AN EVALUATION OF ACOUSTIC AND PERCEPTUAL EFFECTS OF FEEDBACK MANAGEMENT IN HEARING AIDS

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CERTIFICATE

This is to certify that this dissertation entitled **An Evaluation of Acoustic and Perceptual Effects of Feedback Management in Hearing Aids** is a bonafide work in part fulfillment for the degree of Master of Sciences (Audiology) of the student (Registration No. 10AUD017). This has been carried out under the guidance of a faculty of this institute and has not been submitted earlier to any other University for the award of any other Diploma or Degree.

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This dissertation entitled **An Evaluation of Acoustic and Perceptual Effects of Feedback Management in Hearing Aids** is the result of my own study and has not been submitted earlier to any other University for the award of any other Diploma or Degree.

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May, 2012

DEDICATED TO,

AMMA, APPA

AKKA, BHAVA

&

OUR JUNIOR

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Bowing to my beloved god with humbleness and greatfulness

I thank my lord for having bestowed the courage and strength in me which has always helped me to keep moving through all the walks of my life

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TABLE OF CONTENTS

List of Figures

List of Tables

CHAPTER 1

Introduction

Audible feedback is amongst the most prominent problems with hearing aids (Kochkin, 2002a). In a hearing aid, the acoustic feedback or squeal occurs when the output of the hearing aid leaks out of the ear canal and enters the hearing aid microphone and is amplified again. The acoustic leakage is often attenuated by the ear mould coupled to the hearing aid. The conditions necessary for audible feedback oscillation are met when the degree of attenuation is small and/or when the gain of the hearing aid is high (Kuk, Ludvigsen, & Kaulberg, 2002).

Generally, this feedback is associated with high gain hearing instruments. During such times, the annoyance, frustrations and embarrassment caused by the feedback may even outweigh an individual's otherwise perceived benefit from amplification. Acoustic feedback also can indirectly reduce the benefit from amplification. The hearing aid users may prefer to opt less-than-optimal gain to avoid the likelihood of feedback, or only use the hearing aids for situations known to be 'feedback free', or in extreme cases, simply stop using the hearing instruments (Chalupper, Powers, & Steinbuss, 2011). Thus, acoustic feedback is annoying and reduces the maximum usable gain of the hearing devices (Siqueira, Speece, Petsalis, Alwan, Soli, & Gao, 1996). These peaks, which are often high-frequency in nature, may produce an uncomfortable sharpness in the hearing aid processed speech and affect speech recognition (Cox, 1982; Freed & Soli, 2006). Acoustic feedback phenomenon can thus deteriorate the performance of digital hearing aids working at high gains, causing instability and speech degradation (Leira, Bueno, Pita, & Zurera, 2008). This phenomenon even contributes to the “hearing aid effect,” as potential users of amplification view acoustic feedback as a part of the negative stigma (Cox, & Alexander, 2000).

The main challenges faced by the hearing aid users prone to feedback problems are mostly threefold. First, it can distort the sound signal across all the frequencies, causing a noticeable reduction in sound quality. Second, it can be so annoying to the user’s environment that the user is forced to turn the instrument down, thus loosing the crucial speech information when it is most needed. And finally, it can restrict the full use of the volume control, which in turn limits the person’s ability to hear and understand speech.

Despite advances in technology associated with feedback management, feedback remains one of the most common patient complaints, irrespective of the technology used in the hearing aids. According to Kochkin (2005), 28% of hearing aid users report of dissatisfaction with their hearing aids due to feedback. Reasons for complaints regarding feedback may be attributed to several factors. First, audible acoustic feedback presents a high pitch whistling which many patients may find annoying. Second, this whistling may also be audible to others in close vicinity to the patient which can be a source of embarrassment for the potential hearing aid user. Third, the presence of acoustic feedback limits the amount of available gain to the patient. This is especially problematic for more severe hearing losses, as the amount of gain that the patient needs in order to receive benefit and/or satisfaction may not be achieved with specific styles of hearing aids without generating feedback (Chung, 2004). Fourth, the presence of acoustic feedback limits the size of the vent available on the hearing aid or ear mould and may contribute to the occlusion effect (Chung, 2004). Fifth, the presence of acoustic feedback can affect the recognition of speech as well as the sound quality of amplified sound (Freed & Soli, 2006).

Acoustic feedback is also associated, more often, with children. In young children the problem of feedback is exacerbated as the external ear is still growing (Riedner, 1978; Westwood & Bamford, 1995). Consequently, after certain weeks or months, the initially well fitted ear mould can become loose and hence may increase the probability of occurrence of feedback (Flynn & Flynn, 2006).

For the above reasons, consideration on feedback management in hearing aids is of extreme importance in pediatric population with severe hearing impairment, and more so in view of the recently introduced more comfortable open fittings in the ear canal. Acoustic feedback limits the maximum gain that can be used in a hearing aid making it unstable which results in whistling and distortion. Feedback reduction algorithms in hearing aids may provide a solution for some of these problems. Thus, the acoustic feedback suppression in hearing aids can increase the maximum insertion gain of the aid. The ability to achieve target insertion gain leads to better utilization of the speech band-width and through this, improved speech intelligibility for the hearing aid user can be expected as the most probable outcome (Siqueira, Speece, Petsalis, Alwan, Soli, & Gao, 1996).

Furthermore, Flynn and Flynn (2006) have stated reasons for significance of the applicability of Feedback Reduction Method in pediatric population. They are (1) it contains an increased feedback margin, i.e., the amount of gain that can be delivered prior to the occurrence of feedback, which allows the clinician to deliver the prescribed target gain across all frequencies without feedback. (2) As children grow and their ears change physically, the availability of an appropriate auditory signal without feedback enhances the length of time between ear mould remakes. (3) As the child experiences significantly less feedback, the entire amplification experience becomes more pleasant and the child and family are likely to experience less frustration and annoyance associated with the hearing aid fitting, thus yielding greater acceptance and use of amplification.

In addition, a good approach to addressing feedback also improves sound quality, makes soft sounds more audible, works better on the telephone, increases speech understanding in quiet and results in better physical fit and comfort. This is important because the overall satisfaction with hearing aids is known to increase as the number of situations in which the listener is satisfied increases (Kochkin, 2002a). Hence, the effects of feedback reduction are multi-natured. Application of the appropriate feedback management strategy in hearing aids is thus the current state-of-the-art.

Different approaches to feedback management have been introduced in hearing instrument technology (Dillon, 2001). With the advent of digital signal processing (DSP), audible feedback oscillation can be minimized without sacrificing gain, audibility, loudness, and speech intelligibility. The DSP-based electronic controls for minimizing audible feedback oscillations are desirable because 1) they permit greater usable gain 2) they allow the provision of adequate gain with an open ear mould or a shell with a large vent, and 3) certain types of feedback controls can adapt to changing environments, such as when a telephone is placed close to the ear (Merks, Banerjee, & Trine, 2006).

According to a survey report, one aspect of hearing instrument use with which wearers are significantly more satisfied today is feedback (Kochkin, 2010). In this area, the consumer satisfaction ratings improved by 12% compared to the 2004 ratings. However, it is not that the feedback is no longer problematic, as it remains among the most negatively ranked areas related to hearing instruments, but certainly progress in solving this issue is seen (Jespersen & Stender, 2002). This can essentially be attributed to the contributions due to the advent and progress in the field of digital technology in the name of digital signal processing. As a result, the introduction of digital technology has significantly improved the acoustic stability of modern hearing instruments.

The techniques proposed in literature to reduce the feedback in hearing aids can be broadly classified into feed-forward suppression and feedback cancellation techniques. In feed-forward suppression techniques, the regular signal processing path of the hearing aid is modified in such a way that it is stable in conjunction with the feedback path. The achievable increase in maximum stable gain with feed-forward suppression techniques has generally been found to be limited. In addition, feed-forward suppression techniques compromise the basic frequency response of the hearing aid, and, hence, may seriously affect the sound quality.

A more promising solution for acoustic feedback would be the use of a feedback cancellation algorithm. The feedback canceller produces an estimate of the feedback signal and subtracts this estimate from the microphone signal, so that, ideally, only the desired signal is preserved at the input. Since the acoustic path between the loudspeaker and the microphone can vary significantly depending on the acoustical environment the feedback canceller must be adaptive.

Currently available adaptive feedback cancellers can be divided into two classes: algorithms with a continuous adaptation (adaptive/dynamic) and algorithms with a non-continuous adaptation (static). The latter adapt the coefficients of the feedback canceller only when instability is detected or when the input signal level is low. Due to this reactive, rather than proactive, adaptation, such systems may be objectionable. A continuous adaptation feedback (CAF) canceller continuously adapts the coefficients of the feedback path estimate in such a way that the energy of the feedback-compensated signal is minimized.

Among the feed-forward and feedback reduction methods, generally two methods have been used off-late to counteract feedback in the digital hearing aids and they are the gain reduction (notch filters) and phase cancellation methods. The traditional procedure for increasing the stability of a hearing aid is a feed-forward suppression technique which reduces the gain at high frequencies. Gain reduction / notch filters, the approach used in most technology until 2004, reduce the gain in the frequency region where feedback occurs (Park, Kim, & Kim, 1998). While this approach can be successful in eliminating feedback, it also reduces the gain for target signals such as speech, especially in the frequencies which are important for speech resulting in reduced speech intelligibility (Lantz, Jensen, Haastrup, & Olsen, 2007). Thus, such algorithms are less commonly used in hearing aids. Phase cancellation systems, on the other hand, are capable of suppressing feedback without degrading the audibility of speech, and therefore, this type of feedback reduction is preferable (Chalupper, Powers, & Steinbuss, 2011).

For as long as feedback has been a problem, conventional and basic strategies to combat feedback have been utilized. Strategies often begin with making physical changes to the ear mould or shell. These changes include decreasing vent size, increasing canal length, damping the tubing, increasing the outside thickness of the tubing, changing ear mould style and material, and remaking the ear mould or shell to achieve a better fit (Lenzen, 2008).

Techniques for better ear mould venting thus reducing the size of the conventional vents, coupling of hearing aids, and fitting methods have also been proposed to reduce feedback problems (Cox, 1982; Dillon, 1991). Use of dampers in ear moulds can increase the usable gain of hearing aids. They increase the low frequency attenuation and provide greater high frequency output and a smoother frequency response by reducing the peakedness caused at high frequencies due to feedback (Valente, 1984).

According to Chalupper, Powers, and Steinbuss (2011), a good feedback cancellation must fulfill three requirements. It should provide effective feedback suppression without affecting target signals like speech, fast adaptation to changing environments, and robustness against artifacts.

Feedback reduction through phase cancellation differs from other feedback reduction strategies in that it does not reduce the forward gain of the hearing aid under normal operating conditions. Instead, the algorithm reduces feedback by formulating an internal estimate of the true hearing aid feedback signal and then subtracting this estimate from the microphone signal. The internal feedback signal is obtained by passing the hearing aid output signal through an internal, feedback path model. The feedback path model is intended to match the impulse response, or time response, of the external feedback path. If the internal estimate is accurate, then the actual feedback is cancelled and the hearing aid will not squeal. Maintaining an accurate estimate of the true hearing aid feedback signal is essential for the proper operation of the algorithm. Since the feedback signal changes over time, the algorithm must make continuous adjustments of the internal model to ensure accurate matching of the external feedback.

Current feedback-cancellation technologies are functioning efficiently. However, there are a few disadvantages:

1. The adaptation algorithm that is used to update the internal model is sensitive to the microphone signal characteristics. The adaptation algorithm is ideally trained using non-tonal or noise-like signals. Signals that are tonal, or music-like, can cause maladjustment of the internal feedback path model leading to renewed feedback or other audible artifacts. This maladjustment phenomenon is sometimes referred to as ‘entrainment’. This becomes significant when the hearing aid processes a harmonically rich signal such as music. Removing tonal components from the music signal will have an adverse effect on the sound quality.

2. Feedback-cancellation technologies may not respond well to a change in the feedback path. They may fail to continuously adapt to the changing environments and thus may be inefficient for long-term usage.

3. There may be peaky feedback path responses as a result of sharp resonances in transducer responses (most notably receivers); acoustic tube resonances and vibration coupling problems. All of these can lead to peaky feedback responses and, ultimately, to poor feedback canceller performance. For optimum feedback canceller performance, such resonances should be eliminated from the system design, as much as possible.

4. If a sudden change occurs in the external feedback path, there will be a temporary mismatch with the internal feedback path model, even though there is no change in the hearing aid gain. If the gain is sufficiently high, this may lead to a sudden, but temporary feedback event. Also, if the hearing aid gain changes too rapidly, even when the external feedback path is not changed, a small maladjustment of the internal model can be amplified and can lead to a temporary feedback event.

A further challenge of state-of-the-art feedback management systems is that they can cause distortions of natural signals, such as music, telephone ringing or door bells as the feedback cancellation system may mistakenly try to cancel a desired tonal input, producing a form of audible distortion (Merks, Banerjee, & Trine, 2006). Similarly, in hearing aids using dampers as a method for feedback reduction, recurrence of feedback may occur when the dampers are clogged with moisture (Valente, Dunn, & Roeser, 2000). Thus, it is essential to measure the efficacy of different feedback suppression methods in the currently available hearing aids. In addition, their effects in terms of speech perception abilities also need to be evaluated.

Need for the study

Feedback reduction algorithms in hearing aids alleviate the disturbance caused due to feedback (Saunders, 2010). By reducing the peaks in the frequency response with the use of dampers, the output at that frequency is reduced and thus it significantly reduces the low frequency harmonic distortion (Halevy, 1985). Feedback reduction algorithm also increases the gain before the level causing feedback is reached (Greenberg, Zurek, & Brantley, 2000). Hence, an improvement in speech perception abilities is the most probable outcome. There is a dearth of literature regarding the comparison of speech identification scores as a measure of speech perception ability with feedback suppression algorithm in hearing aids, with suitable acoustic modification (damper) and also with the combination of the two methods. Most of the studies taken up are with respect to the added gain provided by the different algorithms (Chalupper, Powers, & Steinbuss, 2011).

The demand for effective feedback cancellation techniques is increasing with the advent of newer technologies like digital hearing aids and open-fit hearing aids (Spriet, Rombouts, Moonen, & Wouters, 2006). Since the research is sparse, more research is required in this area. Also, it is necessary to measure the performance of the feedback reduction algorithms using both electro-acoustic measurement and perceptual measurement in listeners with hearing impairment. The present study focuses on evaluation of strategies of feedback reduction viz., the feedback suppression algorithm and suitable acoustic modification in hearing aids in terms of achievable gain and related benefits in performance.

Aim and Objectives of the study

The aim of the present study was to evaluate the feedback reduction methods in hearing aids, namely the feedback reduction algorithm and use of acoustic modification (damper). The specific objectives are as follows:

1) To evaluate the effect of the feedback reduction strategies in hearing aids, such as the phase cancellation method and use of damper in ear mould, on the insertion gain measures.

2) To evaluate the effect of the feedback reduction method and use of damper in the ear mould, on speech identification scores (SIS).

CHAPTER 2

Review of Literature

One of the major problems faced by hearing aid users when they require more gain is a high intensity oscillation called feedback (Masaki, 1997). The type of feedback commonly encountered in hearing aid fitting is acoustic feedback (Valente, 2002). Acoustic feedback in a hearing aid occurs when the output of the receiver leaks out of the ear canal and enters the microphone of the hearing aid and is amplified again (Chalupper, Powers, & Steinbuss, 2011). When sound again enters the microphone of a hearing aid, it is changed to electrical impulses and sent on to the amplifier. The amplifier intensifies the impulses, and the receiver then translates those stronger impulses into louder sounds. Acoustic feedback can cause oscillations and instability which can lead to a howling sound or squeal produced by the hearing aid.

The acoustic feedback is illustrated in the block diagram given in Figure 1. The input signal (X) is amplified by a gain factor (G) that results in an output signal (Y). If the hearing aid / ear mould provides a complete seal (i.e., no feedback path), the output signal (Y) would simply be determined by the gain (G) of the hearing instrument and the level of the input (X). That is, $Y=G+X$. Figure 1 shows the components and working of a hearing instrument.

Equation 1: Y = G+X

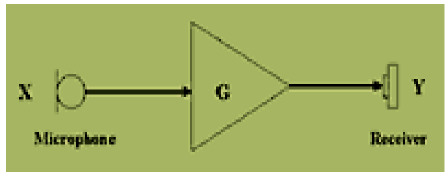


Figure 2.1: Block diagram of a hearing instrument (Kuk, Ludvigsen, & Kaulberg, 2002)

When a feedback path is present, a certain fraction (ß) of the output signal will leak back to the microphone. Figure 2 shows a simple block diagram of a hearing instrument that allows some of the amplified sound to leak back to its microphone (i.e., it has a feedback path). The feedback process is a looped sequence of events. First, the input signal X will create an output G+X. During the first loop, a certain fraction (ß) of the output signal G+X will leak back to the microphone and contribute to the input as ß(G+X). Thus, the combined input at the microphone will be [X + ß (G+X)]. Subsequently, the signal will be amplified by a factor G and contribute to the output signal (Kuk, Ludvigsen, & Kaulberg, 2002). That is, the output of the hearing instrument after one loop becomes: Y=G+X+G [ß (G+X)]. Figure 2 illustrates the feedback path showing the amplified sounds escaping back to the microphone.

Equation 2: Y = G+X + G [ß(G+X)]

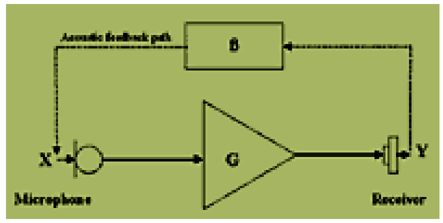


Figure 2.2: Block diagram of a hearing instrument that allows some of the amplified sound to leak back to the microphone (Kuk, Ludvigsen, & Kaulberg, 2002)

In addition to the acoustic feedback, there are other types of feedback in a hearing aid. A second type of feedback is the mechanical feedback. It occurs when physical vibrations are created due to contact between the hearing aid receiver and the hearing aid casing. These vibrations are then transferred through the casing back to the microphone. The third type of feedback is called the electronic feedback. This feedback is caused by a malfunction in the device's complex circuitry, requiring the services of a hearing aid technician to solve the problem of feedback (Chung, 2004). Of all the types of feedback, acoustic feedback occurs more commonly in hearing aids and it is very annoying for the hearing aid wearers (Kochkin, 2002a).

Thus, the aim of the present study was to evaluate the acoustic feedback reduction methods mainly the feedback reduction algorithm and use of acoustic modification namely the use of dampers in the ear mould attached to the hearing aids. The specific objectives were as follows:

1) To evaluate the effect of the feedback reduction methods in hearing aids, the phase cancellation method and the use of damper in ear mould, on the insertion gain measures.

2) To evaluate the effect of feedback reduction method namely the phase cancellation and use of damper in the ear mould (for feedback management), on the speech identification scores (SIS).

In the context of the objectives of the study, literature will be collected and compiled under the following headings:

2.1. Criteria and causes for occurrence of acoustic feedback

2.2. Effects of acoustic feedback

2.3. Need for acoustic feedback reduction methods

2.4. Need for acoustic feedback reduction methods in paediatric population

2.5. Simple troubleshooting procedures to reduce acoustic feedback

2.6. Feedback Reduction through Digital Signal Processing in Digital Hearing Aids and its influence in terms of :

a) Reduction in acoustic feedback

b) Improvement in useable gain, and

c) Improvement in Speech Identification Scores

2.7. Conventionally used methods to reduce the feedback and their efficacy

2.8. Studies in support of use of ‘Phase Cancellation’ method and use of dampers in ear moulds to reduce feedback.

2.9. Studies in support of use of phase cancellation method and use of dampers to reflect an improvement in usable gain
2.10. Studies in support of use of phase cancellation method and use of dampers to improve the speech identification scores
2.11. Studies contradicting the use of Feedback Reduction Algorithms
2.12. Need to study the acoustic feedback management methods and their acoustic and perceptual outcome.

2.1 Criteria and causes for occurrence of acoustic feedback

The occurrence of acoustic feedback is dependent and is determined by several factors which make the squealing phenomenon more likely to result than in other less likely conditions. Feedback occurs due to several causes. Agnew (1996) classified these causes into four categories - Acoustic leakage, hearing aid characteristics, user characteristics, and miscellaneous. Acoustic leakage includes a poor fitting of the ear mould or shell and a crack or break in the seal of the tubing. Hearing aid characteristics include excessive high frequency gain, wide vent, and style of hearing aid in relation to desired gain. User characteristics include traits distinct to the patient such as shape and size of the pinna, high ear canal resonance (i.e., greater than 2800 Hz), excessive cerumen, compliance of the tympanic membrane, and jaw movement. Miscellaneous characteristics consist of reflective surfaces near the hearing aid, such as a hat or wall, which can elicit feedback; as well as improper probe tube measurements. Feedback can occur due to one of these causes or a combination of several of these. A few criteria which are known to increase the possibility of acoustic feedback are listed and explained below.

2.1a. Nyquist Stability Criteria.

Acoustic feedback occurs when sound leaves the receiver and loops back through the microphone, where it is amplified a second time into a whistle or squeal. The presence of a feedback path alone is not sufficient for the hearing aid to squeal; but a condition called *Nyquist Stability Criterion* has to be violated. For sustained feedback to occur, the gain through the forward or direct path of the hearing aid must be greater than the attenuation through the feedback path (Parsa, 2006). The stage prior to the audible oscillatory stage i.e., before an evident feedback is seen, is also important from a sound quality point of view as in this stage, even if the feedback components are not strong enough to induce sustained oscillation, they can still impact the frequency response of the hearing aid (Parsa, 2006). Hence, the clarity of the speech perceived may be compromised even with slight initiation of feedback.

The conditions necessary for audible feedback oscillations are met when the degree of attenuation in the feedback path is small and/or when the gain of the hearing aid is high (Kuk, Ludvigsen, & Kaulberg, 2002; Olson, Müsch, & Struck, 2001a). Generally, this acoustic feedback is associated with high gain hearing instruments and also in cases where vents are used (Olson, Müsch, & Struck, 2001). It is more commonly seen in individuals who have high frequency hearing loss (Ross, 2006). This can be attributed to the fact that high-frequency sounds have a shorter wavelength than lower frequency sounds and hence they are more prone to escape from the ear and re-enter the microphone. Since high-frequency losses are among the most common hearing impairments, a person wearing the hearing aid may not even realize that there is a feedback.

2.1b. Hearing aid style and acoustic feedback.

Feedback is more problematic in CIC and ITC devices than in BTE devices because, in the former, there is less separation between the receiver and the microphone, which results in less attenuation between the receiver and the microphone. This leads to increase in the chances of occurrence of feedback.

2.1c. Acoustic environment near the ear / environmental conditions.

In addition, the feedback path is dynamic, implying that a change in the environment around the hearing aid (e.g., bringing a telephone closer, wearing a hat) will alter the attenuation characteristics of the feedback path and correspondingly affect the tonal composition of the feedback signal. It has been reported that audible feedback occurs only at selected frequencies (Parsa, 2006). The exact frequencies depend on the parameters related to fitting (amount of venting, ear canal characteristics, hearing aid gain curve, etc.) and the surrounding environment. Changes in the environment can result in a change in the composition of the feedback signal. Feedback can affect the hearing aid frequency response and hence its sound quality even when it is inaudible. Reflecting surfaces near the ear, such as a telephone handset or proximity of another person’s head, can temporarily reduce feedback path attenuation by guiding a larger portion of the output signal back to the hearing aid microphone.

2.1d.The fit.

If the hearing aid does not fit perfectly tightly into the ear, it will leave gaps for feedback to occur.

2.1e Ear mould fitting (leakage around ear mould / loose fit) and acoustic feedback.

Another common cause of feedback is poor and loose fitting of the ear moulds. A loose fit of the ear mould can create a more efficient path for the acoustic energy in the ear canal to feed back to the hearing aid microphone, i.e., there is a lesser attenuation between the receiver and the microphone and henceforth there will be greater incidence of occurrence of feedback. So, it is always preferable to use an appropriately fitting ear mould. A new impression and a new ear mould or shell remake may be required to reduce the feedback and the negative effects caused due to feedback, especially in case of children.

2.1g The volume setting used in the hearing aid.

When the hearing aid is set to a high volume, the sounds leaking out of the hearing aid will be louder. Louder sounds can further increase the chances of feedback as they can travel a longer distance.

2.1h. Venting.

Many hearing aids are fitted with a vent. This is a hole drilled into the shell of the ear mould and is a route for sound to come back from the receiver, thus enhancing the chances of occurrence of feedback.

2.1i. Ear wax.

A wall of ear wax blocking the ear canal is a very common cause for hearing aid feedback. Sound waves have pressure as they leave the ear mould. When the sound pressure leaving an ear mould or hearing aid hits a solid wall of earwax, it sprays in all directions, including out through the vent or any gaps between the ear mould or shell and the ear canal. This is the most common cause of hearing aid feedback.

2.1j. Wrong pointing of the ear mould.

One often overlooked problem that sometimes causes feedback is when the end of the hearing aid shell or mould is pointed incorrectly – if the original impression was not made long enough, or if it was too short, the mould/shell sometimes points into the wall of the ear canal instead of at the eardrum. The ear canal normally has two bends, forming an S-shaped curve. The ideal situation for the placement of the ear mould is for the end of the mould/shell to extend slightly beyond the second bend, allowing the sound to be "aimed" at the eardrum. Instead, if the end of the mould / shell terminates by pointing at the wall of the ear canal before the second bend, the sound is forced back out of the ear just as in the case of pointing towards ear wax. This is best solved by a longer canal, but even shortening, re-pointing, or "belling" i.e., using horns at the end of the canal may help.

2.1k. Leakage.

With behind-the-ear aids, there may be a crack or hole in the tubing, especially where it enters the ear mould, or a high power BTE may require a tubing with a thicker wall to prevent leakage. Leakage may lead to increased chances of occurrence of feedback.

Hence, Parsa (2006) opined that, the tonal nature of the feedback signal implies that this condition is met only at a few frequencies. The exact value of these frequencies depends upon a variety of reasons like (1) the gain characteristics of the hearing aid, and (2) the attenuation characteristics of the feedback path, which depend on the individual's ear canal characteristics, the nature of the fit, and the surrounding environment (nearby objects, room reverberation, etc.).

2.2 Effects of acoustic feedback

Audible feedback is among the most prominent problems with hearing aids (Kochkin, 2000b). Acoustic feedback is one of the most negative aspects associated with hearing instruments. In many cases, the annoyance, frustrations and embarrassment caused by feedback may even outweigh an individual’s otherwise perceived benefit from amplification. Acoustic feedback also can indirectly reduce patient benefit with amplification. Mechanical and acoustic feedback limits the maximum gain that can be achieved in most hearing aids (Lybarger, 1982). They may choose to use less-than-optimal gain to avoid the likelihood of feedback, and use the hearing aids only for situations known to be "feedback free," or in extreme cases, simply stop using the hearing instruments. Thus, acoustic feedback is both annoying and reduces the maximum usable gain of the hearing devices (Siqueira, Speece, Petsalis, Alwan, Soli, & Gao, 1996).

The acoustic feedback phenomenon can thus deteriorate the performance of digital hearing aids working at high gains, causing instability and speech degradation (Leira, Bueno, Pita, & Zurera, 2008). Acoustic feedback even contributes to the "hearing aid effect," as potential users of amplification view acoustic feedback as part of the negative stigma (Cox & Alexander, 2000; Chalupper, Powers, & Steinbuss, 2011). Acoustic feedback is also associated, more often, with children due to the loose fitting hearing aids (Flynn & Flynn, 2006). A good approach to addressing feedback also improves sound quality, makes soft sounds more audible, works better on the telephone, increases speech understanding in quiet and results in better physical fit and comfort. This is important because the overall satisfaction with hearing aids is known to increase as the number of situations in which the listener is satisfied increases (Kochkin, 2002a).

2.3 Need for acoustic feedback reduction methods

Acoustic feedback limits the maximum gain that can be used in a hearing aid making it unstable which results in whistling and distortion. To avoid feedback problems, the gain of the hearing aid must be held below a certain limit. However, doing this can create a new problem, since very often patients would benefit from gains above that limit. That extra gain may translate into improved audibility and speech understanding. Thus, feedback problems compromise the effectiveness of hearing aids, particularly for patients with severe losses (Dai & Hou, 2004; Chung, 2004). The presence of acoustic feedback can affect the recognition of speech as well as the sound quality of amplified sound (Freed & Soli, 2006). Sub-oscillatory acoustic feedback occurs when the gain of the hearing aid is slightly below the level at which oscillatory feedback occurs, resulting in peaks in the frequency response of the hearing aid. These peaks, often high-frequency, may produce an uncomfortable sharpness in processed speech and may as well affect the speech recognition (Cox, 1982; Freed & Soli, 2006).

Therefore, to realize the full potential benefit of hearing aids, effective management of feedback problems is essential more so in view of the recently introduced more comfortable open fittings in the ear canal. Feedback reduction algorithms in hearing aids may provide a solution for some of these problems. The acoustic feedback suppression in hearing aids can increase the maximum insertion gain of the aid. The ability to achieve target insertion gain leads to better utilization of the speech band-width and hence improved speech intelligibility for the hearing aid user (Siqueira, Speece, Petsalis, Alwan, Soli, & Gao, 1996).

Despite the advances in technology associated with feedback management, feedback remains one of the most common patient complaints regarding analog and digital hearing aids with 28% of hearing aid patients reporting dissatisfaction with their hearing aids due to feedback (Kochkin, 2005). Hence, there is an immediate need to compare the existing feedback management methods to evaluate their efficacy and to further improve the patient satisfaction in this sphere of management of individuals with hearing impairment.

2.4 Need for acoustic feedback reduction methods in paediatric population

Younger children fitted with hearing aids have a common, pervasive and annoying problem of acoustic feedback. It may make it impossible to provide a paediatric patients with the full spectrum of sound prescribed by the chosen fitting formula (e.g., NAL-NL1, DSL i/o). Currently, the advance in modern digital signal processing in hearing aid technology with regard to sorting the problem of feedback is the development of digital phase cancellation systems. Digital phase cancellation can instantly identify and remove feedback without removing other speech sounds. This allows the clinician to solve feedback issues while continuing to meet gain and frequency response targets (Flynn & Flynn, 2006). Moreover, it allows the full fitting rationale to be delivered to the child without valuable mid- and high- frequency speech information being removed (Flynn & Flynn, 2006).

2.5 Simple troubleshooting procedures to reduce feedback.

For the feedback process, the troubleshooting procedures involve problem solving strategies like a few basic solutions involving techniques for better ear mould venting, coupling of hearing aids (Cox, 1982; Dillon, 1991). Other important troubleshooting methods as suggested by Cox (1982) and Dillon (1991) are:

a) Hearing Aid or Ear mould Fit. Proper fit of the hearing instrument / ear mould to the ear is very important. The ear canal slightly changes in shape and size over time in children with age. As this happens, the hearing aid no longer seals the ear properly. Hence, the ear mould must be remade or modified in such conditions, so that the hearing aid seals properly in the ear. This change in the canal occurs every one to three years for most of the individuals. However, for severe or profound hearing losses remakes or modifications may be required as often as every 3 to 6 months.

b) Vent Size. The vent is a passage through the hearing instrument or ear mould, which equalizes the atmospheric pressure to the ear drum and allows excess amplification at various frequencies to escape from the ear canal. The vent needs to be reduced in size as acoustic feedback becomes more of a problem with its presence. Unfortunately in some cases, closing the vent becomes a necessity and as a result, the users may hear themselves walk, hear their own voices with an echo, and may experience moistness in the ears.

c) Reposition the Microphone and Receiver. Repositioning of the microphone and receiver is done by using longer canals and larger hearing instruments, which increases the distance between the microphone and the receiver. With this, the surface contact in the ear increases, thereby providing a more efficient seal. As the hearing loss becomes more severe, the microphone and receiver may need to be separated further by using behind-the-ear, power CROS or body worn hearing aids.

d) Removing the excess earwax. Earwax causes sound to reflect back to the hearing instrument instead of going through the ear. An indication of this may be the presence of wax on the tip of the hearing aid. Thus, the hearing instrument / ear mould should be cleaned frequently.

2.6 Feedback Reduction through Digital Signal Processing in Digital Hearing Aids.

Digital signal processing in hearing aids with feedback reduction achieves the goal of reaching the amplification targets without the limitations imposed by acoustic feedback. However, the first electronic feedback suppression system worked by reducing the degree of amplification at the feedback frequencies. That is, in response to acoustic feedback at some high frequencies, the hearing aid would automatically reduce the amplification (gain) at these high frequencies or the hearing aid would "notch out" the offending frequency by markedly reducing the gain around that point. Thus, if the feedback frequency were about 2200 Hz, the gain of the aid would be reduced, perhaps from 2000 to 2400 Hz. While both of these feedback reduction methods worked, in that more useable gain was possible before the squealing point was reached. The consequence was less audibility at frequency locations where the person may have required more gain (Ross, 2006).

With the advancement of digital signal processing (DSP) methods, audible feedback oscillation could be minimized without sacrificing the gain, audibility, loudness, and speech intelligibility. DSP-based electronic controls for minimizing audible feedback oscillations are desirable because they permit greater usable gain, they allow the provision of adequate gain with an open ear mould or a shell with a large vent and certain types of feedback controls can adapt to changing environments, such as when a telephone is placed close to the aided ear (Merks, Banerjee, & Trine, 2006).

An optimal feedback cancellation or suppression circuit will reduce the acoustic feedback without any undesirable modifications of the hearing aid's frequency response. A number of manufacturers now include this capability of preserving the frequency response in their hearing aids. Although each company has its own proprietary algorithm, they all apparently have one feature in common i.e., they all permit additional gain before the onset of acoustic feedback and they evidently manage this without any modification in the frequency response.

Research has shown that it is possible to achieve 10 dB, or possibly even more, overall added gain to a hearing aid before the onset of feedback. This is a big technological breakthrough in hearing aid industry. Generally, the feedback cancellation algorithms were less likely to sacrifice gain in specific frequency regions and better at reducing sub-oscillatory peaks, whereas the algorithms that used non-cancellation techniques were more tolerant of tonal input signals. In case of static feedback when the feedback path remains constant, the system will perform as expected and the feedback will be cancelled. The result will be an increase in headroom. The headroom increase is the amount by which the feedback-free gain can be increased with DFS active above the maximal feedback-free gain with inactive DFS.

In a study done by Olson, Musch, and Struck (2001), the feedback problems were found to be common i.e., out of 383 hearing instrument wearers tested in clinical studies, 48% experienced feedback with their preferred gain setting. For 73% of those with feedback, the preferred gain setting could be achieved through DFS. Since FBC algorithms simply cancel out the unwanted feedback, this technique should result in added stable gain (ASG) - an increase in the maximum stable gain (MSG) with 'FBC-on' compared to that with 'FBC-off'. MSG is the greatest amount of gain that can be provided without interference from audible feedback oscillation. Practical considerations such as environmental acoustics and the processing capabilities of the DSP chip in a hearing aid, limit the potential improvement in MSG to a maximum of 15-25 dB (Merks, Banerjee, & Trine, 2006). Figure 3 shows the ASG across the subjects taken for the study.

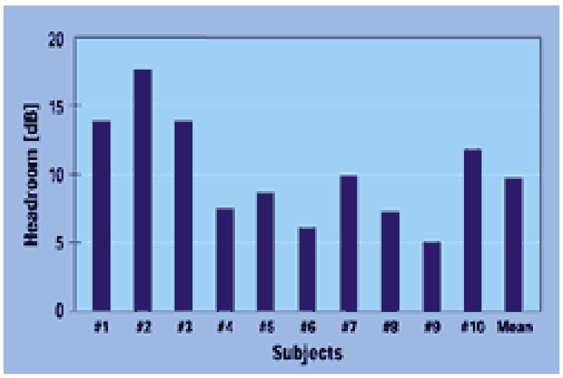


Figure2.3: Increase in feedback-free gain across all the participants with the mean value (Merks, Banerjee, & Trine, 2006).

Improvement in speech intelligibility is an important measure in evaluating the benefit provided by the Digital Feedback Suppression (DFS) method in hearing instruments. In order to evaluate this measure, speech perception measures were evaluated in a study with eight hearing aid users. Sentences from the Hearing in Noise test (HINT) were presented for fittings with and without DFS. When the system was not active, the hearing instrument could not provide as much gain as when it was active and the speech level had to be raised to sustain 50% intelligibility. Figure 4 shows the presentation level of speech in quiet that is required to correctly identify 50% of the sentences in the Hearing in Noise Test (HINT). The data show improved intelligibility with active DFS in eight of the ten subjects. For six of these eight subjects, the improvement is statistically significant. Figure 2.4 shows the performance of individuals on HINT test with and without the activation of digital feedback suppression method.

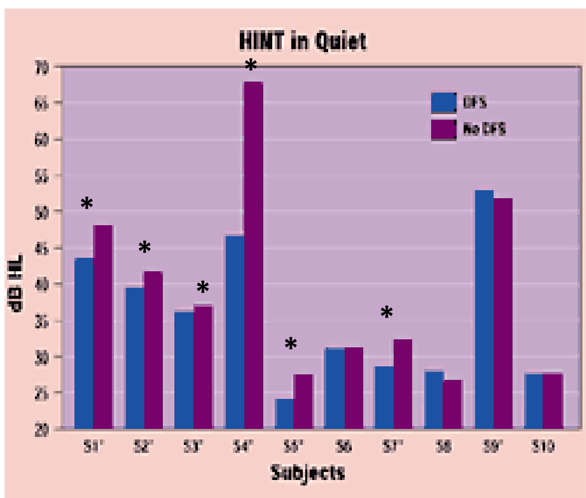


Figure 2.4: Performance on HINT test for ten listeners. Asterisk mark indicates significant performance differences ($p < 0.05$) (Merks, Banerjee, & Trine, 2006).

2.7 Conventionally used methods to reduce the feedback and their efficacy

Generally, two methods have been used most recently to counteract feedback. They are gain reduction (notch filters), and phase cancellation. The other methods of feedback reduction available are reducing the leakage (reduce vent size), reducing the gain, using adaptive notch filters, phase shifting and frequency shifting, moving the knee-point and adaptive phase cancellation. According to Chalupper, Powers, and Steinbuss (2011), a good feedback cancellation must fulfill three requirements. It should carry out effective feedback suppression without affecting target signals like speech, must have fast adaptation to changing environments, and must have robustness against artifacts.

2.7a Gain Reduction.

Use of gain reduction or notch filters was the approach used in most hearing instruments until 2004, which would reduce the gain in the frequency region where feedback usually occurred (Park, Kim, & Kim, 1998). While this approach can be successful in eliminating feedback, it also reduces the gain for target signals such as speech, especially in the frequencies which are important for speech resulting in reduced speech intelligibility (Lantz, Jensen, Haastrup, & Olsen, 2007) by attenuating the desired sound signal and reducing the audibility of important speech components. This is of particular importance as the feedback occurs around the range of 2000 to 3000 Hz which is also the most critical frequency regions for speech understanding, since it contains information such as second formant frequency of vowels and place of consonant articulation (Flynn & Flynn, 2006). This compromise has the greatest negative impact on patients with severe-to-profound hearing losses since they require a greater degree of amplification (Olson, Musch, & Struck, 2001). As in Figure 5, method of gain reduction is implemented in the high frequency region, which is important for the speech understanding. Thus, such feedback reduction algorithms are less commonly used in hearing aids. Figure 5 shows the frequency response with and without the use of gain reduction method.

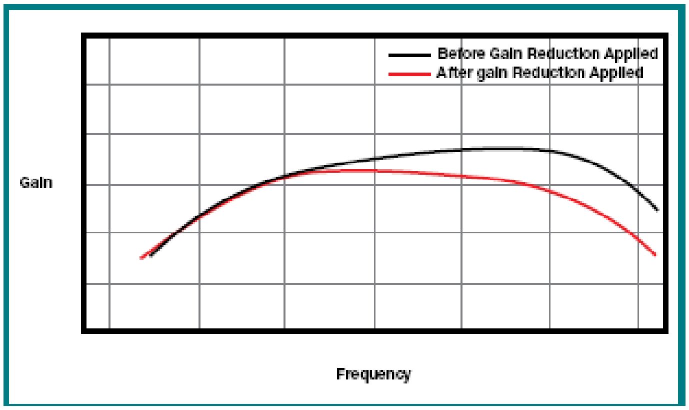


Figure 2.5: Frequency response with gain reduction method of feedback reduction

2.7b Moving the Knee-point of Compression.

Instead of a flat gain reduction across a wide range of frequencies, an alternative is to increase the compression threshold, or knee-point to control the occurrence of feedback. Figure 6 shows the frequency response after moving the knee-point.

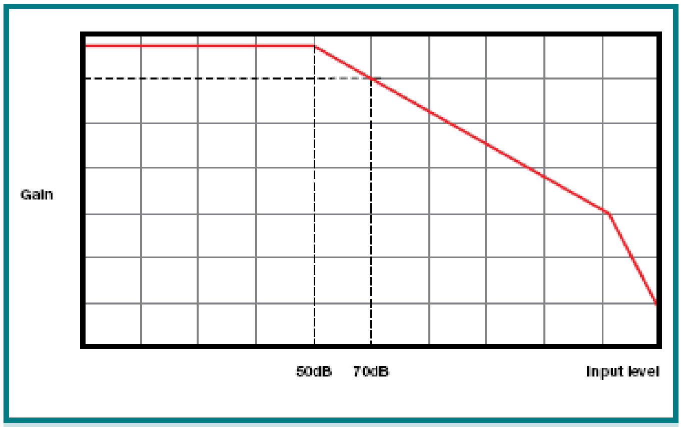


Figure 2.6: Frequency response with 'moving the knee-point' method

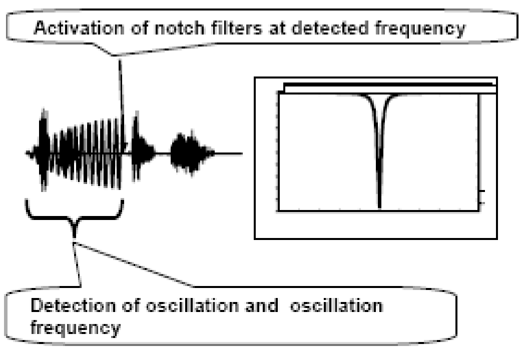
The compression threshold when increased causes the compression circuit to be active only at higher than the pre-set levels of knee-point. Hence, the gain reduction does not take place at low and moderate levels but occurs only at higher levels. The advantage, compared to the gain reduction method, is that an increase in compression threshold does not involve a gain reduction for all inputs. The compression knee-point could be increased up to the point that feedback is present and then only those inputs that are sufficiently loud to cause feedback experience a gain reduction.

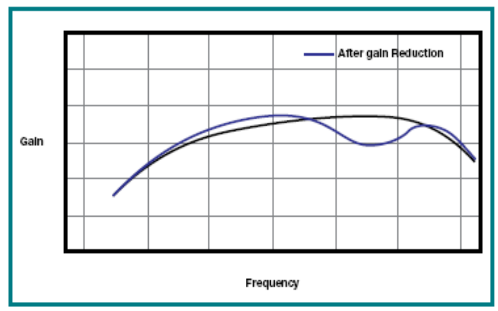
2.7c Notch filters.

Notch filters reduce the gain at those frequencies that cause feedback. The disadvantage is that signal audibility can be compromised around that frequency region. This is especially significant if the bandwidth of notch filter is wide and is further compounded when multiple notch filters are used.

According to a study done by Kates (1991), more than 10 dB of cancellation can be obtained and the Wiener filter is more effective in reducing feedback in the presence of strong interference. However, the frequency response of the device can be compromised even with the narrowest of notch filters. Secondly, there is a partial reduction in gain for frequencies adjacent to the band blocked by the notch filter. Maxwell and Zurek (1995) evaluated the efficacy of adaptive notch filters and found that such a method had a better effect only when the feedback path has a relatively narrow bandwidth. Moreover, intermittent failure of feedback suppression is more likely to occur with narrow notch filters as changes in the wearer's acoustic environment can easily shift the feedback frequency out of the notch filter's narrow suppression band thus making the feedback path more unstable across frequencies of its occurrence.

Thus, in order to achieve stable feedback suppression, a wider notch filter is generally necessary. Unfortunately, the wider the frequency band over which the notch filter reduces the gain, the greater is the loss in signal audibility and speech intelligibility (Olson, Musch, & Struck, 2001). Figures 7 and 8 indicate the spectrum and frequency response of the hearing aid, after using the notch filter method.





Figures 2.7 & 2.8: Spectrum and frequency response with notch filter method respectively

2.7d. Phase cancellation.

Phase cancellation systems, on the other hand, are capable of suppressing feedback without degrading the audibility of speech. Therefore, this type of feedback reduction is preferable (Chalupper, Powers, & Steinbuss, 2011). According to Maxwell and Zurek (1995), the maximum added wideband stable gain was approximately 12 dB with this method.

Generally, phase cancellation systems continually monitor the output of the hearing aid to determine whether some portion of the amplified signal contains elements that have the acoustic characteristics of acoustic feedback. When it does, the feedback circuit first determines the frequency, amplitude, and phase of the feedback component and then generates signals of opposite phase that will cancel (or markedly reduce) the feedback component. Since acoustic feedback is often a complex signal (like a tone with a series of harmonics), the cancellation process requires a complex solution, since more than one frequency is involved. This has to be done very quickly and has to be done adaptively. That is, since the characteristics of acoustic feedback often change (when chewing, talking, sitting in an armchair, etc.), the system must continually generate solutions to the changing feedback frequencies. Figure 9 illustrates the application of phase cancellation method to the feedback signal by creating an out-of-phase signal leading to the effective cancellation of the feedback signal.

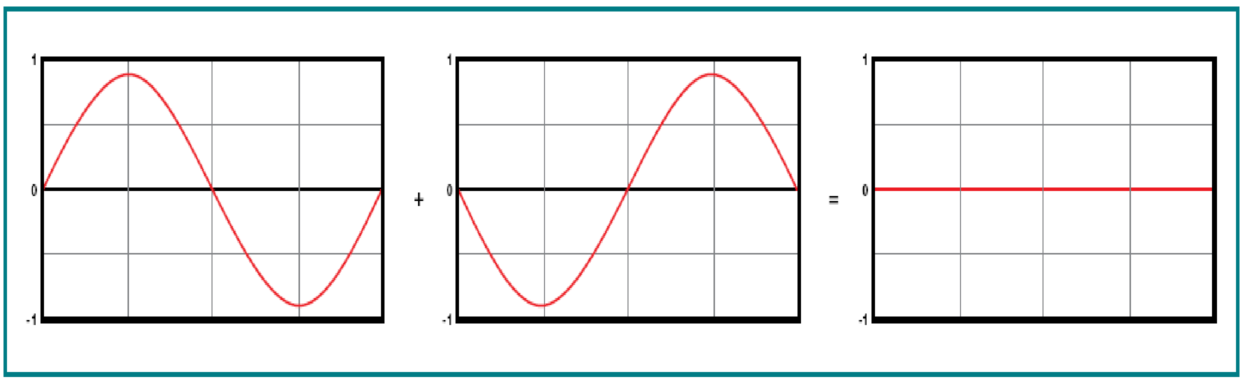


*Figure 2.9: Working of P*hase cancellation method, showing cancellation of out of phase signals.

2.7e. Frequency Shifting.

In this approach, the signal is modified so that its frequency spectrum appears shifted up or down by a certain amount of Hertz. In other words, in this method, the signal is frequency shifted by a given amount, for example 5 Hz, so that the output would be a frequency shifted version of the input (Schroeder, 1961). The output signal spectrum S out (f) is consequently given as S out (f) =Sin (f+ Df) where Df indicates the amount of shifted frequency. Frequency shifting is a very efficient approach for treating the feedback problem. It actually prevents the problem to occur because a microphone-receiver-microphone loop on any particular frequency cannot be created as each frequency is constantly shifted up or down. However, shifting in frequency violates harmonic character of some speech sounds.

For example, for voiced sounds in speech, all constituting frequencies are multiples of a fundamental frequency called pitch. If the pitch frequency is fp then a voiced sound consists of frequencies fp, 2fp, 3fp, … nfp . Shifting these frequencies by Df will give fp+ Df, 2fp+Df, 3fp+Df, … nfp+ Df . Accordingly, the harmonic structure of such a signal is violated as k (fp + Df) kfp+Df is resulted instead of the unshifted harmonics.

2.7f Phase Shifting.

Phase shifting uses all pass filters to avoid 0° phase which is physically not possible at all the frequencies simultaneously. It continuously changes the phase characteristics of the output signal, by the process of changing the phase of the signal during the forward path of the hearing aid. They suppress the feedback but not effectively. Hence, it is not very popular due to its low effectiveness in reducing the problems caused due to feedback.

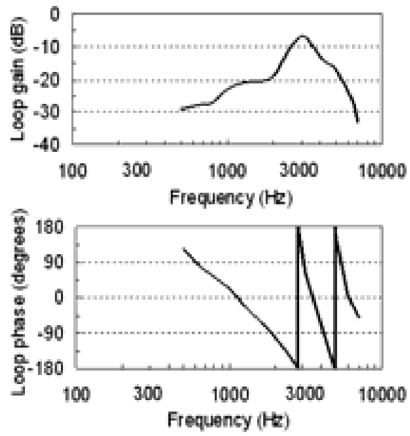


Figure 2.10: Frequency response with phase shifting method

2.7g. Frequency Modulation.

Yet another plausible procedure for reducing feedback in hearing aids was proposed by Nishinomiya (1968) - frequency modulation. In this method, the output signal is frequency modulated so that the stationary feedback relationship between the receiver and the microphone is broken. The modulation will prevent the feedback signal from being continuously in phase with the incoming signal. According to Egolf (1982), Nishinomiya obtained 7 dB additional stable gain using this method. Conclusively, Nishinomiya pointed out that only frequency ranges where feedback is most likely, should be modulated to prevent listener annoyance; frequencies below 500 Hz should be passed without changing them through the system to prevent "warbling". However, it was concluded by Masaki (1997) that the speech quality was degraded even when the gain was 6 dB below the threshold value.

According to this study, two alternative methods of dealing with feedback oscillations were used called Adaptive Feedback Suppression (AFS) and Adaptive

Feedback Cancellation (AFC or FBC). AFS searches for oscillations and reduces the gain at the appropriate frequency when an oscillation is detected. FBC models the feedback path inside the hearing aid, and subtracts the modeled feedback signal from the microphone input signal to cancel the acoustic feedback.

In a study, by Blamey, Hau, and Fulton (2006), the amount of added stable gain (ASG) for the FBC was determined to be 15 dB. ASG is defined by the difference between the maximum amount of gain achievable before feedback occurs with the FBC turned ‘on’ and the maximum amount of gain achievable before the feedback occurs with the FBC ‘off’. These measurements of ASG were made using a BTE hearing aid placed on a Head and Torso Simulator (HATS) in a sound booth. An ASG of 15 dB enables the fitting of greater degrees of hearing loss without feedback and increases the potential for both open fittings and venting in ear moulds and custom aids. The results with the AFS changed slightly as it works on a different principle. With the AFS used in the study, the maximum gain reduction of 12 dB was evidenced.

Apart from the static feedback suppression methods, automatic and adaptive acoustic feedback reduction is one of the most beneficial algorithms offered according to a study which reports an additional 7 dB of added stable gain making added stable gain as high as 25 dB with frequency shifting method (Chalupper, Powers, & Steinbuss, 2011).

2.8 Studies in support of use of 'Phase Cancellation' method and use of dampers to reduce feedback

One of the viable options is to reduce the gain where the feedback occurs i.e., only at the peaks of the frequency response curve without reducing the gain at the other composite frequencies. This can be done suitably in analog hearing aids, as the digital feedback management methods cannot be applied in such hearing aids to reduce the occurrence of feedback. Henceforth, acoustic dampers can be used for this purpose suitably (Dillon, 2001). It is well known that dampers reduce or smoothen the peaks at the mid frequencies and thus gives a smoother frequency response curve. The dampers can reduce the frequency response at higher frequencies where the feedback occurs (Dillon, 2001).

It is convenient to discuss the practical implications of an FBC system in terms of two general hearing loss categories, those with severe to profound hearing losses and those with relatively good hearing in the low frequencies. The major advantage for those with the most severe hearing losses is that an FBC system can help them to reach the target amplification goals without the necessity of tighter earmoulds. An additional 10 or 15 dB gain before the onset of feedback may solve the problem of reduced speech intelligibility. With phase cancellation, approximately 10 dB more gain before feedback can be obtained than without phase cancellation. Practically, the outcome is more complicated. The amount of improvement (increased gain before feedback) depends on many other factors, including the type of fitting, the frequency zone, the type of input signal, length of time, the generation of software, and the anatomy of the ear canal so on and so forth (Ross, 2006).

In an experimental study done by a hearing aid manufacturing company, three aspects of FBC circuits were compared for six hearing aids from six different manufacturers. In the first comparison, the six aids were compared with regard to the additional gain possible before the onset of feedback. This is the most basic comparison and the one that would most directly affect listeners. ASG also referred to as “headroom” or “gain margin” can be defined as the additional amount of gain the feedback algorithm allows before the hearing aid produces an oscillating response (Lenzen, 2008). The second dimension evaluated was termed as "entrainment." This occurs when the FBC system mistakenly tries to cancel a desired tonal input, producing a form of audible distortion. The third area evaluated was the adequacy of the feedback circuit when confronted with an object (such as a phone, hand, hat, etc.) when it was moved closer to or farther from the hearing aid. Such movements often produce undesirable sounds in real life.

The ASG was considered as the primary yardstick used to quantify the adequacy of FBC. However, a high ASG is valuable only if the hearing aid retains good sound quality and does not show entrainment artifacts in response to tonal input signals (Merks, Banerjee, & Trine, 2006). In the key performance dimension, which is the added gain before feedback, the results indicated a fairly substantial range of added gain across manufacturers, from 3.5 dB to about 16 dB, with four of the six hearing aids reaching about 10 dB or more. The ASG is often limited if FBC is not active at low frequencies. This is usually done to avoid entrainment artifacts.

Similarly, in the laboratory tests of a wearable digital hearing aid, a group of subjects with hearing impairment used an additional 4 dB of gain when adaptive feedback cancellation was engaged and showed significantly better speech recognition in quiet and in a background of speech babble (Engebreston & St.George, 1993). Field trials of a feedback cancellation system built into a BTE hearing aid have shown increases of 8-10 dB in the gain used by subjects with severe impairment (Bisgaard, 1993) and increases of 10-13 dB in the gain margin measured in real ears (Dyrlund, Henningsen, Bisgaard, & Jensen, 1994).

Considering, all the caveats/ warnings aside, three major advantages of using phase cancellation as reported by Martin and Robert (2006) are, first - it makes open-canal fittings feasible. Without phase-cancellation technology, it is not possible to develop any appreciable real-ear gain. Second - many patients who have severe hearing loss have large air-bone gaps. They would need as much as gain as possible. Phase cancellation technology improves the capacity to provide them with higher levels of amplification and tremendous improvements in word understanding without feedback. Consequently, most of the improvement will be in the higher frequencies (2000-5000 Hz) where the feedback occurs. Third - people are much satisfied if larger vents were provided. This was difficult to achieve in the past. The instrument would go into feedback too easily for attempts to provide significant gain while venting. Currently, improved hearing and improved word understanding are possible because of enhanced comfort provided by large vents and increased gain in the high frequencies made possible by phase cancellation technology.

Computer simulations and prototype digital systems indicate that increases in gain of between 6 and 20 dB, can be achieved in an adaptive system before the onset of oscillation, and no loss of high frequency response is reported (Kates, 1991; Dyrlund & Bisgaard, 1991; Engebretson & George, 1993; Kaelin, Lyndgren, & Wyrsch, 1998). Figure 2.11 indicates the frequency range over which the feedback management method gives additional gain.

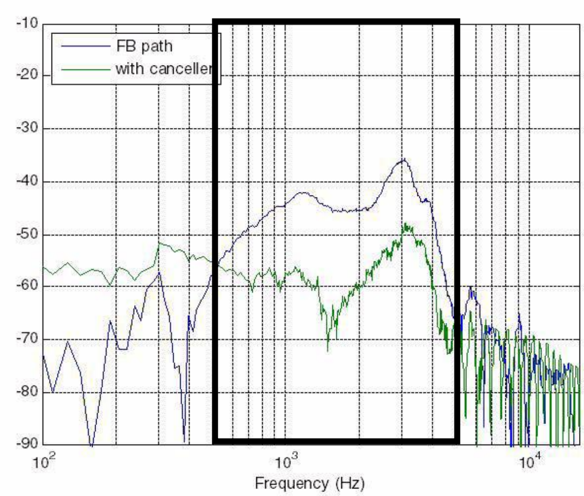


Figure 2.11: Change in feedback-path frequency response before and after the application of feedback canceller along with ASG values across the frequency range (Kates, 1999).

Hence, with the activation of feedback management strategy, an added stable gain can be achieved at frequencies from 500 Hz to 5000 Hz as it is evident from Figure 2.9. The rectangular box in dark solid line is used to indicate the frequency range in which the feedback cancellation is most beneficial to derive an added stable gain. Thus, an overall increase in the gain before the onset of feedback and hence an improvement in speech identification scores with the expected benefit of feedback suppression was the most probable outcome of an effective feedback cancellation algorithm like phase cancellation method of feedback management.

2.8 Studies contradicting the use of /benefit from Feedback Cancellation Algorithms

The efficiency of the feedback management method depends on the amount of added stable gain provided before the onset of feedback. However, the activation of features such as ASG often depends on the lowest frequency at which the FBC operates. FBCs that operate only at higher frequencies will offer no ASG for patients with significant acoustic leakage at low frequencies. Small ASGs offer little or no advantage in delivering necessary gain (Merks, Banerjee, & Trine, 2006).

In a study by Merks, Banerjee, and Trine (2006), there was no evident ASG added for a few patients and the resultant performance was unchanged even when the FBC was 'on'. In general, ASG for a given fitting was dependent on the ASG predicted by the benchmarking method and the maximum gain of the hearing aid (Merks, Banerjee, & Trine, 2006). Another experimental study by a hearing aid company cautions that some FBC systems may produce unacceptable distortion by-products. This study did not report of added gain before feedback. The efficiency of these feedback suppression features on activation in real life situations on performance dimensions is however, yet to be further investigated.

A study carried out by Egolf (1982) revealed that even though a maximum additional stable gain of l0-l2 dB was obtained with the phase shifting method, the intelligibility of the speech was sacrificed. Additionally, the subjects heard “audible beating" when the gain was greater than 6 dB. Therefore, the maximum effective additional stable gain obtainable while retaining good speech quality was only about 6 dB.

Entrainment is the term used to refer to the phenomenon of a feedback-cancellation algorithm which confuses musical tones or the beeps with feedback and puts the hearing aid into an inappropriate cancellation mode that causes disturbances in the output. Most of the amplifiers with feedback algorithms are known to exhibit entrainment under some conditions. But the problem can be overcome by following a few strategies like:

(1) If the entrainment occurs only rarely, it may be enough to adjust the compression carefully and counsel the patient. The correct AGC (compression) setting will reduce the discomfort, even during entrainment.

(2) Switching to another manufacturer as different product brands can vary significantly in how they respond to particular entrainment inputs.

(3) Giving the patient a lower gain program which can be turned on in such situations in which the feedback canceller is turned off to reduce the negative effects caused due to entrainment (Martin, & Robert, 2006).

2.2 Need to study acoustic feedback management methods and their acoustic and perceptual outcome

There are studies on the effect of different feedback management methods with regard to reduction in feedback and hence the added stable gain provided. However, there is a dearth of information regarding the comparison of the methods like phase cancellation which is available as a feedback management option in digital hearing aids and use of dampers in the ear moulds along with hearing aids. Although there much has been written regarding the potential benefits of feedback management methods for both reduction in feedback and improvement in speech intelligibility, there are not many published data to support these claims.

From the review of literature it is evident that there is a need to carry out a comparison of feedback management methods - the digital feedback management strategies (phase cancellation) and the use of damper in ear mould and their acoustic effects in terms of added stable gain and perceptual effects in terms of improvement in speech identification scores. The latter strategy, if found useful, would have application with analog hearing aids. This would help to judiciously evaluate the feedback management methods and to further quantify the benefit in terms of better availability of gain and improved perception of speech.

CHAPTER 3

Method

The aim of the present study was to evaluate the efficacy of the feedback reduction methods in the hearing aid, namely the phase cancellation algorithm and the use of acoustic modification (damper) in the ear mould. The specific objectives were as follows:

1) To evaluate the effect of the phase cancellation method and the use of damper in ear mould, on the insertion gain measures.

2) To evaluate the effect of the phase cancellation method and use of damper in the ear mould, on speech identification scores (SIS).

The following method was adopted to investigate the objectives of the present study.

3.1 Participants

The data were collected from a total of 60 ears of 30 children, in the age range of six to eight years. Table 1 indicates the details of participants.

Table 3.1: *Details of the participants*

Participants	Male	Female	Total
Number (N)	19	11	30
Mean Age (in years)	6.5	6.609	6.55

The participants considered for the current study had Kannada as their mother tongue. All the participants had pre-lingual bilateral severe to profound hearing loss, pure tone average ranging from 75 to 120 dB HL in the speech frequencies. All of them had flat or gradually sloping (with a slope of <15 dB per octave) in both the ears. On immittance evaluation, all the participants got 'A' type tympanogram with reflexes being absent. TEOAEs were absent in both the ears revealing outer hair cell dysfunction in both the ears. Auditory Brainstem Responses (ABRs) were absent in both ears for all the participants. All the participants were using binaural Behind-The-Ear (BTE) hearing aids. Their hearing aid was programmed to a gain lesser than the target gain due to the occurrence of feedback. With this gain setting, all the participants obtained an aided closed-set speech identification (through picture identification task) score of 50% or greater in both the ears with Behind-The-Ear hearing aids.

The thresholds for the frequencies at which the feedback occurred mostly from 1500 Hz to 6000 Hz (Martin, & Robert, 2006) was equal to or greater than 90 dBHL for all the participants, irrespective of minimal residual hearing at low frequencies till1000 Hz.

The participants had no significant history of otologic or neurologic problem. They had a negative history of cognitive or psychological problems. All the participants attended listening and speech therapy for a period of at least of three months and they had the auditory skills at least for the identification of words.

3.2 Equipment and Test Material

3.2.1 Instrumentation.

1) A calibrated GSI Tympstar middle ear analyzer (version 2) was used to rule out the presence of middle ear pathology.
2) A Fonix 7000 hearing aid test system (computer controlled real-time analyzer version 1.70) with probe tube microphone option was used to perform insertion gain measurements.
3) A calibrated dual channel diagnostic sound field audiometer Madsen OB922, with the facility of talk forward and talk back, was used perform puretone, speech and aided testing.
4) A personal computer with NOAH-3 and hearing aid specific software with Hearing Instrument Programmer (Hi-Pro) interface were used to program the hearing aids and to activate or de-activate the feedback reduction algorithm.
5) A digitally programmable two channel Behind-The-Ear hearing aid with a fitting range for severe-to-profound sloping hearing loss with custom made soft shell ear mould was used for the testing. The hearing aid had 2 channels and 8 bands with 4 programmable memories. The hearing aid had a maximum output level of 135 dB SPL with a maximum gain of 70 dB and a reference gain of 52 dB. The basic frequency response was from 200 Hz to 6400 Hz. The hearing aid utilized 'Active Feedback Intercept' which worked on the principle phase cancellation method for feedback management.

The phase cancellation algorithm introduces an additional signal to cancel out the acoustic leakage. When feedback is detected at the output of the hearing aid, a cancellation signal is generated to mimic the feedback. The feedback is eliminated by subtracting the cancellation signal from the input without compromising the gain across frequencies.

When the feedback is generated, an additional signal of same frequency characteristics but opposite phase is introduced to cancel out the feedback signal which is out-of-phase with respect to the generated signal. Figure 3.1 demonstrates the working of phase cancellation method by generating an out-of-phase signal to cancel the feedback.

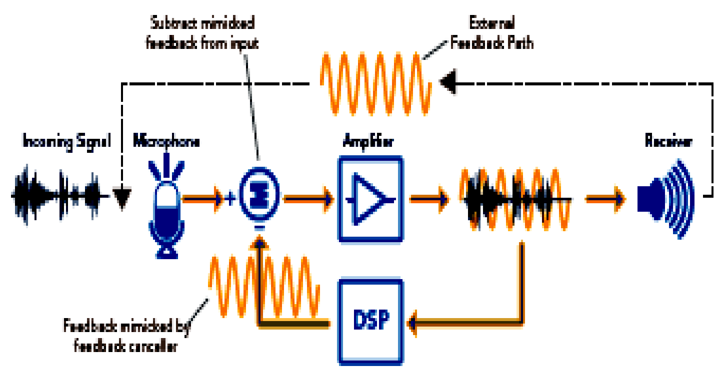


Figure 3.1: The working principle of phase cancellation method (Merks, Banerjee, & Trine, 2006).

3.2.2 Test material.

Phonemically balanced (PB) word list in Kannada developed by Vandana and Yathiraj (1998) was used to find out the closed-set Speech Identification Scores (SIS) in quiet through monitored live-voice presentation.

3.3 Test Environment

All the testing was carried out in an air-conditioned sound treated double room situation.

3.4 Procedure

The study was carried out in three different phases.

Phase I: Selection of participants

Phase II: Insertion Gain measurement

Phase III: Aided behavioural testing

3.4.1 Phase I: Selection of participants.

3.4.1.1 Audiological evaluation.

A detailed case history was taken to confirm the participant inclusion criteria. The routine audiological tests including pure tone audiometry, speech audiometry and immittance evaluation were carried out for all the participants for each test ear. The pure tone audiometry was done by estimating the air-conduction thresholds between 250 Hz to

8000 Hz at audiometric frequencies. The bone-conduction thresholds were also estimated between 250 Hz to 4000 Hz

Speech audiometry was carried out for the participants through which speech detection thresholds (SDT) and Speech Identification Scores (SIS) were measured. Immittance evaluation was carried out using a 226 Hz probe tone. Tympanogram was obtained. Ipsilateral and contralateral reflex thresholds were tested at 500 Hz, 1000 Hz, 2000 Hz and 4000 Hz. These tests were carried out in order to select the participants satisfying the criteria for the study. TEOAEs were also measured which was absent in all the test ears. The ABR were absent at 90 dBnHL in all the test ears.

3.4.1.2 *Speech Identification Scores (SIS) for selection of participants.*

The participants were fitted with a two channel digital BTE hearing aid. The hearing aid was connected to the programming hardware (Hi-Pro) through a cable and was detected by the programming software. The hearing thresholds of each participant were fed into the programming software and target gain curves were obtained using the proprietary prescription formula. Following this, the hearing aid was programmed to match the target gain. The aided closed-set speech identification scores were calculated as the number of words correctly identified out of a total of 25 words presented. The response mode was through the picture identification. A score of 50% and above was considered as the criterion for the inclusion of participants in the current study.

3.4.2 Phase II: Insertion Gain measurement procedure.

In order to measure the amount of gain/output delivered by the hearing aid at the participant's ear canal, insertion gain measurement was performed using a calibrated and levelled Fonix 7000 system. The testing involved a set of systematic steps to obtain the required data. The steps followed to obtain data of insertion gain measurements from the participants are given below.

a) Participant Seating Arrangement.

1) The loudspeaker of the real ear measurement system was placed at approximately 12 inches and at 45^0 azimuth from the test ear of the participants.

2) The height of the loudspeaker was at the level of the test ear.

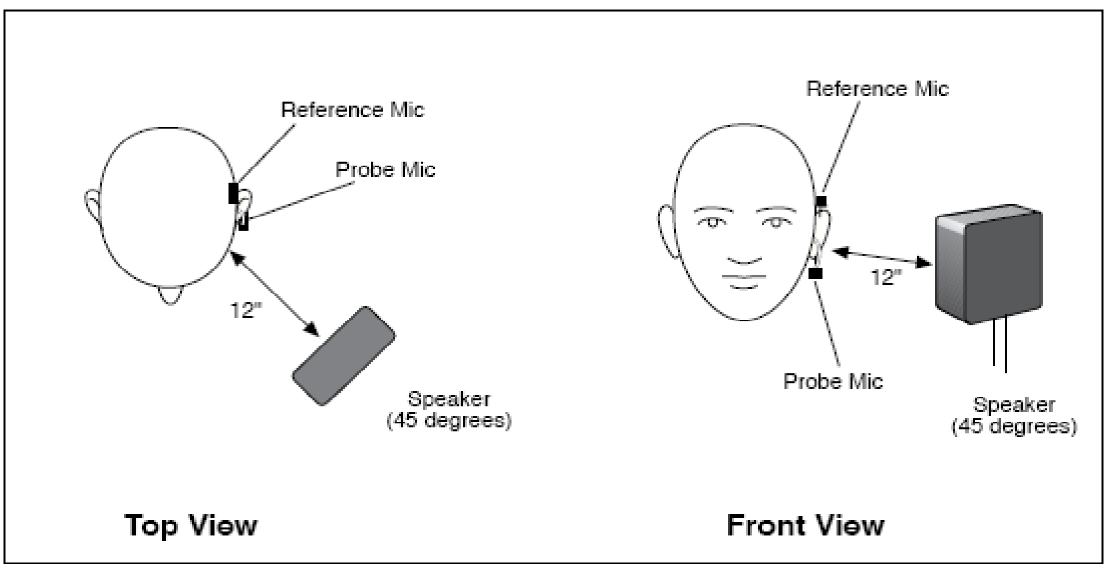


Figure 3.2: Test set-up for insertion gain measurement

b) Placing the earhook and reference microphone.

1) The integrated probe microphone was placed on the test ear of the participant.

2) The reference microphone was secured on the earhook above the ear.

3) The earhook slider was adjusted up or down for optimal positioning of the probe tube into the participant's ear.

4) After the probe tube was inserted, the probe microphone body was pivoted towards the ear to help hold the probe tube in place.

Figure 2 depicts the placement of reference and probe microphones in the ear for data collection. It also depicts the components of the real ear measurement system.

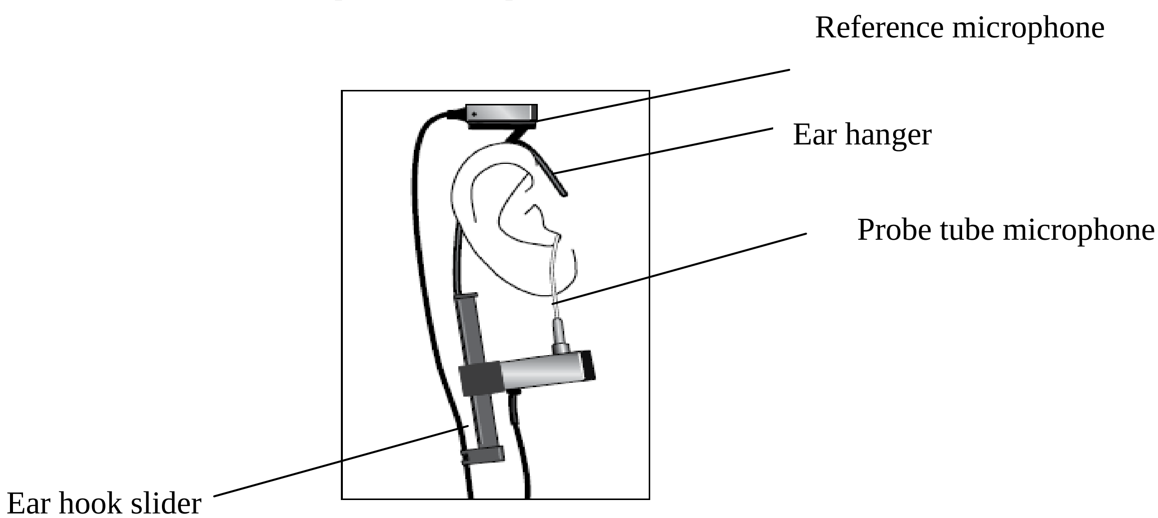


Figure 3.3: Placement of the reference and probe microphones in the ear

c) Inserting the probe tube.

1) The real ear system was calibrated through the ipsilateral comparison procedure.

2) The ear mould was placed next to the probe tube, so that the tube rested along the bottom of the canal part of the ear mould, with the tube extending at least 5 mm from the tip of the ear canal opening.

3) Using the marker, the probe tube was marked where it meets the outside surface of the ear mould.

4) When the probe tube was placed in the ear, it was taken care to see to it that the marking was at the tragal notch of the ear of the participant. Figure 3 shows the marking of the probe tube for aided real ear measurement.

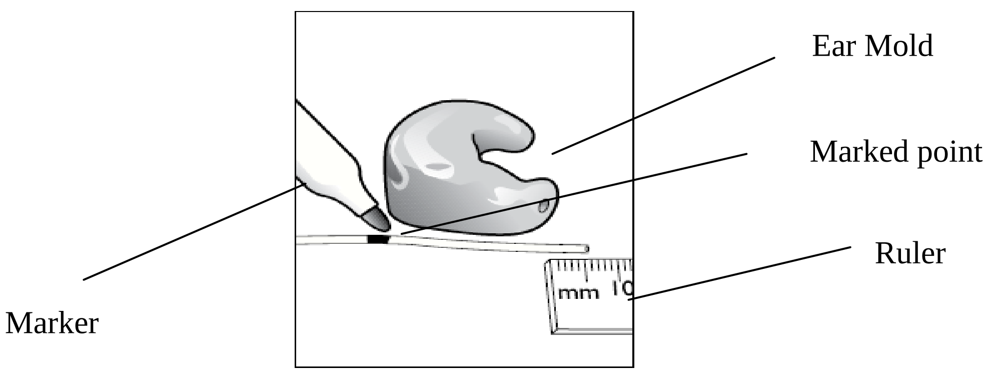


Figure 3.4: Marking the probe tube for insertion in the ear canal

5) At this point, an adhesive tape was used to secure the position of the probe tube, so that head movement of the children did not displace the probe tube.

6) For the aided testing, length of the canal portion of the custom ear mould in addition to a length of 5 mm was considered as the marking point on the probe tube. This point was made to coincide at the tragal notch of the participants.

d) Real Ear measurement procedure.

1) The first step done in the process of real ear measurement was levelling the sound field speaker. This was important so that the input to the hearing aid was controlled evenly across the frequency spectrum. The sound field was levelled by keeping the probe tube near the ear canal. The next step was taken up when the levelling status was ensured.

2) A digi-speech signal was used at 65 dB SPL for the measurement of Real Ear Unaided Response (REUR) which is defined as the SPL measured as a function of frequency, at a specified measurement point in the ear canal, for a specified sound field, with the ear canal unoccluded (ANSI S3.46-1997). This measure gave an estimate of the ear canal resonance characteristics. The response curve giving the intensity levels across the frequency range from 200 Hz to 8000 Hz, at interval of 100 Hz, was obtained which indicated the real ear unaided response (REUR) curve.

3) Hearing aid was fitted with custom ear moulds. The NAL-NL1 fitting formula was used to prescribe the hearing aid gain and accordingly hearing aid was programmed to match the target gain. During this process, if there was occurrence of feedback, the volume control was reduced to a level in which there was no feedback with the programmed gain. Thus, the real ear measurements were obtained at the reduced volume control setting in the ‘feedback reduction - off’ condition. The Real Ear Aided Response was measured at this setting of the hearing aid. Further, closed-set Speech Identification Score (SIS) was obtained with the volume control setting.

4) Similarly, the hearing aid was programmed to reach the target gain with the ‘feedback reduction - on’ condition. The volume control was set to the optimum setting i.e., to a point where there was feedback. The Real Ear Aided Response (REAR) was measured with this setting. With the same volume control setting, the SIS was also obtained from the participant.

5) All the participants were tested with the ‘Active Feedback Intercept - on’ and in ‘Active Feedback Intercept - off’ conditions.

6) A damper of resistance 4700 Ω was inserted into the tubing of the hearing aid at a distance of 9 mm from the tubing end. This amount and placement of the damper was used assuming better effects of smoothening of the mid frequency response which would in turn reduce the peaks caused due to feedback. The damper was placed as close as possible to the earhook end of the hearing aid. With the use of dampers, the volume control setting that was attainable without the occurrence of feedback was noted and Real Ear Aided Response (REAR) was obtained with this volume setting without activation of feedback management method. With the same volume control setting, the SIS was obtained from the participant in the closed-set condition.

7) The Real Ear insertion Gain (REIG) was found which is the difference between the Real Ear Aided Response (REAR) and the Real Ear Unaided Response (REUR) across all the frequencies for the three aided conditions, namely with and without feedback management and with damper for each test ear.

8) Two other measures were included which were calculated based on the real ear gain obtained at different frequencies. The two measures were the High Frequency Average Real Ear Insertion Gain (HFAREIG in dB) and Added Stable Gain (ASG in dB).

9) HFAREIG was obtained by averaging the gain values across the speech frequencies, i.e., 1000 Hz, 1600 Hz and 2500 Hz. It was calculated with the assumption that the high average gave a better estimate of speech perception abilities than other frequency averages (Lenzen, 2008).

10) Added Stable Gain (ASG in dB) was calculated by subtracting the REAG in 'without feedback management' condition from REAG in 'with feedback management condition' i.e., REAG (WFBM) - REAG (WOFBM) or subtracting REAG in 'without feedback management' condition from REAG in 'with damper' condition i.e., REAG (WDAMP) - REAG (WOFBM). This measure was obtained as it could be used as a quantitative measure to compare the benefit from different feedback management in comparison with no feedback management. Also, the effect of the available gain on the improvement in speech identification scores could be quantified.

3.4.3 Phase III: Behavioural Testing

The closed-set Speech Identification Scores (SIS) in quiet were obtained through monitored live voice presentation of the phonemically balanced word lists for children developed by Vandana and Yathiraj (1998). For this the gain was set at just below the level causing feedback. The presentation level of monitored live voice speech was 45 dBHL through. The stimuli were presented through a loudspeaker of the audiometer from 0^0 Azimuth placed at a distance of 1 meter from the head of the participant. The response mode was pointing to the appropriate picture out of a group of 4 pictures. A total of 5 practice trials were given to the participants before starting the actual testing. The scoring was done based on number of words correctly identified out of the total number of 25 words presented. This was followed by the administration of the test to obtain speech identification scores with the feedback suppression algorithm activated and subsequently with the use of dampers with appropriately adjusted volume control settings.

Thus, a total of two sets of measurements (insertion gain measures and behavioural measures) for the two ears of each participant were made for three aided conditions (WOFBM, WFBM and WDAMP) giving a total of six measures for each test ear of the participant considered for the study.

For each participant, the following data were collected:

1) Real Ear Aided Response (REAR, in dB SPL)
2) High Frequency Average Real Ear Gain (HFAREIG, in dB)
3) Added Stable Gain (ASG, in dB)
4) Aided Speech Identification Scores (SIS, maximum score being 25).

3.4 Statistical Analysis

Appropriate statistical analysis was carried out for the data to verify the objectives of the study. The mean and standard deviation of the REAR (in dB SPL), HFAREIG (in dB), Added Stable Gain (ASG in dB) and SIS (with maximum score of 25) were obtained. There were three independent variables in the present study namely the three conditions ‘without feedback management method (WOFBM), ‘with feedback management method (WFBM) and ‘the use of dampers (WDAMP)’. The dependent variables in the present study were Real Ear Aided Response (dB SPL), High Frequency Average Real Ear Insertion Gain (HFAREIG), Added Stable Gain (ASG) and Speech Identification Scores (SIS). The scores obtained using REAR (in dB SPL), Added Stable Gain (ASG), HFAREIG (in dB) were compared across the three aided conditions (WOFBM, WFBM and WDAMP) and were compared across eleven discrete frequencies from 200 Hz to 8000 Hz. Data on SIS were compared across the three aided conditions (WOFBM, WFBM and WDAMP) to check for the significant differences, if any.

CHAPTER 4

Results and Discussion

The study aimed at evaluating the efficacy of two different feedback management methods in hearing aids, namely the feedback reduction algorithm (i.e., phase cancellation) and the use of acoustic modification (i.e., damper in ear mould) in hearing aids. The effect of these two feedback management methods namely the phase cancellation method and the use of damper in the ear mould, on the insertion gain measures and speech identification scores (SIS) was evaluated in the current study.

The data for the present study were collected from 60 ears (N=60 ears) of thirty children having bilateral severe to profound hearing loss. The following insertion gain and behavioural data were evaluated and compared

4.1 Data from insertion gain measurements:

4.1.1 Comparison of Real Ear Aided Response (in dB SPL) obtained for eleven frequencies (from 200 Hz to 8000 Hz) in the three aided conditions.

4.1.2 Comparison of the frequencies High Frequency Average Real Ear Insertion Gain (HFAREIG in dB) for 1 kHz, 1.6 kHz and 2.5 kHz in the three aided conditions.

4.1.3 Comparison of Added Stable Gain (ASG in dB) for the three aided conditions for eleven frequencies (from 200 Hz to 8000 Hz).

4.2. Data from behavioural measurement:

4.2.1 Comparison of Speech Identification Scores in the three aided conditions.

These data were collected in three aided conditions -

a) Hearing aid, without feedback management (WOFBM),

b) Hearing aid, with feedback management (WFBM), and

c) Hearing aid with feedback management deactivated, use of damper in the ear mould (WDAMP).

The data collected were tabulated and subjected to statistical analysis using Statistical Package for Social Sciences (SPSS 17.0 for windows version). Descriptive statistics and analysis of variance were computed to evaluate the objectives of the study. The results are discussed under the following headings:

4.1 Insertion Gain Measures and

4.2 Behavioural Measure

4.1 Insertion gain measures

The data on insertion gain measure obtained for all the 60 ears were analyzed in the three aided conditions, viz., without feedback management with the feedback management and by inserting the damper in the ear mould.

4.1.1 Real Ear Aided Response, REAR (in dB SPL).

Real Ear Aided Response (in dB SPL) was obtained at eleven discrete frequencies across the frequency range from 200 Hz to 8000 Hz, for 60 ears in three aided conditions. Descriptive statistics was used to compare the mean and standard deviation measures of the REAR values (Table 4.1) in the three aided conditions.

Table 4.1:

Mean and Standard Deviation (SD)values of REAR (in dB SPL) for the three aided conditions (WOFBM, WFBM and WDAMP) across eleven discrete frequencies from 200 Hz to 8000 Hz (N=60 ears).

	REAR in dB SPL										
Conditions	**200 Hz Mean (SD)**	**500 Hz Mean (SD)**	**1000 Hz Mean (SD)**	**1500 Hz Mean (SD)**	**2000 Hz Mean (SD)**	**3000 Hz Mean (SD)**	**4000 Hz Mean (SD)**	**5000 Hz Mean (SD)**	**6000 Hz Mean (SD)**	**7000 Hz Mean (SD)**	**8000 Hz Mean (SD)**
WOFBM	87.42 (5.76)	95.98 (6.71)	102.99 (6.23)	102.02 (7.90)	103.54 (6.73)	93.45 (5.97)	87.21 (7.22)	80.89 (8.51)	73.15 (9.07)	67.96 (9.76)	56.79 (7.96)
WFBM	95.55 (5.69)	105.24 (5.57)	110.14 (6.23)	109.15 (5.71)	111.99 (5.72)	101.26 (5.26)	96.61 (5.89)	92.51 (7.46)	84.20 (7.43)	77.32 (10.13)	68.44 (9.88)
WDAMP	90.43 (6.72)	96.99 (6.01)	100.14 (6.15)	106.24 (6.44)	107.71 (6.50)	94.88 (6.20)	90.95 (6.99)	83.41 (8.41)	78.68 (9.12)	67.59 (10.56)	61.13 (10.8)

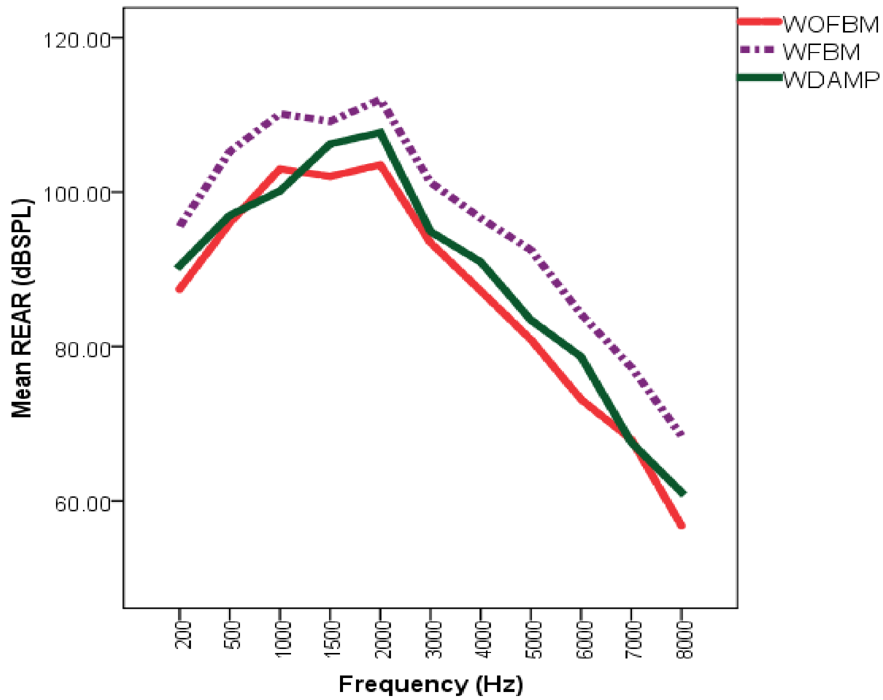


Figure 4.1: Mean REAR (in dB SPL) for the three aided conditions (WOFBM, WFBM and WDAMP).

Figure 4.1 shows the mean REAR values across the eleven frequencies for the three conditions (WOFBM, WFBM and WDAMP). From Table 4.1 and Figure 4.1, it is evident that the mean REAR values was highest for WFBM condition followed by WDAMP condition and then by WOFBM condition, across all the frequencies except for 1000 Hz and 7000 Hz. Further Table 4.2 shows the mean REAR values of different frequencies in the three aided conditions.

Table 4.2:

Results of descriptive statistics for Rear Ear Aided Response (REAR in dB SPL) indicating mean and SD values across the three aided conditions (WOFBM, WFBM and WDAMP).

Conditions	**REAR (in dB SPL)** **Mean (across frequencies)** **(Standard Deviation)**
Without feedback management (WOFBM)	86.49 (15.42)
With feedback management (WFBM)	95.67 (14.14)
With damper (WDAMP)	88.92 (14.99)

As indicated in Table 4.2, a high mean REAR value was evidenced in with feedback management condition (WFBM) compared to without feedback management (WOFBM) and with damper (WDAMP) condition. To determine if this difference in REAR in the three aided conditions was significant, repeated measure ANOVA was done. Table 4.3 shows the results of two way repeated measure ANOVA indicating F value with degrees of freedom and level of significance.

Table 4.3:

Results of two-way repeated measure ANOVA showing the F value, degrees of freedom (df) and level of significance (p) of REAR at different frequencies and conditions

Parameters	***F value***	***df***	***p***
Conditions	113.550	2	0.000*

Note: *: p<0.01=highly significant difference

As depicted in the Table 4.3, a highly significant difference resulted on repeated measure ANOVA ($p<0.01$). Hence, Bonferroni's multiple group comparison was carried out to evaluate the pairs of conditions which showed a significant difference. Table 4.4 shows the results of Bonferroni multiple group comparison showing the level of significance for the three pairs of conditions.

Table 4.4:

Results of Bonferroni multiple group comparison showing the level of significance across the pairs of conditions

Difference between conditions	**p values**
WOFBM & WFBM	0.000*
WFBM & WDAMP	0.000*
WDAMP & WOFBM	0.025*

Note: *: p < 0.01=highly significant difference

As indicated in Table 4.4, there was a highly significant difference between the WOFBM and WFBM conditions, and WFBM and WDAMP ($p<0.01$). In addition, significant difference was also noted between WDAMP and WOFBM ($p<0.05$).

The observed differences between the conditions along with WFBM having a significantly greater output compared to other two conditions (WOFBM and WDAMP) may be attributed to a greater available gain and hence a greater output was possible with the activation of the feedback management (Freed & Soli, 2006). These differences in the REAR is possible due to the use of digital technology in hearing aids, because of which mathematical estimations of the feedback path can be made and used to compensate for the feedback, essentially without affecting the input signal, while ideally preserving the desired output. Such a method provides an added 6 to 10 dB average headroom improvement and possibly avails more useable gain compared to without the feedback management activated (Edwards, 2000; Olson, Musch, & Struck, 2001).

The type of feedback management method used in the hearing aid in the present study was 'Active Feedback Intercept' which is based on the principle of working of phase cancellation algorithms. Since phase cancellation algorithms simply cancel out unwanted feedback, there is no gain reduction associated with elimination of feedback. This simply involves subtracting an artificially generated signal which is opposite in phase but having the same amplitude and spectral characteristics as the feedback signal. On the contrary, this technique results in added stable gain (ASG) - i.e., an increase in maximum gain with feedback management enabled compared to that with feedback management disabled, which is why there was an increase in the output with the activation of feedback management method (Freed & Soli, 2006; Kates, 2001; Merks, Banerjee, & Trine, 2006).

Mean REAR values were higher in aided condition with dampers for all the frequencies compared to 'without feedback management' method. This increase in the REAR values (and thus the real ear insertion gain values) may be because, the dampers give a higher gain at higher frequencies and smoothen the frequency response at mid- to high- frequencies where the resonances caused by feedback results in sharp peaks in the frequency response (Valente, 1984). Moreover, in the WOFBM method, the prescribed gain is reduced till the point where feedback does not occur, the gain across all the frequencies will be effectively lesser compared to WDAMP condition. Hence, a reduced output and thus the gain is the most likely outcome in the Aided condition without feedback management. However, the dampers reduce the sharp peaks caused due to the feedback and hence they reduce the gain and output SPL especially at low- to mid- frequencies (Valente, 1984). Hence, there are reduced REAR values in WDAMP condition compared to WFBM condition.

Also, frequency differences were noted for REAR (in dB SPL) across the three aided conditions. There was a greater output at mid- and high- frequencies from 500 Hz to 4000 Hz (with an average REAR of 105.73 dB SPL) compared to lower frequencies below 500 Hz (with an REAR of 95.55 dB SPL). Because the frequency response becomes sharper with the occurrence of feedback, there may be a significant difference of REAR across the frequencies in the without feedback management condition (Valente, 1984).

Because the feedback management methods provide added stable gain, the frequency response also changes accordingly across the two conditions namely with and without feedback management conditions (Freed & Soli, 2006; Kates, 2001; Merks et al., 2006). Since the phase cancellation method, is functional at 1500 Hz to 6000 Hz and more effectively it operates at 3000 Hz to 6000 Hz (Dyrlund, Henningsen, Bisgaard, & Jensen, 1994), the peaks caused due to the presence of feedback are reduced. And this explains the observed differences in REAR values across the conditions for different frequencies. According to Valente (1984), there will be smoothening of the frequency response with the use of dampers. As a result, there will be a change in the frequency response with the use of dampers compared to WOFBM and WFBM, hence a significant difference is expected across the three conditions.

Moreover, there are only certain frequencies where the feedback management method operates (Dyrlund et.al., 1994) and only certain frequency range which is smoothened by dampers (Valente, 1984). In addition, the frequency range where the feedback management method and dampers operate, may be different, that is there may be a difference in the REAR values across the frequency range.

A similar finding was noted in a study by Kuk and Ludvigsen (2002), who reported an increase in the available gain and hence the output across the frequency range of 200 Hz to 8000 Hz with the phase cancellation method. According to Robert (2006), phase cancellation not only preserves gain, but also because of its increased feedback margin, makes approximately 10 to 15 dB SPL more amplification available in the mid to-high frequencies.

However, in a study by Lenzen (2008) the mean ASG ranged from 1.6 dB for low frequency band to 2.8 dB for the high frequency band. The mean difference of 1.2 dB in ASG between the low frequency band and the mid-frequency band was statistically significant (two tailed proportion $p<0.001$). Also, it was reported that the mean difference of 0.9 dB in ASG between the mid-frequency band and the high frequency band was statistically significant (two tailed proportion ($p<0.001$). There were no significant differences in mean ASG between the low frequency and high frequency band. The reason attributed to the reduced added stable gain values was that the maximum gain was reached initially and hence further improvement was not effective due to the 'ceiling effect'.

The ear mould dampers have an effect of smoothening the peaks from 1000 Hz to 3000 Hz (Taylor & Teter, 2009). As a result of this, the peaks are reduced and a smoother frequency response with higher gain is possible at mid frequencies (Dillon, 2001).

4.1.2 High Frequency Average Real Ear Insertion Gain (HFAREIG) calculated for the frequencies 1 kHz, 1.6 kHz and 2.5 kHz in all the three conditions.

This data was obtained for all the three conditions across the frequencies 1 kHz, 1.6 kHz and 2.5 kHz for 60 ears. Table 4.5 shows the descriptive statistics for the HFAREIG values at 1000 Hz, 1600 Hz and 2500 Hz obtained across the three conditions (WOFBM, WFBM and WDAMP).

Table 4.5:

Descriptive statistics showing Mean and Standard Deviation for the HFAREIG values across the three conditions (WOFBM, WFBM and WDAMP)

Conditions	**HFAREIG (in dB): Mean (Standard Deviation)**
Without feedback management	34.99 (6.09)
With feedback management	42.04 (5.25)
With damper	37.69 (4.34)

Table 4.5 indicates that the mean HFAREIG value for WFBM condition is greater than mean HFAREIG values for WOFBM and WDAMP conditions. Repeated measure ANOVA was done to find the significant differences across conditions (WOFBM, WFBM and WDAMP), if any. Table 4.6 shows the results of repeated measure ANOVA with F value, degrees of freedom and level of significance.

Table 4.6:

Results of repeated measure ANOVA for HFAREIG with F value, degrees of freedom with error degrees of freedom and level of significance

Parameter	**F value (Degrees of freedom, error degrees of freedom)**	**Level of significance (p)**
HFAREIG	55.92 (2, 118)	0.01*

Note: **: $p < 0.01$=highly significant difference

As indicated in Table 4.6, a highly significant difference was evidenced for HFAREIG across the three conditions [F (2,118) =55.92, $p < 0.01$]. Bonferroni's pair-wise comparison was done to assess the significant differences between the conditions for HFAREIG. Table 4.7 indicates results of Bonferroni's pair-wise comparison done across the three conditions for HFAREIG.

Table 4.7:

Results of Bonferroni's pair-wise comparison for HFAREIG across the three conditions (WOFBM, WFBM and WDAMP)

Conditions HFAREIG (in dB)	**Level of Significance (p)**
WOFBM & WFBM	0.000*
WFBM & WDAMP	0.000*
WDAMP & WOFBM	0.002*

Note: *: $p < 0.01$=high significant difference

A highly significant difference was present across the three pairs of conditions for HFAREIG values as revealed from Table 4.7. The significant difference across the conditions may be attributed to the frequency range at which the feedback management is functional. It is supported by the fact that most of the feedback management is activated at frequencies between 1500 Hz to 6000 Hz. Accordingly, there would be a gain enhancement in this frequency range. This reason can be attributed to the observed differences between the HFAREIG values across WOFBM and WFBM conditions. Also, study by Flynn and Flynn (2006), showed a greater available gain with feedback management strategy at frequencies from 1.5 kHz to 3 kHz which is why a significant difference between the two conditions might have resulted.

The present study shows an increase in HFAREIG of 7 dB with and without feedback management conditions supports the findings of the study by Merks et al.

(2006). He compared the feedback reduction performance of two hearing aids on 20 ears. Maximum Stable Gain was calculated by averaging the gain at 1 kHz, 1.6 kHz, and 2.5 kHz. The authors found that ASG ranged between 9 and 12 dB across the two hearing aids. The average difference in ASG between the two study hearing aids was 3 dB (Lenzen, 2008).

The difference in HFAREIG values for WOFBM and WDAMP conditions was 2.7 dB. There was a high significant difference between WOFBM and WDAMP conditions which may be because dampers decrease the gain and the maximum output (Valente, 1984). Since, they are more effective in reducing the peaks from 1 to 3 kHz, there will be reduction in the gain at this frequency range, compared to that obtained from WFBM condition. However, the gain reduction with the use of 'yellow' color coded dampers was on an average 9 dB (Valente 1984), which was mostly lesser compared to the gain reduction caused in an attempt to reduce the feedback in WOFBM condition.

4.1.3 Added Stable Gain (ASG) across the frequency range from 200 to 8000 Hz (for eleven discrete frequencies.

Added Stable Gain (ASG) which is the difference between the gain for the WOFBM and WFBM condition or WOFBM and WDAMP condition was calculated for the two conditions (WFBM and WDAMP). Since the ASG gives an idea of an increase in the available gain with feedback management and with dampers, comparison of the obtained ASG was made to account for the efficacy of the feedback management methods. Table 4.8 shows the ASG for the two conditions (WFBM and WDAMP) across the eleven frequencies.

Table 4.8:

Mean ASG (in dB) across eleven discrete frequencies from 200 Hz to 8000 Hz for WFBM and WDAMP conditions

Mean ASG for conditions (in dB)	**Frequencies (in Hz)**										
	200	**500**	**1000**	**1500**	**2000**	**3000**	**4000**	**5000**	**6000**	**7000**	**8000**
WFBM	*8.13*	*9.26*	*7.14*	*8.44*	*7.80*	*9.40*	*11.61*	*11.04*	*11.04*	*9.36*	*11.64*
WDAMP	*3.01*	*1.01*	*-2.85*	*4.16*	*1.42*	*3.74*	*2.52*	*3.52*	*3.52*	*-0.37*	*4.33*

Table 4.8 indicates that the mean ASG for WFBM condition was greater than WDAMP condition across all the frequencies. Also, mean ASG values were found to be greater for higher frequencies compared to mid- and low- frequencies. To find the average value of ASG across the frequencies, descriptive statistics was used. Table 4.10 shows the results of descriptive statistics giving mean, median and standard deviation for the ASG averaged across the eleven frequencies.

Table 4.9:

Results of descriptive statistics for Added Stable Gain (ASG) indicating mean, median and SD values for the two conditions (WFBM and WDAMP)

Conditions	**ASG (in dB): Mean (Standard Deviation)**
With feedback management	9.17 (1.65)
With damper	2.24 (2.26)

From Table 4.9, it is evident that the mean values for WFBM condition were greater than WDAMP. A difference of 6.0 to 7.0 dB was evidenced for ASG across the two conditions. Figure 4.2 shows the gain (in dB) across the frequencies for the three conditions (WOFBM, WFBM and WDAMP).

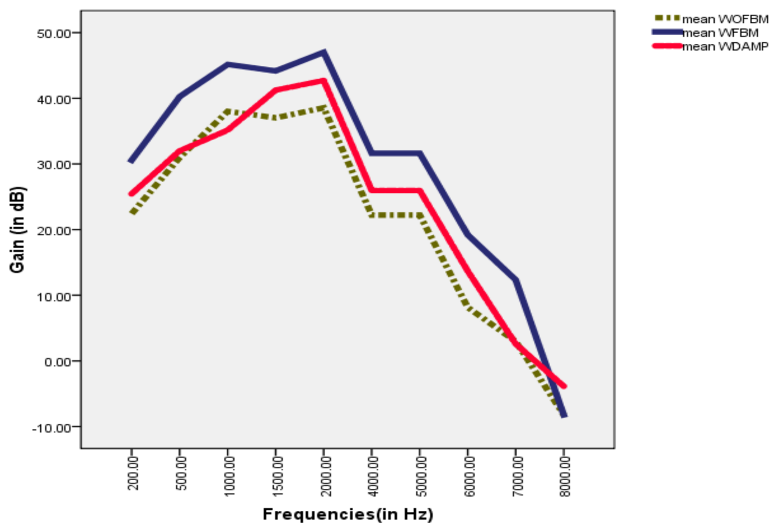


Figure 4.2: Gain (in dB) across the frequency range from 200 Hz to 8000 Hz for three conditions (WOFBM, WFBM and WDAMP)

Figure 4.2 indicates that the gain for WFBM was greater compared to without feedback management and with damper. This finding was evident across the frequency range from 200 Hz to 8000 Hz. Significant differences between the two conditions were determined using pair-wise comparison, if indicated. Table 4.10 indicates results of paired t-test.

Table 4.10:

Results of paired t-test showing t value, degrees of freedom and level of significance

Parameter	**t- value (Degrees of freedom)**	**Level of sig.2-tailed (p)**
ASG (WFBM & WDAMP)	10.035 (10)	0.000*

Note:: p< 0.01=highly significant difference*

Table 4.10 is indicative of a highly significant difference between the two conditions for ASG ($p < 0.01$). Several studies have reported the ASG values obtained with and without feedback management methods. The increase in ASG values with feedback management method can be attributed to the principle of working of the phase cancellation, which effectively reduces the feedback without reducing the gain. Moreover, it gives a greater available gain across the frequency range. Mean ASG values obtained through the method of feedback management varies across the frequencies. Few studies reporting the amount of ASG and ASG values across the frequency range are discussed below.

According to Robert (2006), it was seen that the ASG was maximum at around 1 kHz and 3 kHz followed by higher frequencies. In addition, the study done by Robert (2006) further emphasized a greater ASG at higher frequencies (from 2 k to 5 kHz). Approximately 10 to 15 dB more amplification was made available in the mid-high frequencies through the phase cancellation method. According to Maxwell and Zurek (1995), the maximum added wideband stable gain was approximately 12 dB through the method of phase cancellation.

Field trials of a feedback-cancellation system built into a BTE hearing aid have shown increases of 8 to 10 dB in the gain used by individuals with severe hearing impairment (Bisgaard, 1993) and increases of 10 to 13 dB in the gain margin measured in real ears (Dyrlund et al., 1994). Computer simulations and prototype digital systems indicate that increases in gain of between 6 and 20 dB can be achieved in an adaptive system before the onset of oscillation, and no loss of high-frequency response is noted (Kates, 1991; Engebretson & St.George, 1993; Kaelin, Lindgren, & Wyrsch, 1998).

Greenberg, Zurek, and Brantley (2000) reported that ASG ranged between -1 to 25 dB with a mean ASG of 8.5 dB for the experimental algorithm and approximately 5 dB for the other algorithms. Banerjee, Recker, and Paumen (2006) compared the feedback reduction performance of two hearing aids on 20 ears. Gain for each hearing aid was increased in 1 dB steps until the hearing aid was just below audible feedback. At this point, real ear aided gain (REAG) was measured using a 60 dB SPL composite noise signal with the feedback reduction algorithm disabled and enabled. Maximum stable gain (MSG) was calculated by averaging the gain at 1 kHz, 1.6 kHz, and 2.5 kHz. The authors found that the ASG ranged between 9 and 12 dB across the two hearing aids.

However, a study by Lenzen (2008) indicated that the mean ASG ranged from 1.6 dB for low frequency band to 2.8 dB for the high frequency band. A mean difference of 1.2 dB in ASG between the low frequency band and the mid frequency band was statistically significant (two tailed proportion $p < 0.001$). There were no significant differences in mean ASG between the low frequency and high frequency band.

Merks et al., (2006) did not report average ASG, but reported that ASG ranged from 3.5 dB to 16.3 dB. Banerjee et al. (2006) reported that ASG ranged between 2 dB to 18 dB with an average of 9 dB to 12 dB. Freed and Soli (2006) did not report average ASG, but reported that ASG ranged between 0 to 18 dB across all frequencies. Greenberg et al. (2000) reported an ASG ranging between -1 dB to 25 dB with an average ASG of

8.5 dB for one experimental algorithm and approximately 5 dB for the other experimental algorithms.

According to Kuk, Ludvigsen, and Kaulberg (2006), there is a wide range of Added Stable Gain increase between 2 kHz to 4 kHz from as little as 8 dB to as much as 19 dB. An average of 12 to 13 dB was noted for the group. No increase in AGBF was noted below 1 kHz, possibly because feedback being a high frequency phenomenon usually occurs above 1 kHz and target gain is typically reached below 1 kHz. Thus, there is probably no need for gain increase.

Thus, the mean ASG values obtained in the present study was in accordance with the previous reports. However, the ASG obtained for the WDAMP condition was lesser compared with WFBM condition. This may be because of dampening effect from 1 to 3 k Hz which might have resulted in lower gain compared to the gain from WFBM condition.

4.2. Behavioural measure

4.2.1 SIS values for the three conditions (WOFBM, WFBM and WDAMP).

The mean and standard deviation values of the Speech Identification Scores for each participant in three aided conditions were obtained. The maximum SIS was 25. Table 4.11 gives the mean and SD of the SIS.

Table 4.11:

Mean and Standard Deviation (SD) values of SIS (Max. 25) in three aided conditions (WOFBM, WFBM and WDAMP)

Conditions	**SIS - Mean (Standard Deviation)**
Without feedback management	17.45 (2.15)
With feedback management	22.25 (2.39)
With damper	20.78 (2.54)

From Table 4.11, it is evident that the mean SIS for WFBM condition was greater than the SIS values for WOFBM and WDAMP conditions. It was found that the SIS was better in the WFBM condition than WOFBM and WDAMP conditions. Figure 4.4 shows the mean SIS scores with standard error (95% confidence level) across the three conditions.

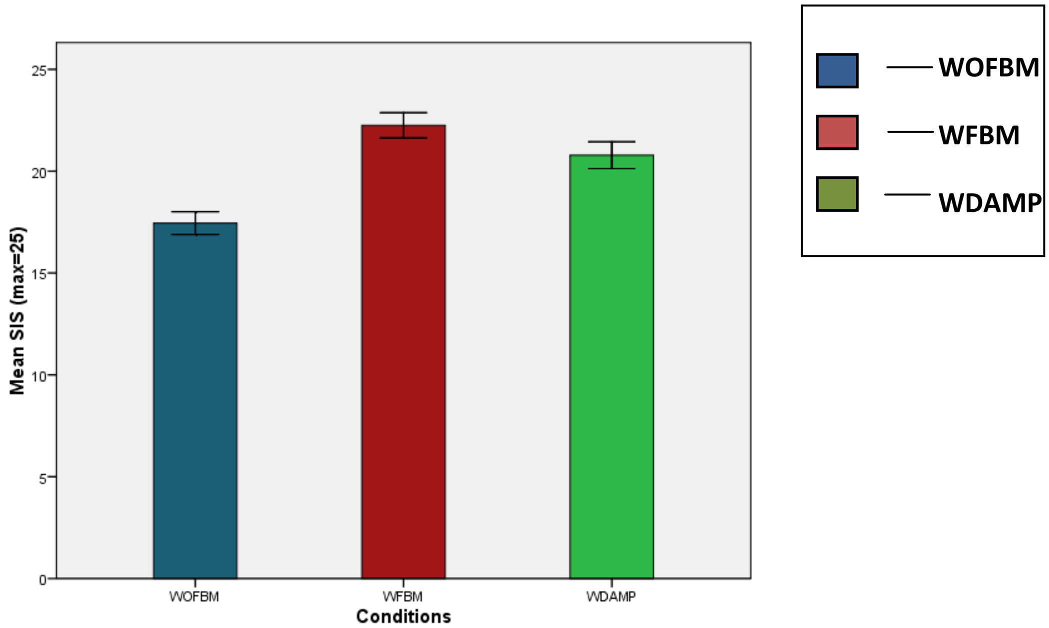


Figure 4.4: Mean SIS (Max.=25) for 60 ears across the three conditions (WOFBM, WFBM and WDAMP) (two-tailed with 95% confidence level)

Figure 4.4 shows the mean SIS values across the three aided conditions. SIS scores WFBM was greater compared to SIS in the WDAMP and WOFBM conditions. In order to see if there was a significant difference, repeated measure ANOVA was done. Table 4.12 shows the results of repeated measure ANOVA with F value, degrees of freedom and level of significance.

Table 4.12:

Results of repeated measure ANOVA with F value, degrees of freedom and level of significance for SIS across the three conditions (WOFBM, WFBM and WDAMP)

Parameter	**F value (Degrees of freedom, error degrees of freedom)**	**Level of significance (p)**
SIS (max value=25)	275.851 (2, 118)	0.000*

Note:*: $p < 0.01$=high significant difference

From Table 4.12, a high significant difference between the conditions is evident. Bonferroni's pair-wise comparison was done to find the pair of conditions which showed a significant difference. Table 4.13 shows the results of Bonferroni's pair-wise comparison with significance levels.

Table 4.13:

Results of Bonferroni's pair-wise comparison for SIS across three conditions (WOFBM, WFBM and WDAMP)

SIS scores across conditions	**Level of Significance (p)**
WOFBM & WFBM	0.000*
WFBM & WDAMP	0.000*
WDAMP & WOFBM	0.000*

Note:*: $p < 0.01$=high significant difference

Table 4.13 revealed that a highly significant difference existed across the conditions on SIS. This can be attributed to the added stable gain which was more in WOFBM condition than for the WFBM than WDAMP condition. Nevertheless, the increase in ASG allows the majority of wearers to achieve their desired gain without feedback in many more listening situations. The ability to use the target gain more consistently could result in better speech intelligibility, better sound quality, and a hassle-free listening experience. In addition, the HFAREIG values were 7 dB greater for WFBM condition and 4.35 dB greater for WDAMP condition compared to WOFBM condition. The increase in the available gain at higher frequencies leads to better speech perception.

Dyrlund et al (1994) reported that since there is a greater ASG at high frequencies, a tremendous improvement in speech identification performance without causing feedback is evidenced. Since feedback is a high frequency phenomenon, phase cancellation method would result in more of high frequency gain by cancelling the peaky responses (Martin & Robert, 2006). Christensen, Winfrey, and Stelmachowitz (2006) investigated the effectiveness of phase cancellation method and it was noted that a high frequency gain and improved perception of high frequency consonants resulted. The authors concluded that using feedback management helps to meet the mid- and high-frequency targets while providing maximum audibility for speech sounds. Hence, an improvement in SIS was noted in the WFBM condition. Figure 4.4 shows the mean SIS scores (with max. score of 25) across the conditions WOFBM, WFBM and WDAMP.

The SIS in WDAMP condition was better than WOFBM condition. Peaks and troughs in the gain frequency response adversely affect the speech intelligibility and quality of amplified sound. Peaks become objectionable if they rise above 6 dB above the smooth curve joining the dips, which affects the maximum output of the hearing aid also the gain frequency response (Valente, 1984). Since smoothening of the frequency response takes place with the use of dampers and the peaks in the frequency response are reduced, an improvement in SIS was seen in WDAMP condition in comparison with WOFBM condition. However, the SIS obtained in WFBM condition was significantly better than the SIS obtained in WDAMP condition. This may be attributed to the added stable gain possible with the phase cancellation method where as in WDAMP condition, only the peaked responses are reduced providing a more stable output, without providing the ASG equivalent to that of the WFBM condition. This is expected since the purpose of the two is different. Hence, a significant difference was noted across the two conditions (WFBM and WDAMP).

To summarize the results of the study:

1) Data were obtained on the insertion gain measures and behavioural measures across all the participants in three aided conditions (WOFBM, WFBM and WDAMP) across eleven discrete frequencies (form 200 Hz to 8000 Hz).
2) Data on insertion gain measures included comparison of data on Real Ear Aided Response (REAR), High Frequency Average Real Ear Insertion Gain (HFAREIG in dB) and Added Stable Gain (ASG in dB) across the three aided conditions.

3) Data on behavioural measure included comparison of data on Speech Identification Scores (SIS) across the three aided conditions.
4) Analysis of the data obtained on REAR across the three conditions showed a high significant difference between the WOFBM and WFBM, also for WFBM and WDAMP ($p<0.01$). This may be attributed to the principle of working of phase cancellation method which effectively increases the amount of gain available and also the output with the activation of feedback management. With the use of dampers, the gain and hence the output across the mid frequencies is likely to be reduced due to the dampening effect. Hence, a significant difference may have resulted for REAR values across WFBM and WDAMP condition ($p<0.01$). The gain and hence the REAR values are reduced to the point till the feedback does not occur, hence WOFBM condition has significantly poorer score on REAR compared to WDAMP condition.
5) HFAREIG values were significantly higher for WFBM condition compared to WOFBM and WDAMP conditions ($p<0.01$). Since feedback is a high frequency phenomenon, the action of phase cancellation makes more gain available at high frequencies. WDAMP condition had a significantly higher HFAREIG values compared to WOFBM condition ($p<0.01$). As the dampers increase more of high frequency gain, the observed differences can be well explained.
6) ASG was significantly better for WFBM compared to WOFBM and WDAMP conditions. ($p<0.01$). This is possibly because of working of phase cancellation method which increases the available gain before the occurrence of feedback.

ASG for WDAMP condition was significantly higher compared to WOFBM, because of the enhanced high frequency gain availability with the use of dampers.

7) Speech Identification Scores (SIS) showed a significant improvement in WFBM condition compared to WOFBM and WDAMP conditions ($p<0.01$). Since the HFAREIG and ASG for the WFBM condition were significantly more than WOFBM and WDAMP conditions, there might be an enhancement in the SIS for WFBM condition. SIS in WDAMP condition was significantly higher than the SIS in WOFBM condition ($p<0.01$). This could be due to the significantly higher HFAREIG and ASG values for WDAMP compared to WOFBM ($p<0.01$).

Thus, the improvement on insertion gains measures with the activation of feedback management methods namely the feedback reduction strategy and the use of dampers, leads to a parallel improvement in terms of perceptual measures as well.

CHAPTER 5

Summary and Conclusions

The aim of the present study was to evaluate the efficacy of the feedback reduction methods in the hearing aid, namely the phase cancellation algorithm and the use of acoustic modification (damper) in the ear mould. The specific objectives were:

1) To evaluate the effect of the phase cancellation method and the use of damper in ear mould, on the insertion gain measures.

2) To evaluate the effect of the phase cancellation method and use of damper in the ear mould, on speech identification scores (SIS).

The following method was adopted to investigate the objectives of the present study.

The data for the present study were collected from 60 ears of 30 children having bilateral severe to profound hearing loss (N=60 ears). The insertion gain and behavioural measures were evaluated and compared in the following three aided conditions:

a) With the hearing aid deactivated for the feedback management method (WOFBM)

b) With the hearing aid activated for feedback management method, and (WFBM)

c) With the use of damper in the ear mould (WDAMP)

The following insertion gain and the behavioural data were analysed in the three aided conditions:

I) Data from insertion gain measurements

a) Comparison of Real Ear Aided Response (in dB SPL) obtained for eleven frequencies

(From 200 Hz to 8000 Hz) in the three aided conditions.

b) Comparison of the High Frequency Average Real Ear Insertion Gain (HFAREIG in dB) for 1 kHz, 1.6 kHz and 2.5 kHz in the three aided conditions.

c) Comparison of Added Stable Gain (ASG in dB) for the three aided conditions for eleven frequencies (from 200 Hz to 8000 Hz).

II) Data from behavioural measure in the three aided conditions

Comparison of Speech Identification Scores:

The data collected were tabulated and subjected to statistical analysis using Statistical Package for Social Sciences (SPSS 17.0 for windows version). Descriptive statistics and analysis of variance were computed to evaluate the objectives of the study. To investigate the presence of interaction effect and main effects for the presence of significant differences between the conditions (WOFBM, WFBM and WDAMP) and frequencies (200 Hz to 8000 Hz), two-way repeated measure ANOVA was carried out. Bonferroni's multiple group comparison was carried out to check for the significant differences between the pairs of conditions, if any. The results of the study revealed that:

I. On Comparison of Real Ear Aided measurement obtained for eleven frequencies (from 200 Hz to 8000 Hz) in the three aided conditions, the following were noted:

a) REAR:

REAR was significantly greater in the WFBM condition compared to WOFBM and WDAMP for all the participants ($p < 0.01$), with REAR for the WDAMP condition being significantly higher than in WOFBM condition.

b) High Frequency Average Real Ear Insertion Gain (HFAREIG in dB):

High Frequency Average Real Ear Insertion Gain 1 kHz, 1.6 kHz and 2.5 kHz values were significantly greater for WFBM condition compared to WOFBM and WDAMP condition ($p < 0.01$). The HFAREIG values were significantly greater for WDAMP condition compared to WOFBM condition ($p < 0.01$).

c) Added Stable Gain (ASG in dB):

The difference between the Added Stable Gain for the WOFBM, WFBM and WDAMP conditions were calculated and analyzed. The results revealed that there was a significant difference between the ASG values for WFBM and WDAMP conditions ($p < 0.01$). The ASG with the feedback management condition was significantly greater than that for with dampers condition.

II. Behavioural measure in the three aided conditions - Speech Identification Scores (SIS):

The three conditions were compared for the speech identification scores. Statistical analysis revealed that for all the participants, there was a highly significant difference between the three pairs of conditions.

From the study, it can be concluded that there was a highly significant improvement with the feedback management condition (WFBM) than without the feedback management (WOFBM) in hearing aids. The improvement in the WFBM condition was mostly due to the principle of working of phase cancellation method, because of which gain is not compromised while reducing the feedback. On the contrary, available gain increases with the activation of phase cancellation. Hence, greater output and gain value result with phase cancellation (Freed & Soli, 2006; Kates, 2001; Merks et al., 2006). With dampers, the amount of gain available and hence the output given also is significantly higher because of the effect of smoothening of the peaks in the frequency response. This leads to increased high- and mid- frequency response as the hearing aid wearers can increase the volume control. This is mainly because the peaks in the frequency response are reduced (Valente & Dunn, 2007).

Increase in the Added Stable Gain and high frequency average real ear insertion gain (HFAREIG) led to better speech identification scores in feedback management activated condition than when it was de-activated. However, Dyrlund et al. (1994) reported that since there is greater ASG at high frequencies there is a improvement in speech identification performance without causing feedback. The SIS obtained in the WDAMP condition was better than in WOFBM condition due to increased gain available in mid- and high- frequencies (Valente, & Dunn, 2007), however the amount of gain available was lesser compared to WFBM condition.

5.1 Clinical implications

From the results of the current study, the following clinical implications were arrived at:

1. Real ear measurements such as the REAR, HFAREIG and ASG provide quantitative information on the added benefit obtained from the feedback reduction methods (i.e., the phase cancellation method and the use of damper in the ear mould) evaluated in the study. Thus, these methods can provide optimum gain and added benefit required for clients. Hence, the REAR, HFAREIG and ASG measures can be used as an effective verification tool.

2. Since use of the feedback reduction methods evaluated in the study has demonstrated benefit in both objective and behavioural measures, fitting individuals with severe to profound hearing loss will not pose any problem with reference to reduction in feedback and compromise in the gain provided by the hearing aid.

3. Real ear measures being objective measurement, this can be of help while evaluating young children or difficult-to-test population.

4. Though the damper in the ear mould is traditionally used for smoothening the frequency response of the hearing aid, from this study it has been shown to be advantageous as a feedback reduction strategy. Thus, even if the hearing aid does

not have the option of phase cancellation algorithm, a damper can be used for the purpose.

5. Once the problem of feedback is solved, as documented in the present study, there will be better acceptance of the hearing by the client as well as the family members.

Thus the present study throws light on the usefulness of real ear measures and behavioural measures with feedback reduction strategies.

5.2 Future directions for research

1. The study can be replicated using other feedback reduction algorithms for comparison in terms of the real ear gain and speech perception measures.

2. The quality of speech perceived is an important aspect of patient acceptance of the hearing aid. In the present study, this aspect was not considered for evaluation since the participants were children. The quality assessment with feedback management strategies can be can be done in the adult to further validate the results.

3. Efficiency of the feedback management works is dependent on the environment around the hearing aid wearer. Hence, applicability of feedback management strategy must be tested in adverse environmental conditions.

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