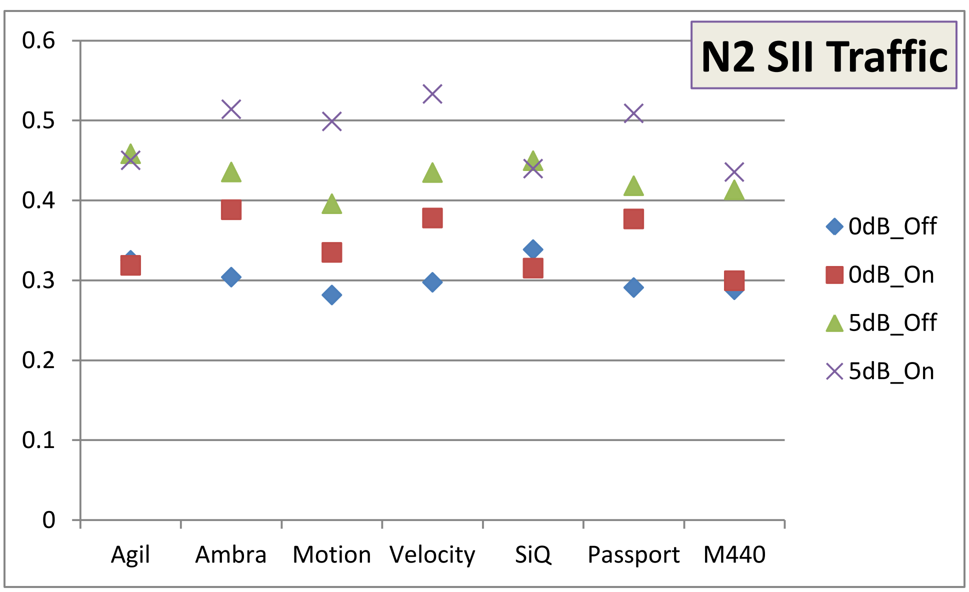


**Figure A-1: Comparison of SII values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

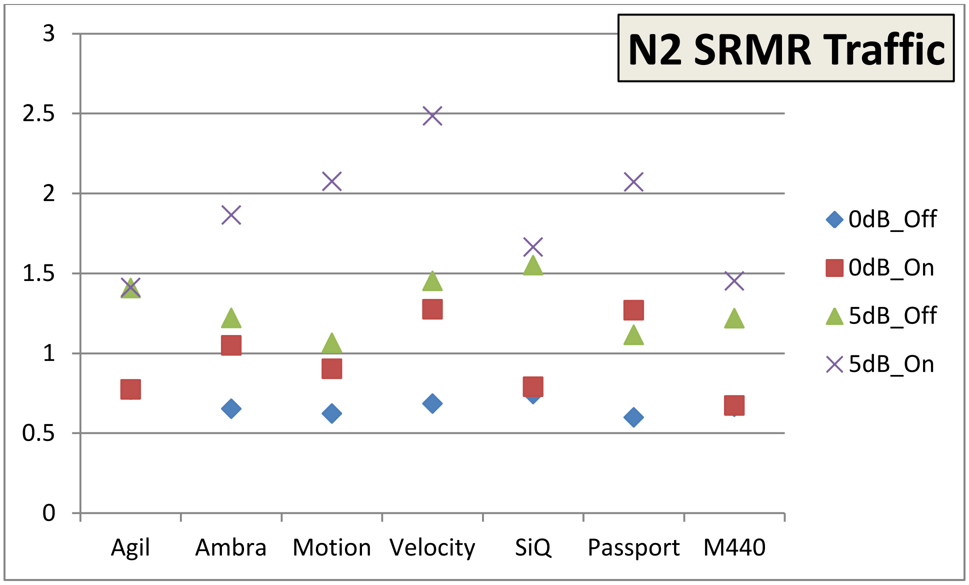


**Figure A-2: Comparison of SRMR values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

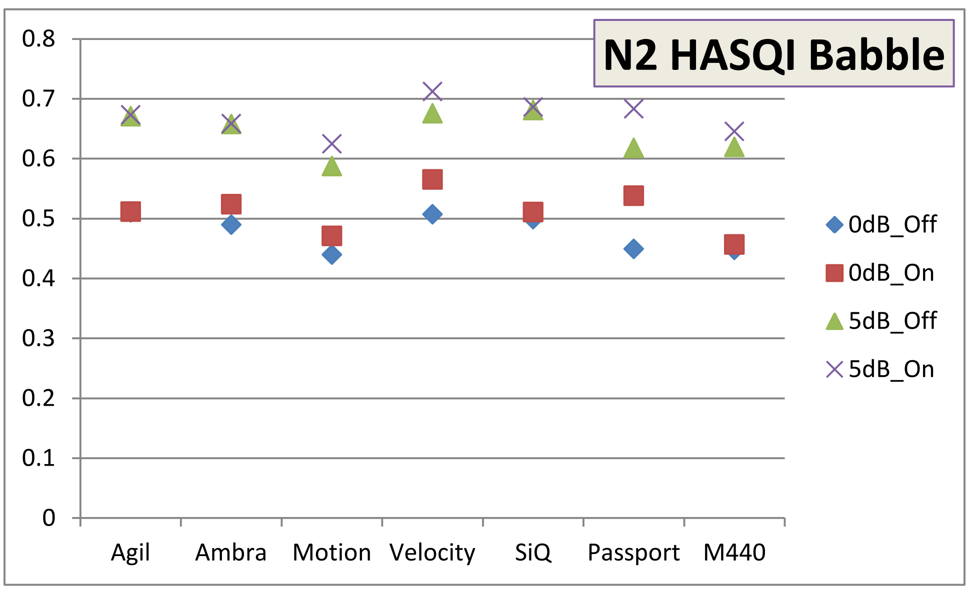


**Figure A-3: Comparison of HASQI values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

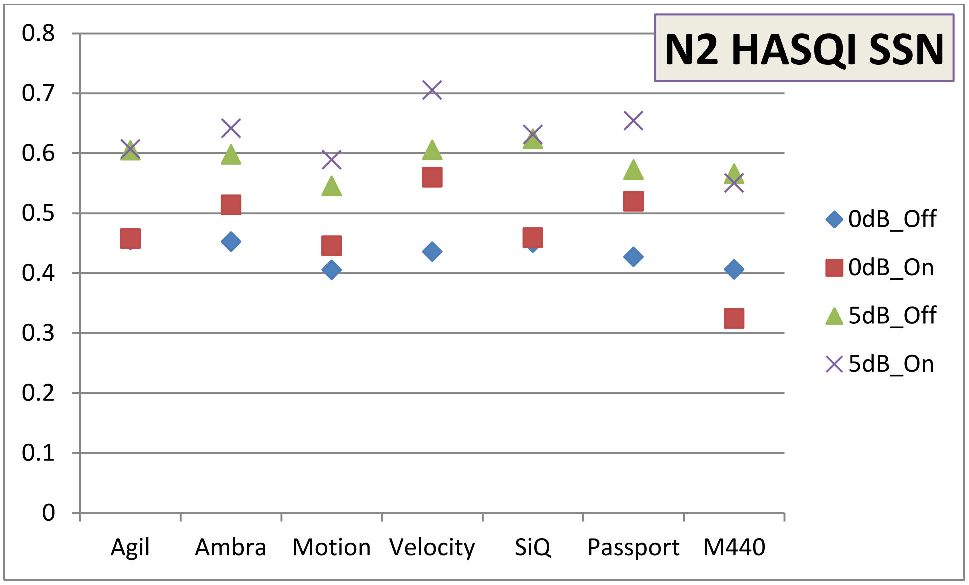


**Figure A-4: Comparison of HASQI values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

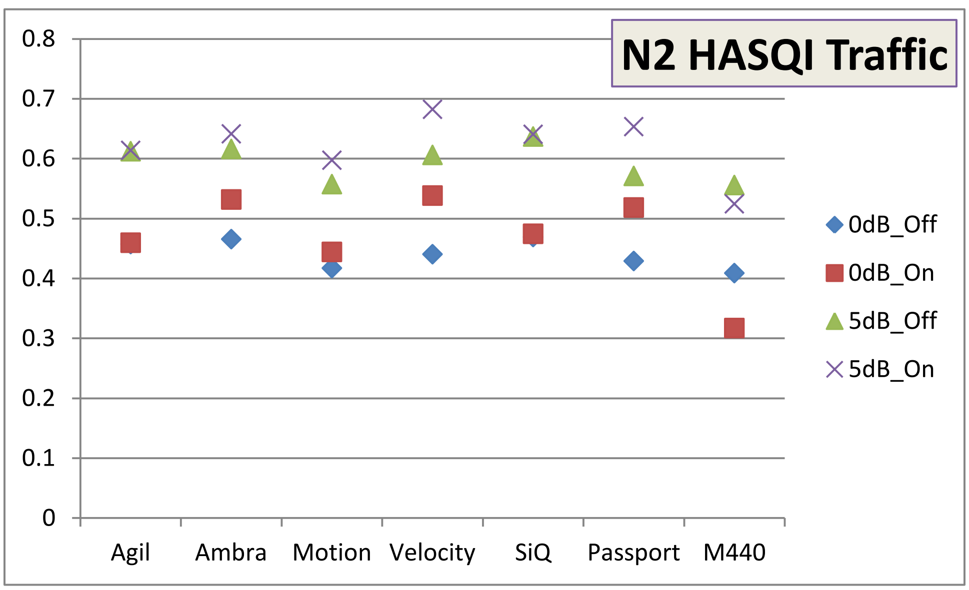


**Figure A-5: Comparison of HASQI values between the N2 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

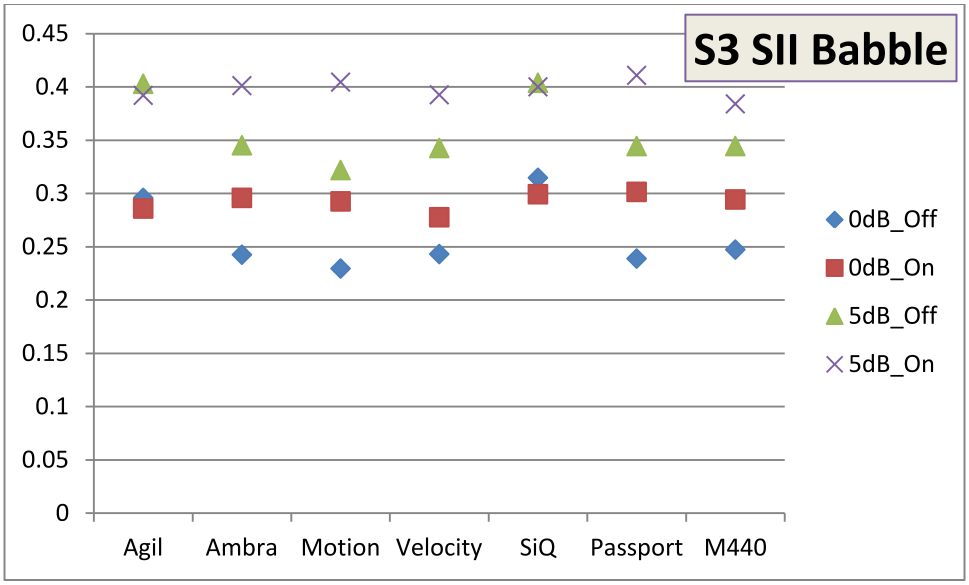


**Figure A-6: Comparison of SII values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

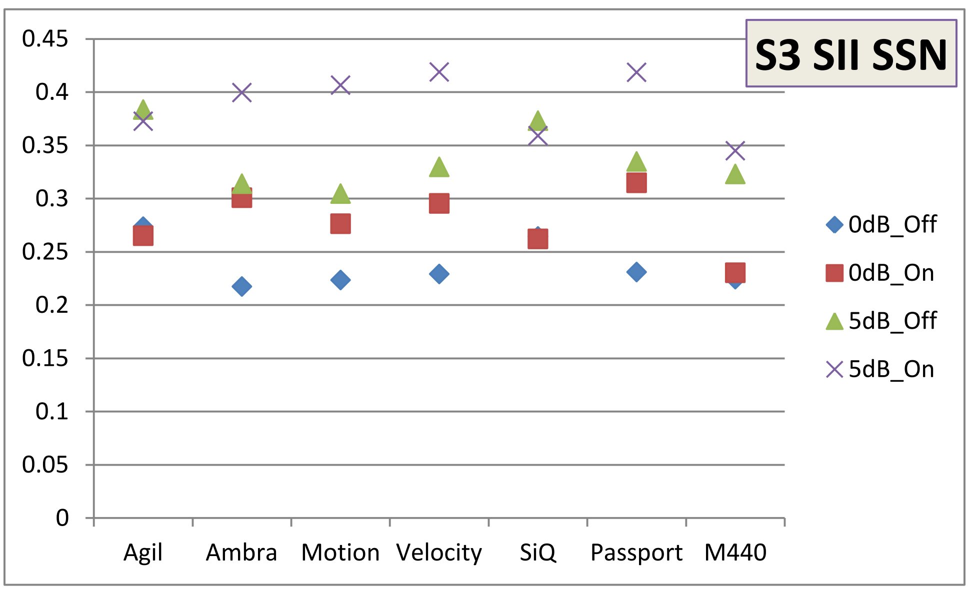


**Figure A-7: Comparison of SII values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

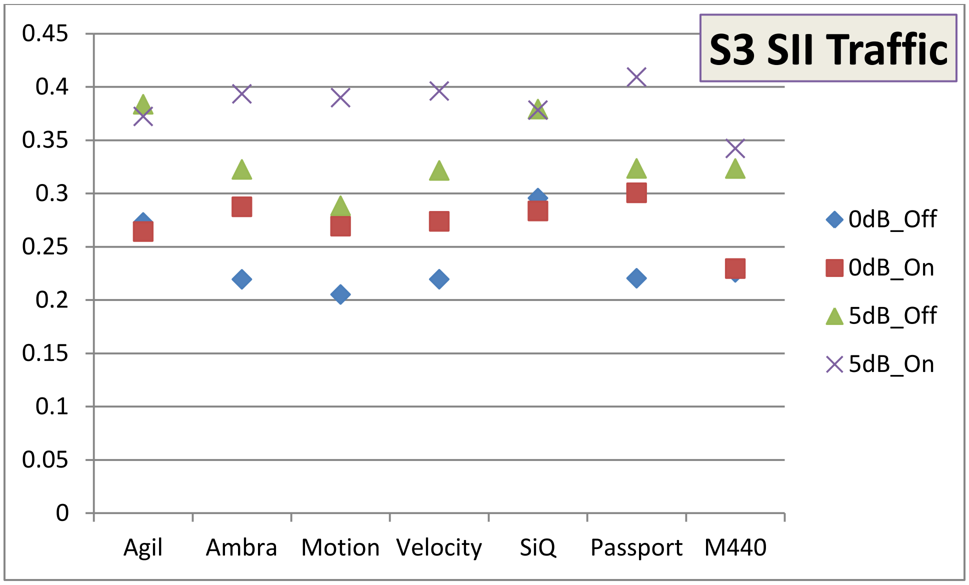


**Figure A-8: Comparison of SII values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

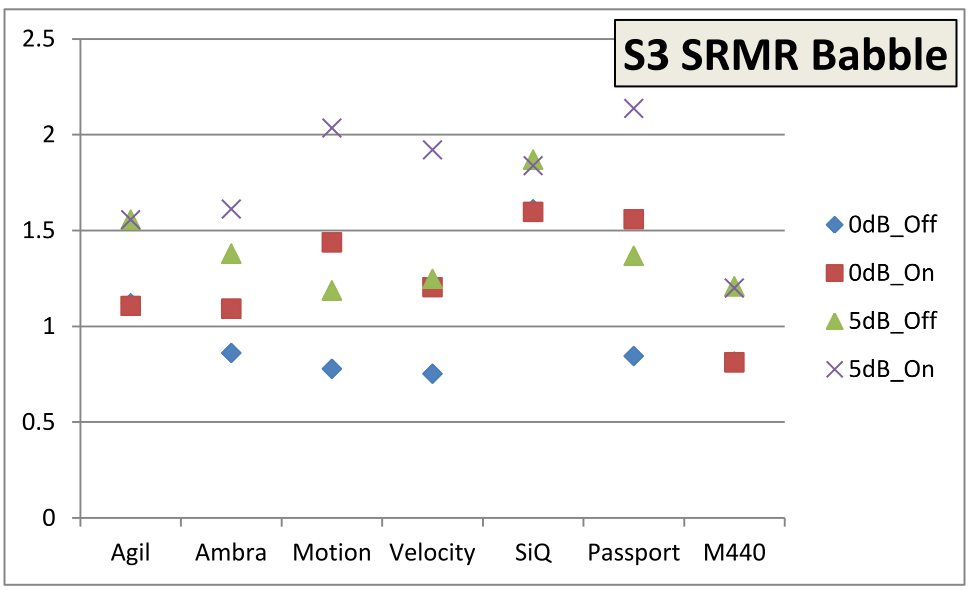


**Figure A-9: Comparison of SRMR values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

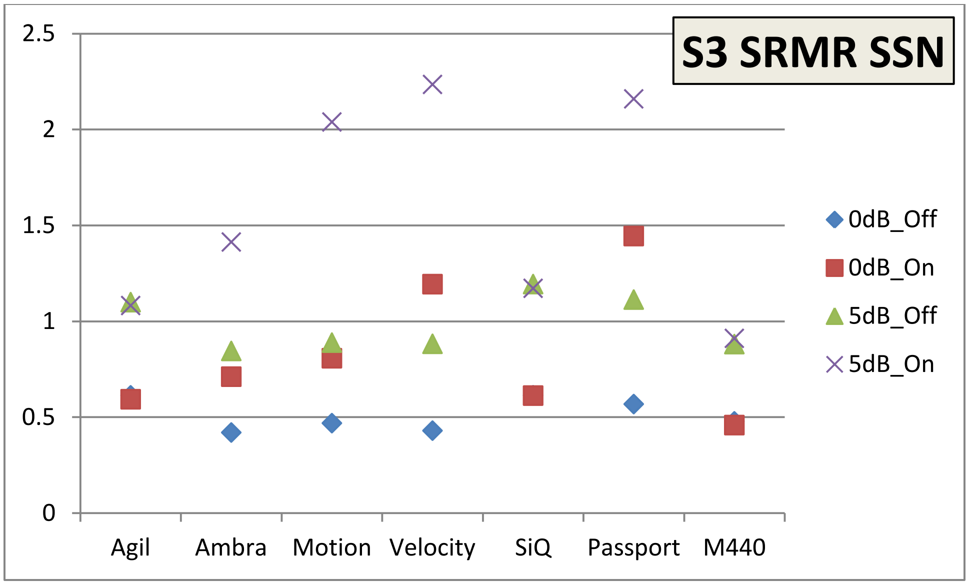


**Figure A-10: Comparison of SRMR values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

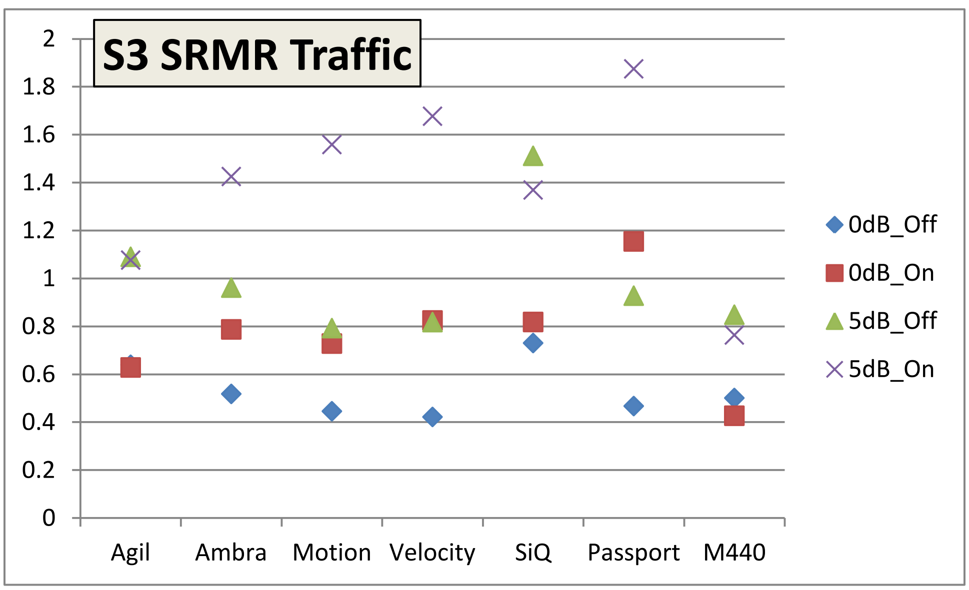


**Figure A-11: Comparison of SRMR values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

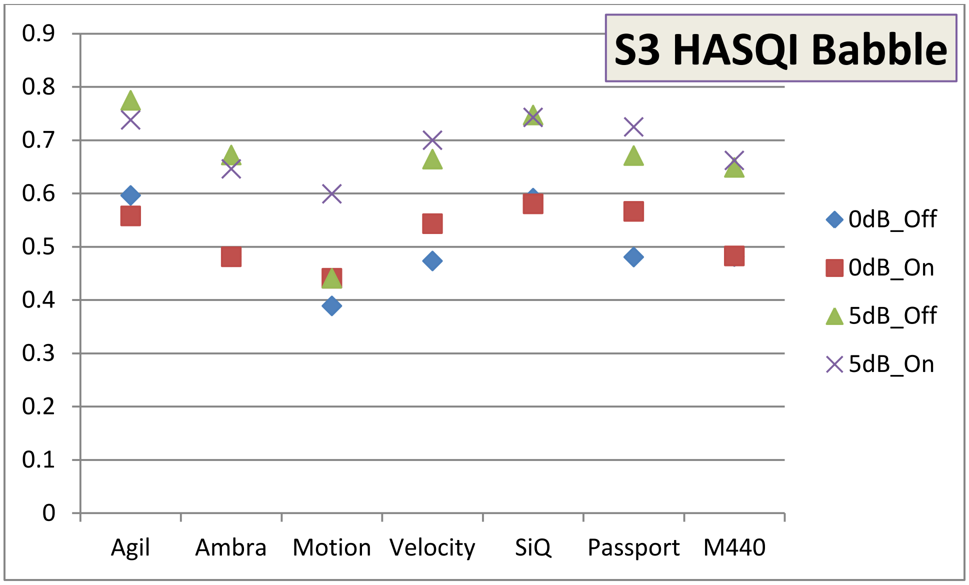


**Figure A-12: Comparison of HASQI values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

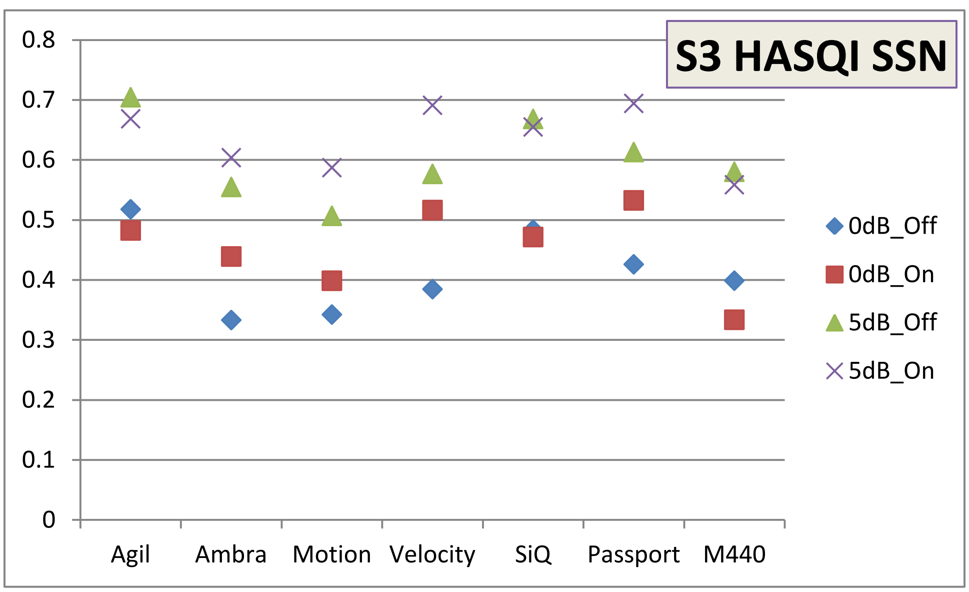


**Figure A-13: Comparison of HASQI values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

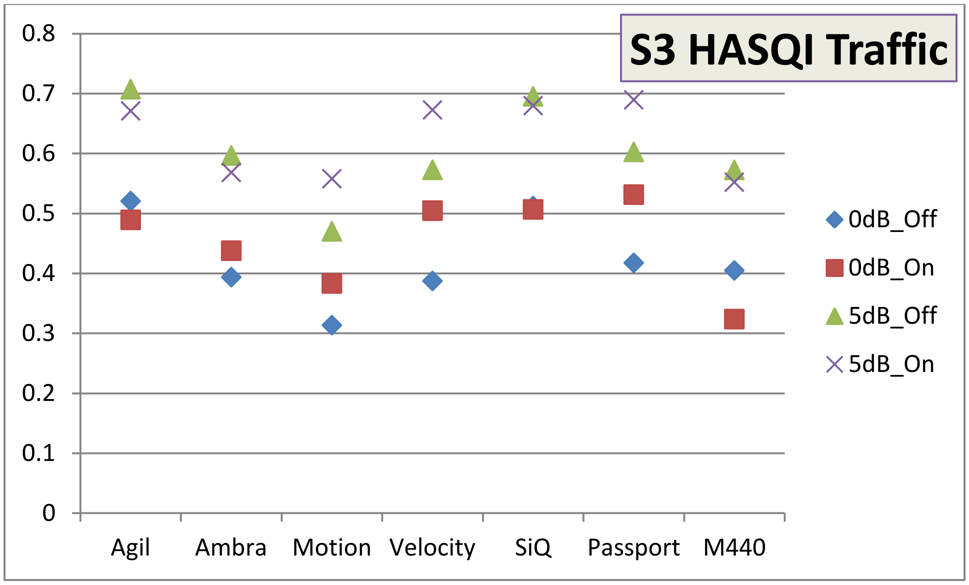


**Figure A-14: Comparison of HASQI values between the S3 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

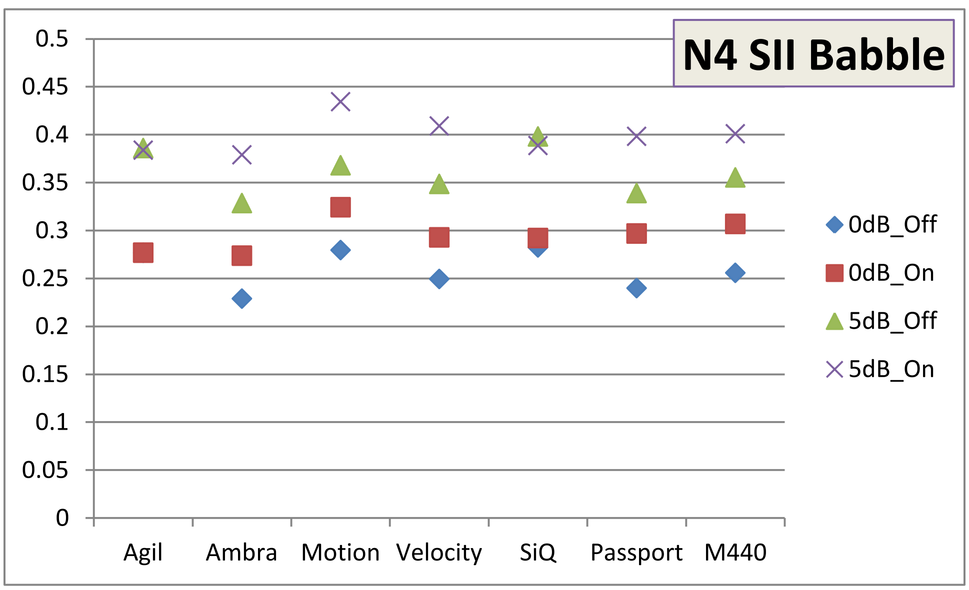


**Figure A-15: Comparison of SII values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

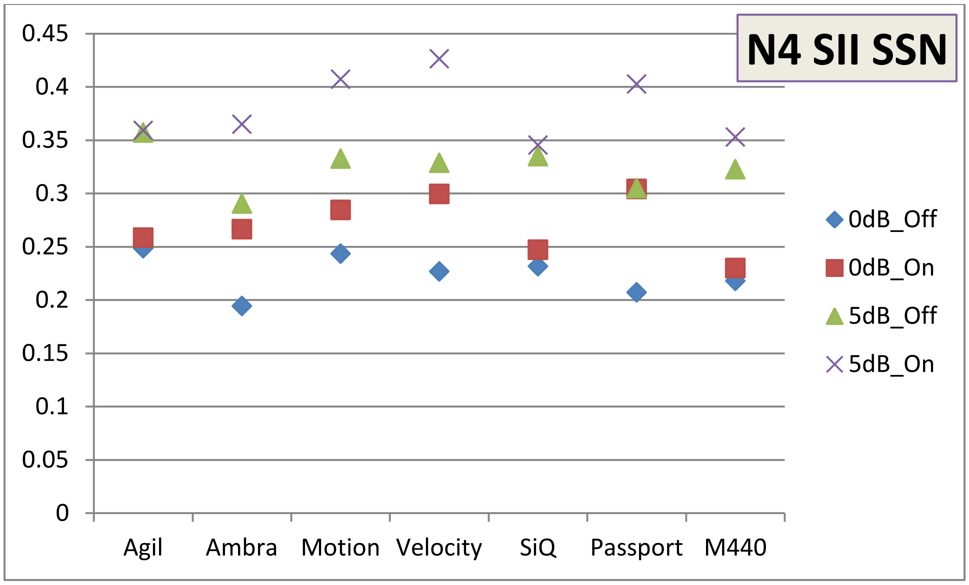


**Figure A-16: Comparison of SII values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

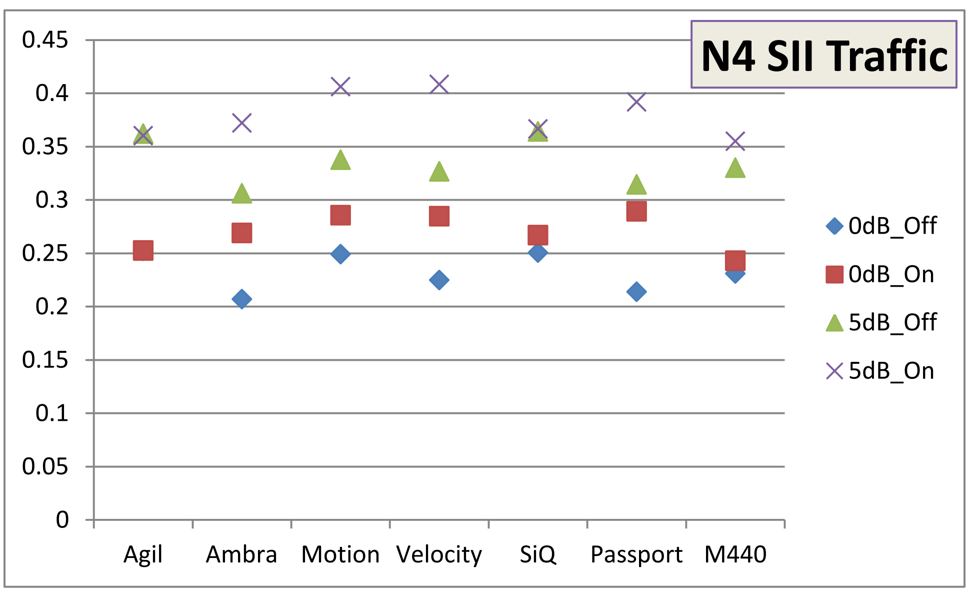


**Figure A-17: Comparison of SII values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

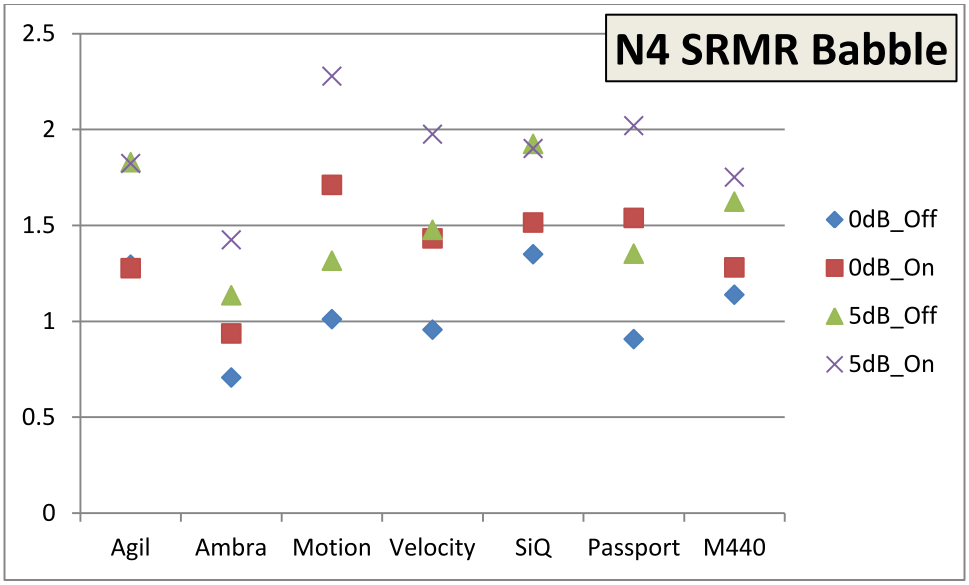


**Figure A-18: Comparison of SRMR values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

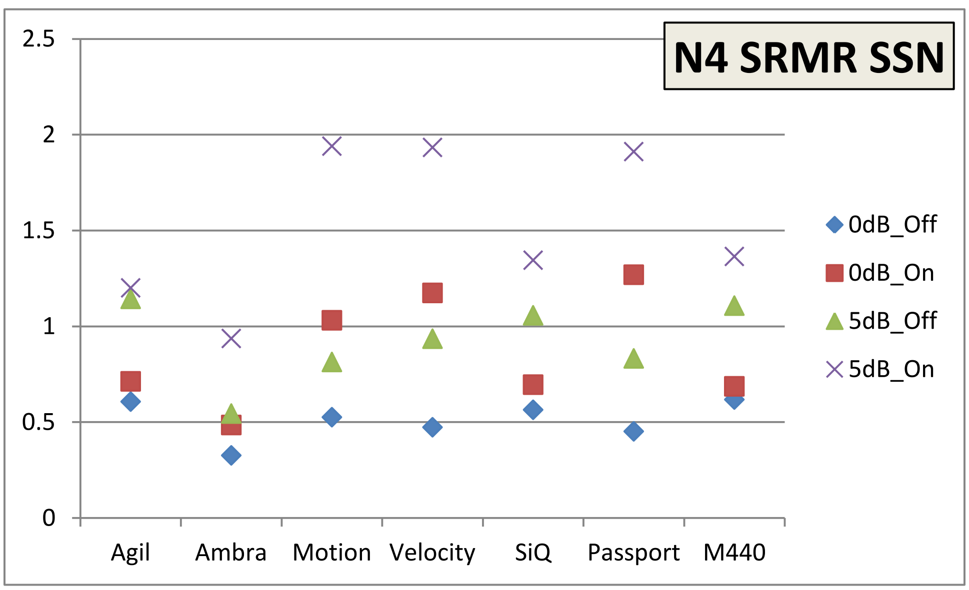


**Figure A-19: Comparison of SRMR values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

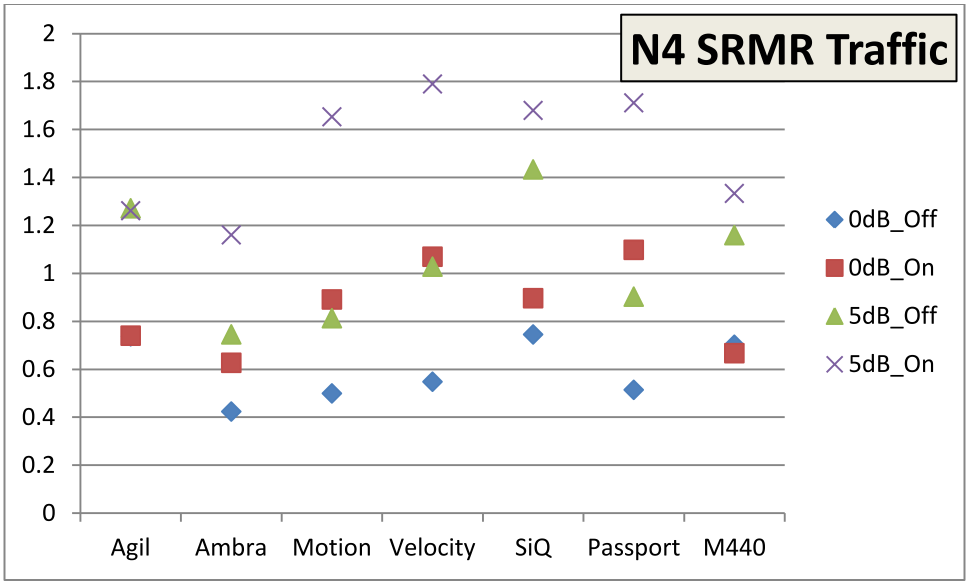


**Figure A-20: Comparison of SRMR values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

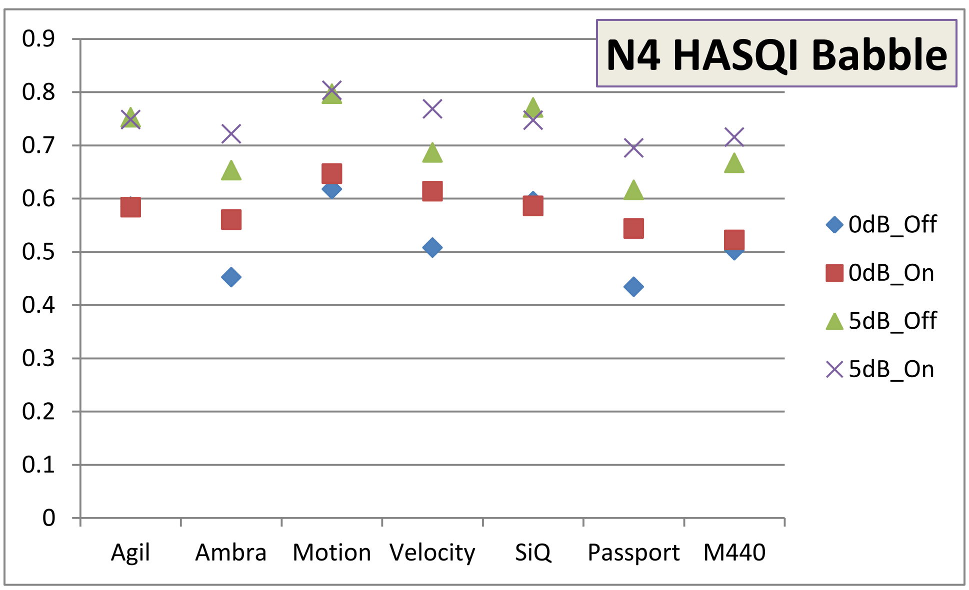


**Figure A-21: Comparison of HASQI values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is multi-talker babble.**

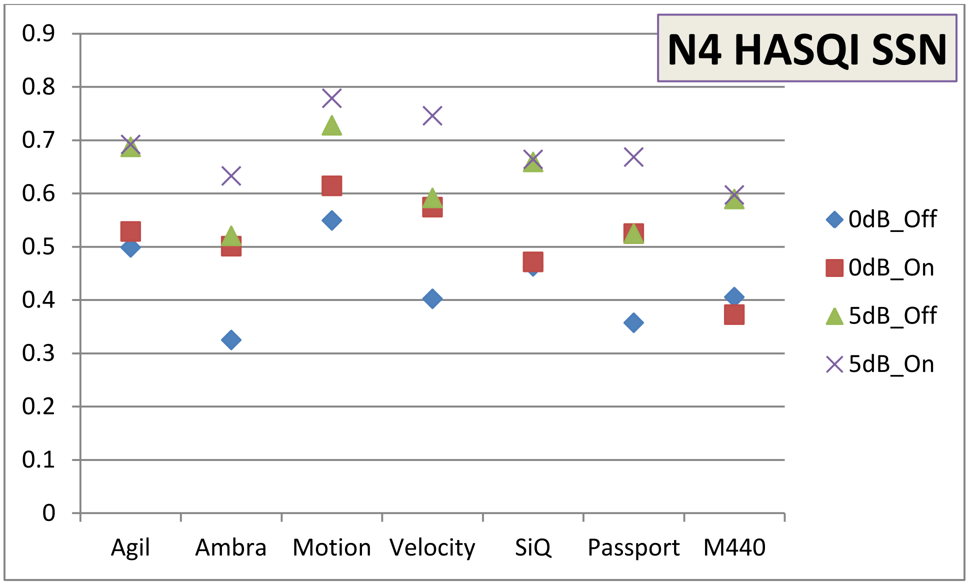


**Figure A-22: Comparison of HASQI values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is speech shaped noise.**

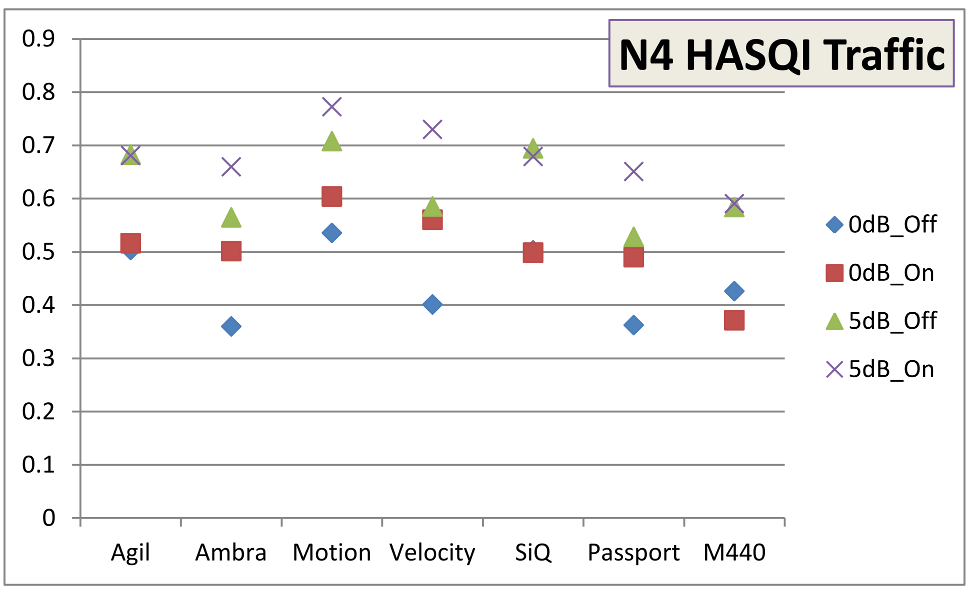


**Figure A-23: Comparison of HASQI values between the N4 DNR OFF and DNR ON conditions for 0 dB SNR and 5 dB SNR where the noise type is traffic noise.**

# Curriculum Vitae

| | |
|---|---|
| **Name:** | David Suelzle |
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**Publications:**

D. Suelzle, V.Parsa, T. Falk, "On a reference-free speech quality estimator for hearing aids," in *Journal of the Acoustical Society of America,* 133(5), EL412-EL418.

T. Falk, S. Cosentino, J. Santos, D. Suelzle and V. Parsa, "Non-intrusive objective speech quality and intelligibility prediction for hearing instruments in complex listening environments," in *IEEE International Conference on Acoustics, Speech and Signal Processing, Vancouver, Canada.* 2013.

D. Suelzle, I. Ibrahim, J. Pietrobon, V. Parsa, M. Cheesman, A. Farid, "Objective and Subjective Sound Quality Measurements of Binaural Wireless Hearing Aids," in *Proceedings of the International Hearing Research Conference, 2010. IHCON 2010,* August, pp. 45-46.

N. Pourmand, D. Suelzle, V. Parsa, Y. Hu, and P. Loizou, "On the use of Bayesian modeling for predicting noise reduction performance," in *IEEE International Conference on Acoustics, Speech and Signal Processing, 2009. ICASSP 2009,* April, pp. 3873–3876.