COMPARISON OF FUNCTIONAL GAIN AND INSERTION GAIN

IN LINEAR AND NON-LINEAR HEARING AIDS

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A dissertation submitted in part fulfillment for the degree of
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ALL INDIA INSTITUTE OF SPEECH AND HEARING

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May, 2009

Dedicated to

My Family

&

Manjula Ma'am

Certificate

This is to certify that this Dissertation entitled "Comparison of functional gain and insertion gain in linear and non-linear hearing aids" is the bonafide work submitted in part fulfillment for the degree of Master of Science (Audiology) of the student with Registration No. 07AUD009. 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 Diploma or Degree.

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This is to certify that this Dissertation entitled "Comparison of functional gain and insertion gain in linear and non-linear hearing aids" is the result of my own study under the guidance of Dr. Manjula P., Reader, Department of Audiology, All India Institute of Speech and Hearing, Mysore, and has not been submitted earlier in any other University for the award of any Diploma or Degree.

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INTRODUCTION

Young children listen and learn language both actively and passively in a variety of auditory environments throughout the day (Bess, Dodd-Murphy, & Parker, 1998; Flexer, 1999). For example, in addition to being taught words directly by their parents or siblings, children may learn words through over hearing the conversations of others. Thus, normal hearing is essential for spoken language comprehension in young children. Therefore, children with hearing impairment often need specialized management and intensive instruction for the development of language (Skinner, 1978). Consistent audibility of speech at levels ranging from soft to loud is a necessary pre-requisite for spoken language development; its importance is reflected in the Paediatric Amplification Guidelines (2004) by American Academy of Audiology. These guidelines state that the goal of amplification for children with significant hearing impairment is “to provide a signal that makes low, moderate, and high intensity sounds audible but not uncomfortable and provide excellent sound quality in a variety of listening environments”.

Hearing aids partially overcome the deficits associated with hearing loss. A hearing aid amplifies the weak sounds as well as moderate to loud level of sounds. Selection of hearing aid is a step-wise procedure which involves hearing evaluation as *first* step. The *second* step involves pre-selection hearing aid which in turn depends on the technology (digital, analog, programmable) and the type of hearing aid being selected. The *third* step involves hearing aid fitting based on different approaches, such as threshold based and suprathreshold (most comfortable level) approaches. The *fourth* step involves verification as it is necessary to ensure that the electroacoustic parameters that were selected for the client accurately matches with the device on client's ear. Validation, the *fifth* step, refers to measuring whether the client actually performs well with and likes what was prescribed. The final and *sixth* component of hearing aid selection is counselling on the use and care of hearing aids (Palmer, Lindley, & Mormer, 2000).

The *fourth* step of verification differs for different types of circuitry of the hearing aid, i.e., linear or non-linear. In linear hearing aids, same amount of gain is applied to incoming sounds of a given frequency regardless of the level of sound entering the hearing aid (Palmer, Lindley, & Mormer, 2000).

Up to the maximum output level of the hearing aid, the effect of linear amplification is that a set of volume loud sounds may be uncomfortably loud. In order to overcome this problem most of the individuals are now fitted with non-linear hearing aids which provide more gain for the weak sounds than to the intense sounds. The non-linear hearing aids mainly compress most of the speech spectrum into the residual range, giving increased audibility and comfort and making loudness perception similar to that in individuals with normal hearing.

Currently available measures like functional gain measurements are adequate for the verification of the performance of children through the linear hearing aids (Stelmachowtcz, Hoover, Lewis, & Brennan, 2002). The difference in audiometric dial settings between the unaided and the aided thresholds is defined as the functional gain provided by the hearing aid at a specific frequency (Kuk & Ludvigsen, 2003). The importance of this definition is that the aided threshold is a behavioral response at one perceptual level.

Threshold is a behavioral response that reflects the lowest sound pressure level (SPL) at which a listener barely detects the presence of a sound. In a clinical situation, the unaided threshold represents the lowest dial setting that produces a signal at the eardrum that reaches the threshold criterion. The aided thresholds, represent the lowest dial setting that produces an input (I) to the hearing aid microphone which, when added to its in-situ gain, results in an output (O) at the wearer's eardrum that reaches the threshold criterion (Kuk & Ludvigsen, 2003).

For a given hearing aid gain setting and test environment, there is only one aided threshold and one value of functional gain. It also means that although the wearer may experience a higher hearing aid output in the ear when the input is higher, the aided threshold and thus the functional gain remains the same (Kuk & Ludvigsen, 2003). Despite its usefulness, information provided by sound-field measures is not the same as that provided by probe tube microphone measures, especially with non-linear hearing aids. Thus, these two indices must be determined separately for a complete verification of the wearer's performance with the different hearing aids.

However, as the functional gain (FG) is the measurement done only at one level, it reflects the hearing aid gain only at one input level or at low input levels. Thus, the FG seems to be more appropriate for evaluation of linear hearing aids that give a constant gain irrespective of the level of the input signal (Kuk & Ludvigsen, 2003). For evaluating the non-linear hearing aids, one of the limitations of FG lies in the fact that the FG represents only the response of the hearing aid for low level of signal (Tecca, Woodford,

& Kee, 1987). Thus, FG is not an appropriate measure to evaluate non-linear hearing aids that provide different gain at different levels of input signals.

In order to measure the performance of the hearing aid in the real ear or in case where there is a need to replace the current hearing aid due to unsatisfactory behavioral measures clinical judgment of benefit can be documented with objective hearing aid tests. Real ear measurements have contributed to the understanding of the role of outer ear and middle ear in influencing the hearing aid performance. If insert earphones are used in determining thresholds, the volume of the ear canal directly affects the sound pressure level generated at the eardrum. The canal volume increases as the child grows, and the SPL generated at the eardrum decreases. This, in turn, causes apparent hearing thresholds to ‘deteriorate’ with increase in age. If sound field assessment is used, the length of the ear canal determines its resonance properties, which determines the real-ear unaided gain (REUG) of the ear, and hence the real ear threshold. In addition, real ear measurements contribute to better understanding of inter-subject and intra-subject variability that occur in fitting hearing aids (Tecca, 1994).

The Real Ear Insertion Gain (REIG) measurements take all these parameters into consideration. The Real Ear Insertion Gain (REIG), the theoretical equivalent of functional gain, is determined by measuring the Sound Pressure Level (SPL) at the ear drum without a hearing aid (the Real Ear Unaided Gain, or REUG) and subtracting that from the SPL at the ear drum with the hearing aid in the ear (Real Ear Aided Gain, or REAG) (Hawkins, 2004). The FG and IG difference, in dB, represents the actual amount of gain achieved by the person wearing the hearing aid, because both methods account for individual differences in ear-canal geometry, acoustic characteristics of the ear, and coupling methods. Tharpe, Fino-Szumski, and Bess (2001) reported that approximately 60% of the audiologists verify hearing aid gain and frequency response settings for young children using behavioral measures such as sound field thresholds. In the school settings, nearly 80% of audiologists use these measures to adjust and fit the hearing aids.

Functional gain measures are typically limited to discrete audiometric frequencies, which rules out identifying acoustic interactions that may occur at octave or mid-octave frequencies, for example at 2700 Hz. By comparison, speech harmonics occur every 100 to 200 Hz, and most real-ear test systems use similar precision for their composite or warble-tone stimuli. Moreover, functional gain measurements are impossible to perform on clients who are unable to respond reliably (Zemplenyi, Dirks, & Gilman, 1985).

Numerous investigations have demonstrated that the test-retest reliability for probe microphone measurement between 1 and 5 dB (Ringdahl & Lejon, 1984; Hawkins, 1987; Hawkins, Alvarez, & Houlihan, 1991). For functional gain, separate measurements for aided and unaided threshold are required, and reliability may vary by 15 dB or more under ideal test conditions (Hawkins, Montgomery, Prosek, & Walden, 1987; Humes & Kirn, 1990; Stuart, Durieus-Smith, & Stenstrom 1990). That is, unless electroacoustic differences exceed 15 dB between the two test conditions (unaided vs. aided, or aided vs. aided), they cannot be measured accurately and repeatedly with functional gain. Further, these measurements may be more variable or impossible for paediatric or difficult-to-test clients, for wide dynamic range compression circuits, or outside of audiometric test booths. If behavioral measures are to be used, however, optimal aided threshold has one advantage over functional gain in that it uses only one rather than two highly variable measurements. At best, this translates to about a 10-dB difference between the two hearing aid settings to be interpreted as being significantly different from each other.

In recent years, there has been increased interest in the use of articulation index (AI) for assessing the hearing handicap and for measuring the potential effectiveness of amplification systems. This interest has been encouraged because of the ability of AI to explain the amount of difficulty the person with hearing impairment will have in understanding speech (Kamm, Dirks, & Bell, 1985). The practical application of AI also has been fueled by the popularity of prescriptive fitting strategies and the development of computerized probe-microphone measures (Mueller and Killion, 1990).

In order to know how the hearing aid functions at different input levels, insertion gain measurement would be more appropriate. Insertion gain (IG) measurement provides quick, more reliable and efficient method of quantifying the in-situ performance of hearing instruments than functional gain method (Stelmachowtcz, Hoover, Lewis, & Brennan, 2002).

Need for the study

People who have hearing impairment from early life may rely on acoustic cues for speech perception that are different from those used by adults with acquired hearing loss (Nittrouer & Boothroyd, 1990). When assessing the suitability of different signal processing strategies for children who have hearing impairment from early life, it is important to evaluate the strategies using a representative sample of those children, rather than generalizing from results derived from adult subjects with acquired hearing loss. There is relatively little published research on the use of compression amplification for young children, although there are a few studies with older children and adolescents (Bamford, McCracken, Peers, & Grayson, 1999; Jenstad, Seewald, Cornelisse, & Shantz 1999; Stelmachowicz, Kopun, Mace, Lewis, & Nittrouer, 1995). Hence, there is a need to compare the functional gain measurement with insertion gain measurement, as the functional gain is difficult to obtain from peadiatric population.

The functional gain measurement mainly reflects the functioning of the hearing aid for low level signals which is appropriate for linear hearing aids, as the linear hearing aids provide a constant gain irrespective of the level of the input signal. However, the gain in a non-linear hearing aid depends on the level of the input signal, i.e., a non-linear (NL) hearing aid provides more gain for soft signals and lesser gain for loud signals. Thus, the functional gain may not represent the gain that the hearing aid provides to an acoustic event in the person's environment with different levels of sounds. This limitation of the FG can be overcome by insertion gain measurement (Hawkins, 2004).

For evaluation of the non-linear hearing aid, it is possible to measure the gain provided by the hearing aid at different levels of the input signal using insertion gain measurement (ASHA, 1997, Paediatric Working Group, 1996). Objective measures such as insertion gain can depict the gain provided by different hearing aids at different levels as it can assess the hearing aid circuitry at different levels which is not possible through the subjective measures. The current study attempts to compare the usefulness of insertion gain measurement with that of functional gain measurement for verification of the performance of non-linear hearing aids. The prescriptive procedures for non-linear hearing aids use different target gains for different levels of the input signals and the hearing aid gain is adjusted to match these targets. This provides valuable information like audibility of speech over a range of commonly experienced input levels such as soft, average and loud speech. Hence, it is necessary to compare the FG and IG of hearing aids to see if one can be used instead of the other for hearing aids using different technologies (Hawkins, 2004).

Objectives of the study

1) To compare the insertion gain (IG) of hearing aid across the age groups.
2) To compare the different types of signals used for insertion gain measurement.
3) To compare linear and non-linear program modes for insertion gain measurement.
4) To compare insertion gain (IG) and functional gain (FG) for linear and non-linear hearing aids.
5) To investigate the relationship between the speech identification scores and the articulation index derived from the insertion and functional gain measures.

Chapter 2

REVIEW OF LITERATURE

The principle aim of any hearing aid selection and fitting procedure is to ensure that the environmental sounds, especially the conversational speech, is audible without being extensively loud (McCandless, 1994). Amplification for children must provide access to the wide variation in speech levels and spectra that occurs in different listening environments and with different speakers (Pearsons, Bennett, & Fidell, 1977; Stelmachowicz, Mace, Kean, & Carney, 1993). Non-linear amplification that incorporates compression with a low compression ratio has advantages over linear amplification for children. Because children, especially infants, cannot adjust the volume control of a hearing aid, a non-linear hearing aid that automatically provides higher gain for soft sounds and lower gain for loud sounds ensures adequate audibility for low input levels and listening comfort for high input levels (Kuk & Marcoux, 2002).

Hearing aids with digital signal processing (DSP) and compression circuitry have the potential to address the goal of providing audibility and comfort for speech signals ranging from very soft to high intensity levels. Wide dynamic range compression (WDRC) is particularly well suited to address this goal. This type of compression circuitry is designed to adjust the gain of the hearing aid without the need for a manual volume control. The amount of gain applied will be dependent on the input level (Stelmachowicz, Mace, Kopun, & Carney, (1993). That is, greater amount of gain is provided for soft sounds (e.g., speech at a distance), while less gain is provided for loud sounds (e.g., speech at close proximity).

The review of literature of the current study mainly focuses on the verification procedures of hearing aid selection for linear and non-linear hearing aids. The review of literature has been sub-divided into following topics:

2.1. Introduction to selection and verification procedure for hearing aid fitting

2.2. NAL-NL1 prescriptive formula

2.3. Subjective verification procedure - the functional gain measurement

2.4. Objective verification procedure - the insertion gain measurement

2.5. Verification procedures for the linear and non-linear hearing aids

2.6. Studies on comparison of the linear and non-linear hearing aid processing strategies

2.7. Comparison of the functional gain and the insertion gain measurement procedures for different processing strategies (linear and non-linear)

2.8. Potential sources of error in the various methods of insertion gain Measurement.

2.1. Introduction to subjective selection and verification procedure for hearing aid fitting

Selection and fitting of hearing aid is a step-wise procedure which involves hearing assessment, hearing aid pre-selection, hearing aid selection, verification of hearing aid and validation (Palmer, Lindley, & Mormer, 2000). The two fundamental objectives when fitting hearing aids are to ensure audibility of the speech input and that sounds are not uncomfortably loud (Skinner, 1980; Mason & Popelka, 1987; Byrne, 1992). A third important objective, especially in fitting hearing aids to children is to ensure consistent audibility in both ears over time. Currently used assessment tools for verification of the hearing aid selection include the Functional Gain (FG) and Insertion Gain (IG) measurement.

2.2. NAL-NL1 prescriptive formula

The National Acoustic Laboratories, (NAL) had developed a procedure for prescribing non-linear hearing aids, which is known as the NAL-NL1 procedure. The aim of this procedure is to maximize the speech intelligibility while maintaining the overall loudness at a level similar to that perceived by a person with normal-hearing, listening to the same sound (Dillon, 1999). For each input level, a different gain-frequency response is prescribed to achieve these aims.

2.3. Subjective verification procedure - the functional gain measurement

The FG measurement is a behavioural measure. The FG is defined as the difference, in dB, between the aided and unaided sound field thresholds. It is possible that many clients wear the hearing aid with the unsatisfactory real ear frequency response in spite of the use of sound field testing and other classic measurement for appropriate fittings.

The functional gain provided by the hearing aids, or the difference in audiometer dial readings between the unaided and the aided thresholds, was suggested to reflect gain for sounds presented at a conversational input level (Pascoe, 1975). The importance of this definition is that the aided threshold is a behavioral response at one perceptual level (i.e., threshold). The measurement of functional gain is needed because the coupler gain of a hearing aid is different from the gain that the wearer receives when it is worn. For a given hearing aid gain setting and test environment, there is only one aided threshold and hence one value of functional gain. It also means that although the wearer may experience a higher hearing aid output in the ear when the input is higher, the aided threshold (and functional gain) remains the same.

Aided sound field testing can also include articulation index (AI) which is an indicator of speech intelligibility of a person. The AI is an expression of the proportion of the average speech signal that is audible to a given client and therefore it can vary from 0 to 1.0. The calculation procedure for AI usually consists of dividing the speech signal into several frequency bands, each weighted according to theoretical contribution of the band to speech intelligibility. The frequency region surrounding 2000 Hz is normally rated as highest.

In the last two decades, there has been increased interest in the use of the articulation index (AI) or speech intelligibility index (SII) for assessing the hearing handicap and for measuring the potential effectiveness of amplification systems (Mueller & Killion, 1990). This interest had been encouraged by studies demonstrating the ability of AI to explain how much of difficulty the person with hearing impairment has in understanding speech. The AI was first proposed by French and Steinberg (1947). The practical application of AI also had been fueled by the popularity of prescriptive fitting strategies and development of computerized probe-microphone measures. Compared to the SII, which has correction factors for speech level distortion and hearing loss desensitization, the unmodified AI procedure was accurate for the relative comparison of various listening conditions in a single listener (Pavlovic, Studebaker, and Sherbecoe, 1986).

2.4. Objective verification procedure - the insertion gain measurement

Real ear test and sound box measurement are essential tools that allow quick documentation of whether or not hearing aid is performing upto some pre-determined goal. These tests are objective and can be completed within a few minutes. The insertion gain measurement is an objective measure using probe tube microphone (Stelmachowicz, Hoover, Lewis, & Brennan, 2002). Insertion gain measurement includes real ear testing.

For many years, probe microphone technology was the only method available to quantify the in-situ performance of hearing instruments. In the early 1980s, probe-microphone technology was introduced as an alternative to functional gain measures and the procedure was proved to be much faster than functional gain, multiple gain and output curves can be obtained in a few minutes. Early studies using this technique revealed large individual differences in the gain and SPL delivered to the ear using the same hearing instrument (Wetzell & Harford, 1983). Studies also showed age related differences in SPL developed in the ear canal for children at birth to seven years of age range (Barlow, Auslander, Rines, & Stelmachowicz, 1988).

The targets of the hearing aid is governed by the prescriptive formulae and are defined for speech inputs, hearing aids are tested using artificial test signals such as tones or noises. These measurements are then used to represent hearing aid performance for speech inputs. Majority of the times, a pure tone sweep is used as the test signal. But pure tone signals are narrow band signals that are sequentially presented across the frequencies. The broadband test signals, such as composite signal, are speech-weighted and have the broader bandwidth. The composite noise at the high signal levels underestimate speech levels, primarily in the mid- and high- frequency regions. The modulated narrowband test signals are narrower in bandwidth and fluctuate over time like speech. The high level speech weighted test signals tended to overestimate the aided speech.

Currently, probe microphone technology can provide a more efficient and reliable method of quantifying in-situ performance of hearing instrument than a functional gain method. However, in the current audiological practice, aided sound field threshold alone often are used to determine the appropriateness of hearing aid fitting.

Mueller and Killion (1990) had simplified the articulation index by providing the count-the dot method for the calculation of AI for insertion gain measurement. To determine the potential benefit from the hearing aid, insertion gain was added to the unaided threshold of the participant. Then, the aided thresholds were plotted in the count-the-dot audiogram using the real ear insertion response. Then, with count-the-dot procedure suggested by Mueller and Killion (1990), the AI can be computed. This audiogram format also served to make the standard audiogram more meaningful to clinician as well as clients.

2.5. Verification procedures for the linear and non-linear processing in hearing aids

Earlier form of amplification for children were primarily linear amplification with either peak clipping or compression limiting to limit the output which in turn affects the development of speech and language in young children. Although much is known about the merits and demerits of different forms of amplification, almost half of the hearing aids dispensed for children, especially those with severe to profound hearing loss, are still linear hearing aids with peak clipping or compression limiting (Tharpe, Fino-Szumski, &

Bess, 2001). Equal gain at all input levels maintain the relative intensity contrast of input signals and preserve the intensity cues within the speech signals. Such cues could be important in speech understanding, especially with increasing hearing loss (Van Tasell, 1993).

The advent of non-linear signal processing and the application of digital techniques in hearing aids brought new opportunities as well as challenges. One of the challenges is the appropriateness of using behavioral sound-field measurement for verification of hearing aid performance (Kuk & Ludvigsen, 2003).

The use of non-linear signal processing can alter the interpretation of the aided threshold (and functional gain). Furthermore, the approach to determine the aided threshold may need to be modified for non-linear hearing aids because the measured outcome depends on the properties of the signal used for the measurement (Kuk & Ludvigsen, 2003). Thus, a re-examination of the concept and the usefulness of the aided threshold for appropriate application and interpretation is needed.

The advantage, as well as disadvantage, of linear hearing aid is that equal gain is provided at all input levels until saturation. A fixed gain maintains the relative intensity contrast of the input signals and preserves the intensity cues within speech signals. Such cues are important in speech understanding, especially with increasing hearing loss (Van Tassel, 1993). On the other hand, children seldom stay in the listening environments that have the same range of intensity fluctuation. Indeed Stelmachowicz, Mace, Kopun, and Carney (1993) illustrated the problem by showing the variation in the input levels at child’s ear when their parents vocalize at different distances. At a fixed gain settings, as in the case of linear hearing aids that are fitted to optimize the amplification for a medium or conversation level input, the child may be over amplified in some instances and under amplified in others. This problem is especially aggravated in the case of severe to profound hearing loss in which the necessary high gain on the hearing aid may result in a constantly high output that saturates the hearing aid at even medium and high input levels (Macrae, 1991, 1993, 1995).

2.6. Studies on comparison of the linear and non-linear hearing aid processing strategies

A possible and practical solution for consistent audibility at more (if not all) input levels is the use of Wide Dynamic Range Compression (WDRC) hearing aids. By design, such hearing aids can provide more gain for low input levels and less gain for high input levels than linear hearing aids when both are matched in gain for medium input level. The exact amount of output for a given input level can be set by the clinician adjusting the hearing aids (Kuk & Marcoux, 2002).

The studies by Laurence, Moore, and Glasberg (1983); and Moore, Johnson, Clark, and Pluvinage (1992) compared linear and compression circuitry using a two channel fast acting compression aid that also served as linear aid by turning ‘off’ the compression. In a study by Laurence, Moore, and Glasberg (1983), the hearing aids were fitted using a method which ensured that speech at 70 dB SPL was comfortable and that the speech at 50 dB SPL was audible. Speech intelligibility was measured in quiet at three different levels and in presence of noise. The compression aid maintained high speech perception scores for all three speech inputs, whereas, the linear and unaided condition had lower scores for low input levels. Benson, Clark, and Johnson (1992) compared a multiband compression device with the client’s own device. Speech recognition testing at three input levels revealed better scores with compression device compared with subject's own aid for the two lower inputs but not at the higher levels of input.

Stelmachowicz, Kopun, Mace, Lewis, and Nittrouer (1995) showed that adaptive compression led to improved consonant discrimination (relative to linear amplification) for one out of three of their clients with moderate degree of hearing impairment.

Jenstad, Seewald, Cornelisse, and Shantz (1999) measured speech perception scores and their loudness rating of speech spectra for moderate to severe hearing loss. Results confirmed that Wide Dynamic Range Compression, (WDRC) processing provided greater audibility and comfort across a wide range of input levels than that provided by linear processing. For both measures, the types of circuits were matched for the average input levels, such as average speech at one meter, own voice at ear level and classroom at one meter. The circuit resulted in equivalent comfort and intelligibility for these average input levels. Greater benefit from WDRC was seen for soft speech inputs whereas, speech was consistently rated as more comfortable than linear hearing aids and speech intelligibility scores were higher. Greater benefit was also demonstrated for loud speech inputs for both the loudness comfort ratings and speech intelligibility.

Marriage and Moore (2003) conducted a study on children in the age ranging from four to fourteen years with moderate and severe to profound hearing loss. All children were fitted bilaterally. The frequency response of the test hearing aid in linear condition was adjusted to be similar to that of each child's own hearing aid for the same ear. The manufacturer's software automatically selected the appropriate gain values when the processing was switched to the compression. Hearing aid was programmed in two different ways (1) single channel WDRC, (2) single channel linear amplification with peak clipping and (3) single channel linear amplification with output limiting. Results revealed that WDRC can provide significant improvement in consonant discrimination for children with moderate and severe to profound hearing loss. There was no significant difference in the performance was obtained for moderate and severe to profound degree of hearing loss with WDRC for vowel discrimination task. The benefits of compression obtained for the children with moderate losses were comparable to those found in Jenstad, Seewald, Cornelisse, and Shantz (1999) in young adults. They found larger benefit for soft speech than for speech at moderate levels.

Bamford, McCracken, Peers, and Grayson (1999) compared a two channel aid (WDRC in low frequency channel and linear amplification in high frequency channel) with the children's own single channel aids (a mixture of linear aids and compression aids), using children in the age range from six to fifteen years of age. The cross over frequency was kept as 1600 Hz but it was varied according to audiogram configuration, with filter slope of 24 dB / octave. The compression in the low frequency channel had a knee-point lower than 50 dB SPL, making it constantly active. The two channel aids led to higher scores for speech perception in noise and higher satisfaction, as determined by the questionnaires.

Souza, Jenstad, and Folino (2005) compared speech recognition scores across different amplification strategies for adult listeners with severe hearing loss. The amplification options included conventional options (linear with peak clipping and linear with compression limiting) and newer strategies (multichannel wide dynamic range compression WDRC). Results demonstrated significantly poorer recognition and preference for a three channel WDRC system compared with a compression limiting system. It was observed that compression benefit was linked to the degree of hearing loss, with the smaller improvements observed when pure tone threshold exceeded 70 dB SPL. Boothroyd, Springer, Smith, and Schulman (1988) reported that the amplitude distortions were responsible for poorer performance of eight listeners with hearing impairment using two channel fast acting compression.

Henning and Bentler (2005) studied the effect of release time, compression ratio, and number of compression channels as well as interactions of these parameters, on gain difference between common non-speech test signals and speech. The results showed that the speech signals with speech-like spectral and temporal properties such as ICRA (International Collegium of Rehabilitative Audiology) signal showed the smallest gain differences from speech. Higher input levels (80 dB SPL) led to gain differences between speech and non-speech signals that were several dB greater than the difference observed for middle input levels (65 dB SPL). In their opinion, this was probably due to different signals causing varying amount of compression.

Davidson and Skinner (2006) examined the relationship of audibility for frequency specific sounds and the Speech Intelligibility Index (SII) to speech perception abilities of children with severe to profound sensorineural hearing loss using digital signal processing hearing aids with wide dynamic range compression. SII is accumulation of the audibility across the different frequency bands; weighted by the band importance function (not all frequency bands are equally important for speech intelligibility. SII can be seen as the total speech information available to the listener (Rhebergen & Versfeld, 2005). The results of their study showed that across 26 children, there was a significant correlation between aided pure tone averages (PTA) and their scores for monosyllabic words on Lexical Neighborhood Test (LNT) test presented at the soft level of 50 dB SPL. This was because children with better (lower) thresholds were able to hear more of the acoustic cues including consonant sounds. Compression cues also provided appropriate cues for recognition of speech at a loud overall level of 70 dB SPL.

2.7. Comparisons of the functional gain and the insertion gain measurement procedures for different processing strategies (linear and non-linear):

Functional gain is the difference between aided and unaided sound field thresholds with each of these measures having its own inherent variability. The variability of functional gain could be larger and smaller than that associated with sound field thresholds depending upon the assumed correlation between unaided and aided thresholds. The variation of aided sound field thresholds appears to be larger than unaided thresholds because of the additional source of variation associated with the removal and replacement of the hearing aid.

In addition, methods or procedural variables may contribute to a large range of individual data. The stimuli must be narrow-band with sharp rejection rates for the purpose of frequency specificity. As the frequency band becomes narrower, however, threshold measurements in a sound field may become more variable (Walker, Dillon, & Byrne, 1984). Participants must be selected with hearing losses that can be measured accurately in the sound field. Also, hearing aids need to be selected that allow for accurate threshold measurement for each individual (Macrae, 1982). The use of a single hearing aid to make threshold measurements on all subjects may be inadequate. If a high-gain instrument is used with individuals with mild hearing losses, accurate threshold measurements will not be possible because of ambient room noise or the internal noise of the hearing aid.

Large variations in individual data would also be expected if functional gain was not a valid estimate of real-ear gain. The validity of functional gain measures had been questioned on the basis that aided sound-field thresholds may be masked by the amplified noise of the test room or by the internal noise of the hearing aid (Macrae, 1982; Macrae & Frazier, 1980; Rines, Stelmachowicz, & Gorga, 1984; Walden & Kasten, 1976; Zemplenyi, Dirks, & Gilman, 1985).

Aided sound field testing (functional gain) is fraught with measurement error. And, even if reasonably valid thresholds are obtained it does not provide real ear maximum output of the hearing aid, which, for the paediatric client might be more important to know than the aided thresholds. Moreover, if a child is fitted with a WDRC hearing aid which is quite likely when digital instrument is fitted. In such a condition, the aided sound field thresholds do not represent the gain that will be present for average speech signal (Mueller, 2001). Most digital hearing aids employ WDRC. Hence, it is required to examine gain provided at various input levels which can be implemented through probe microphone measurements.

Mason and Popelka (1986) calculated the real-ear acoustic gain from sound pressure levels measured in the ear canal with a probe-tube microphone in unaided and aided conditions, at levels well above the ambient room noise. Results from 12 participants with hearing impairment suggested average difference between IG (probe tube microphone gain) and FG of 0.87 dB indicated increased sound pressure level in the ear canal during the aided condition relative to the unaided condition as an accurate estimate of functional gain. FG was compared with probe-tube measurements made in the ear canal with and without the hearing aid. The variability was reduced for measurements made at 250, 500, 3000, and 4000 Hz. The mean data indicated that FG and IG would be essentially equivalent methods for measuring the real ear gain of hearing aids. Both methods account for individual differences in the ear geometry, acoustic characteristics of the ear, and coupling factors. The limitations of the FG method were those of threshold measurements in general. First, obtaining unaided and aided thresholds is time consuming relative to the IG method. Second, both instrumentation and room noise conditions impose limitations on how low a threshold can be measured. Third, in the case of severe-to-profound hearing losses, unaided sound field thresholds may not be obtainable, which would prevent the calculation of FG.

Stelmachowtcz and Lewis (1988) represented hypothetical estimates of functional and insertion gain derived solely from the electroacoustic characteristics of three commercially available post-auricular hearing aids. They concluded that in the case of a non-linear hearing aid or a hearing aid in saturation, insertion-gain measures were more valid than functional-gain measures only because it is necessary to obtain these measures at relatively high levels due to the internal noise of measuring equipment and/or undesirable room noise. Functional gain can overestimate the gain for an average speech input if the aided thresholds occur at relatively low levels.

The insertion gain measurements, however, more accurately reflected the fact that the components of average speech at 4000 Hz would not be audible. Real ear SL was more accurate estimate as the loud conversation or the peaks of speech causes the hearing aid to saturate in the high frequencies. In cases of a more profound hearing loss, a severely reduced dynamic range may point to the need for compression amplification or an increase in maximum output, if possible. Because both threshold and aided measures are referenced to ear canal SPL, the estimated SL of speech at each frequency is accurate and requires no complicated conversions. Ensuring the audibility of as much of the speech spectrum as is possible may be the largest single factor in providing an optimum hearing-aid fitting (Killion, 1985; Seewald, Ross, & Spiro, 1985). Regardless of the approach used to select amplification characteristics, this measurement technique may provide a more direct method by which to characterize aided results.

Fortunately, a 60 dB SPL pure-tone input corresponds fairly well to most estimates of the average long-term speech spectrum. The idea of an ‘aided audiogram’ can be a misleading concept in that a hearing aid does not ‘shift’ an individual's thresholds (Gittelman & Popelka, 1987; Schwartz & Larson, 1977). As had been suggested by Erber (1973); and, Schwartz and Larson (1977) it may be more appropriate to think in terms of an ‘amplified speech spectrum’ and how it relates to an individual's unaided thresholds. The in-situ measurement, in which test signal is generated by hearing aid itself, rather than insertion gain was used because the quantity of interest is the actual SPL in the ear canal with the hearing aid and ear mold in place. The dB difference between the unaided thresholds and the amplified speech spectrum yields an estimate of the sensation level (SL) of average conversational speech as a function of frequency.

The SL estimates will be accurate only if the unaided thresholds represent the SPL in the ear canal at threshold. Hawkins (1987) had suggested that this can be accomplished easily by placing a probe-tube microphone into the ear canal prior to obtaining thresholds. Although the physical constraints of different probe-tube microphone systems may influence how the signal is to be delivered to the ear, the tympanic membrane SPL at threshold should not depend upon how the signal is transduced such as through the hearing aid, circumaural earphone, insert earphone, sound-field speaker (Killion, 1978). After the threshold is determined at a given frequency, a steady-state tone can be presented at threshold and the ear canal SPL can be recorded. If the measured thresholds are below the noise floor of the measuring system (usually 40-50 dB SPL), then a constant dB increment can be added to the signal and subtracted from the measured probe tube SPL.

Zemplenyi, Dirks, and Gilman (1985) conducted a study on 12 adults to determine the hearing-aid insertion gain (from a probe measurement of aided thresholds) and to compare these results with gain determined from couplers and functional gain. The results demonstrated a difference exceeding 15 dB at 3.0 and 4.0 kHz between the two measurements (the coupler gain and functional gain) in the middle and high frequencies. The dB differences between these measurements were reduced if the ear-simulator measurements (ESM) were made using the ear molds of the individual. These differences are even further reduced (except at 5.0 and 6.0 kHz) if the ear-simulator measurements with ear molds are determined in the ear simulator whose ear canal length corresponds to the measured length of the individual subject's occluded ear canal (ESM_L). A close agreement was found between the probe method and the simulator with ear mold or functional-gain measurements through 4.0 kHz. At 5.0 and 6.0 kHz, the probe system underestimated the gain particularly when compared to the measurement using the simulator with ear mold.

2.8. Potential sources of error in the various methods of gain measurement:

There were several factors contributing to the differences observed at 5.0 and 6.0 kHz during functional and insertion gain measurement.

1. *Standing waves* - As reported by Stinson, Shaw, and Lawton (1982), a standing wave exists in the ear canal which has a minimum that occurs at a position dependent on the stimulus frequency. Their results demonstrated that the minimum of the standing wave occurred at about 12 mm from the eardrum for a stimulus frequency of 6.5 kHz. At lower frequencies, the position of the minimum occurs at distances further away from the eardrum. Since the slope of the standing-wave curve is steep near its minimum, the error in probe SPL measurement can be quite large even for a small difference in the position of the probe with respect to the eardrum. Thus, differences in ear canal lengths could affect the position of the measurement point with respect to the minimum of the standing wave and generate frequency-dependent errors (Gilman, 1984).

2. *Coupling of earmold to simulator* - The individual earmold is coupled to the ear simulator using a concave adapter into which the earmold is mounted with silicone putty. The volume of air trapped between the medial end of the earmold and the adaptor will affect the high-frequency gain response i. e., gain response at 5.0 kHz and 6.0 kHz. This type of error will not create a difference between the probe and eardrum SPL per se, but could create an error between functional gain and insertion gain.

3. *Resonance in the hearing aid response* - The hearing-aid frequency response showed a sharp anti-resonance in the output of the hearing aid starting near 5.0 kHz and resulting in a rapid drop in the output of 9-12 dB in the range from 5.0 to 6.0 kHz. Since the test tone was frequency modulated, the sharp-negative slope at 5.0 to 6.0 kHz could

create an error in the SPL measured at both the simulator eardrum and at the probe.

People who have hearing impairment from early life may rely on acoustic cue for speech perception that are different from those used by adults with acquired hearing loss (Nittrouer & Boothroyd, 1990).When assessing the suitability of different signal processing strategies for children who have hearing impairment from early life, it is important to evaluate the strategies using a representative sample of those children, rather than generalizing from results derived from adult subjects with acquired hearing loss. There is relatively little published research on the use of compression amplification for young children, although there are a few studies with older children and adolescents (Bamford, et al. 1999; Jenstad, et al. 1999; Stelmachowicz, et al. 1995).

A number of studies have suggested that the small differences observed between functional gain and insertion gain are due to measurement error (Dillon & Murray, 1987; Harford, 1981; Mason & Popelka, 1986). One exception to this rule occurs when an individual has normal or near-normal hearing sensitivity in a frequency region in which significant gain is provided. Under these circumstances, internal hearing aid noise masks the aided sound-field thresholds, invalidating estimates of functional gain. In this situation, insertion gain provides a valid estimate of real-ear gain because these measures are obtained at supra-threshold levels. Although, this effect had been well documented (Dillon & Murray, 1987; Macrae, 1982; Macrae & Frazier, 1980; Mason & Popelka, 1986; Rines, Stelmachowicz, & Gorga, 1984), there were at least three other circumstances in which insertion gain and functional gain may not agree. These included: 1) a high gain hearing aid with a relatively low maximum output, 2) certain types of non-linear hearing aids, and 3) in some clients with profound hearing loss.

Digital processing strategies have brought possibilities as well as challenges for persons with hearing impairment and for hearing professionals. Hearing aid fitting involves several stages. They are evaluation, pre-selection of hearing aid, hearing aid fitting, validation, and the final stage includes counseling. Verification stage includes both subjective and objective measures. The functional gain measurement which is done only at one level correlates well with the soft or moderate (conversational) level of signals which is appropriate for linear hearing aids (Kuk & Ludvigsen, 2003). Because linear hearing aid provides equal amount of gain irrespective of soft, moderate, loud level of signals until saturation. Whereas, non-linear hearing aids provide different amount of gain for different levels of signal. So, functional gain may not be appropriate for such processing strategies because it assesses only the soft or moderate levels.

On the other hand, objective measures such as insertion gain can depict the gain provided by different hearing aids at different levels as it can assess the hearing aid circuitry at different levels which is not possible through the subjective measures. But insertion gain is susceptible to measurement errors like location of the probe inside the ear canal, depth of probe insertion, location and azimuth of loudspeaker. It's difficult to carry out insertion gain measurement in young children as co-operation and fatigue increase the variability. Thus, there is a need for evaluating the objective verification procedure which can supplement as well as complement the subjective verification procedure, especially for children.

Chapter 3

METHOD

The present study was conducted to evaluate the effectiveness of the verification of linear and non-linear hearing aid fittings using functional gain and insertion gain.

Participants

Twenty children with hearing impairment using hearing aids participated in the study. The children used a range of different models of Behind-The-Ear (BTE) hearing aids, and all of them wore their hearing aids through most of their waking hours, i.e., not less than eight hours per day.

Inclusion criteria

- Age range: Four to six years [mean age: 4.98 years, standard deviation (SD): 0.68 year]. The participants were further divided into two groups. Group I with participants in the age ranging from 4+ to 5 years, and Group II with participants in the age ranging from 5+ to 6 years. The demographic details of the participants in the two age groups are provided in the Table 3.1.

Table 3.1

Details of the participants in the study.

Groups	*Male*	*Female*	*Total N*	*Age (in years) Mean (SD)*
Group I (*4+ to 5 years)*	4	6	10	4.41 (0.33)
Group II (*5+ to 6 years)*	2	8	10	5.56 (0.37)
Total / Mean (SD)	6	14	20	4.98 (0.68)

- Degree and type of hearing loss: From moderately severe to profound sensorineural hearing loss.
- Native speakers of Kannada language attending the pre-school and/or individual therapy session at All India Institute of Speech and Hearing, Mysore. All the participants were at or above the stage of word identification.
- On otoscopic examination, all participants had ear canals that were free from cerumen, debris or foreign body.
- They had normal middle ear functioning as indicated by middle ear analyzer. The middle ear peak pressure was ranging from +50 to -100 daPa, and the admittance was ranging from 0.5 to 1.75 ml with the probe tone frequency of 226 Hz. The acoustic reflexes were absent for all the participants.

Exclusion criteria

- Presence of associated problems like mental retardation, cerebral palsy.
- Indication of retrocochlear pathology.
- Indication of cognition problem.

Test Environment

The testing was performed in an air conditioned sound treated double or single room environment.

Instruments / Material used

1. A calibrated sound field audiometer (Madsen OB922, version 2).
2. A calibrated hearing aid analyzer (Fonix 7000 Hearing Aid Test System, version 1.8).
3. A digital BTE hearing aid, coupled with custom ear mold. The hearing aid had six channels with a fitting range from moderate to profound degree of hearing loss. The hearing aid was programmed in two different program modes:

   i) Non-linear program mode

   ii) Linear program mode

4. Hardware and software to program the hearing aids. A personal computer connected to HIPRO for programming the hearing aid. The NOAH software (version 3.1.2) and the hearing aid specific software (Aventa, version 2.6) along with WinCHAP

(Computerized Hearing Aid Program for Windows, version 2.82) software were installed in this personal computer.

5. Picture identification test material in Kannada developed by Vandana (1998). This had four lists, each with 25 bi-syllabic PB (phonemically balanced) words. The test material included words which were in the vocabulary of the children in the age ranging from four to seven years. List A consisted of 50 words, the order of these words were randomized to construct the List B. From each of the lists (List A and List B) half lists were constructed and the phonemic balance was maintained for both half and the full list. Thus, the test consisted of four Phonemically Balanced (PB) word lists, each with 25 words.

For response identification, a picture booklet was used which had four pictures on each page. Among four picturized words, one was the test word and the other was the distractor word. The distractor word had same vowel ending as the test word. The other two pictures were randomly selected.

Procedure

The experiment was conducted in three stages –

Stage I: Optimization of parameters for linear and non-linear program modes. The NAL-R was used for linear program mode and NAL-NL1 was used for non-linear program .

Stage II: Verification of hearing aid fitting through insertion gain (IG) measurement.

Stage III: Verification of hearing aid fitting through functional gain (FG) measurement.

Stage I: Optimization of Parameters for Non-linear and Linear Program Modes

The test hearing aid was connected to the HIPRO which in turn was connected to the personal computer. The personal computer had the NOAH and the hearing aid specific software Aventa. The participant's demographic as well as the audiometric data were entered into the NOAH software. Then the hearing aid was detected and programmed in the Aventa software.

For optimizing the hearing aid program in non-linear mode, insertion gain measurement was carried out.

The hearing aid was matched with the target gain using the NAL-R formula using auto fit feature in the hearing aid specific software. The NAL-R formula in hearing aid specific software uses only one target gain curve. That is, for a linear hearing aid, the overall gain remains the same irrespective of the input level of the sound. Hence, the NAL-R formula was used. This was stored as Program 1 (P1) of the hearing aid.

In a similar way, the gain was also matched with the non-linear fitting technique using NAL-NL1 formula (Dillon, 1999). As the NAL-NL1 formula is for non-linear hearing aids, it provides more gain for the soft level of sounds, and lesser gain for higher level of sounds. Hence, the hearing aid specific software contained two target curves - one for 50 dB SPL, and the other for 80 dB SPL.

As there were two separate programs available in the test hearing aid, the NAL-R settings was stored in Program 1 (P1) and the NAL-NL-1 settings was stored in Program 2 (P2) of the hearing aid.

In each age group and for each participant, the measurement was done only for one ear, equal numbers of right and left ears were considered. Custom made soft ear molds were used to couple the test hearing aid to the ear of the participant during the measurement. The hearing aid program in linear mode was optimized using insertion gain measurement option of the Fonix 7000 hearing aid test system.

In each age group and for each participant, the measurement was done only for one ear, equal numbers of right and left ears were considered. Custom made soft ear molds were used to couple the test hearing aid to the ear of the participant during the measurement. The hearing aid program in linear mode was optimized using insertion gain measurement option of the Fonix 7000 hearing aid test system.

Stage II: Verification of Hearing Aid Fitting Through Insertion Gain Measurement

Verification through insertion gain measurement utilized pressure method. In this method the reference microphone was placed as close as possible to the hearing aid microphone while the measurement was being done. The reference microphone monitors the SPL reaching the hearing aid from the loudspeaker. If the input level is higher or lower than the desired output level, the reference microphone and the regulating circuitry automatically turned the volume of the sound coming from the speaker down or up, until the required level is obtained. For verification of the hearing aid fitting the insertion gain measurement were carried out in linear and non-linear program modes and the hearing aid was optimized using insertion gain measurement option of the Fonix 7000 hearing aid test system.

Insertion Gain Measurement Procedure:

The Fonix 7000 was connected to the personal computer. The personal computer had the WinCHAP installed in it. The step-wise procedure given below was used for insertion gain measurement through Fonix 7000 and WinCHAP, using the protocol given in Table 3.2.

1) The instrument (Fonix 7000) was switched ‘on’.
2) Leveling of the instrument was ensured before carrying out the insertion gain measurement.
3) From the opening screen, the real ear navigation was accessed in the Fonix 7000 module. Later, the insertion gain measurement mode was used.
4) Placement of the sound field loudspeaker for real ear measurement: The sound field speaker of Fonix 7000 was placed 12 inches from the participant’s head. The sound field speaker was at an angle of 45 ° (half-way between the participant’s nose and ear), as shown in the Figure 3.1.

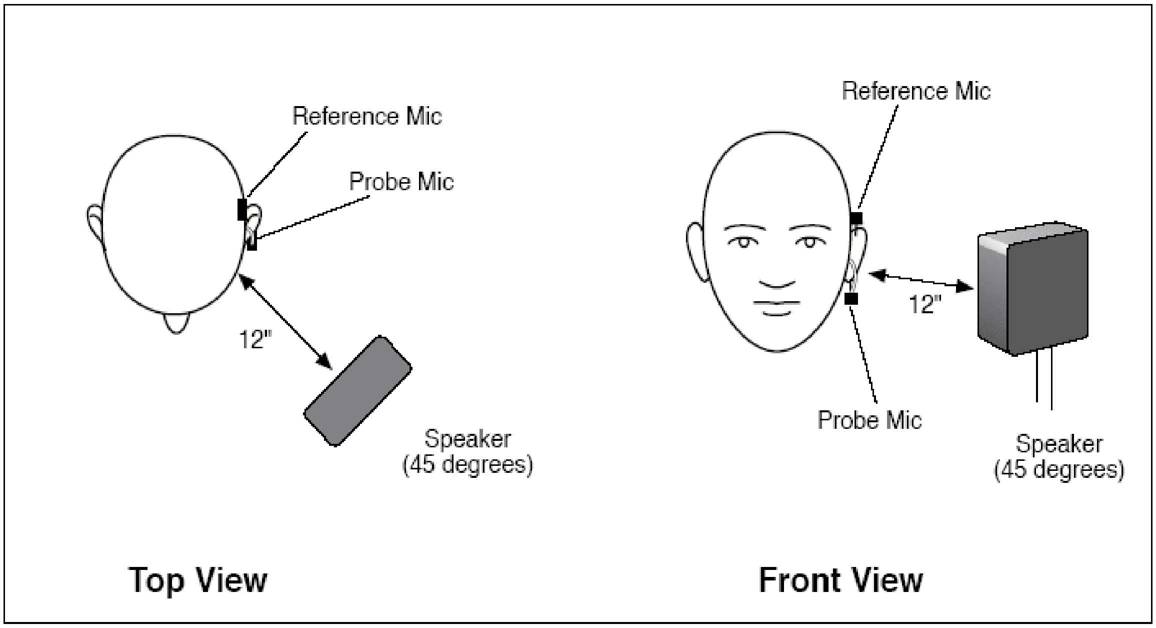

Fig. 3.1: Location of the loudspeaker and the participant for real ear measurement.

5) Marking the probe tube for insertion gain measurement.

- An unattached probe tube was placed on a flat surface along with the participant's earmold.
- The ear mold was held next to the probe tube, so that the tube rested along the bottom of the canal part of the earmold, with the tube extending approximately 5 mm past the canal opening. This was done as shown in the Figure 3.2.
- The probe tube was marked where it met the outside surface of the earmold with a marking pen.
- The probe tube was attached to the body of the probe microphone. The probe microphone was attached to the velcro pad on the ear hook.

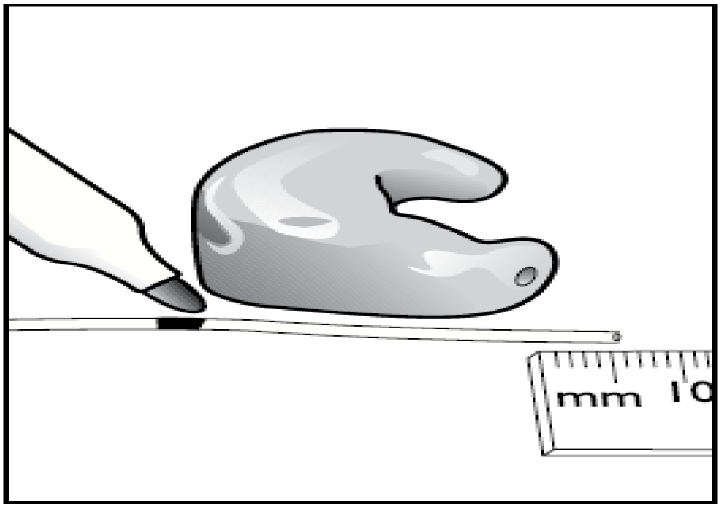

Fig. 3.2: Marking of the probe tube to be inserted in the ear canal.

6) Placement of ear hook, reference microphone, and probe microphone:

- The integrated probe microphone set was positioned on the participant’s ear.
- The small reference microphone was secured on the ear hook above the ear.
- The ear hook slider was adjusted up or down for optional positioning of the probe tube into the ear, as shown in the Figure 3.3.
- The probe tube was inserted into the participant’s ear (without the earmold or aid), so the mark was at the location where the bottom of the outer surface of the ear mold was, once the ear mold was in place.

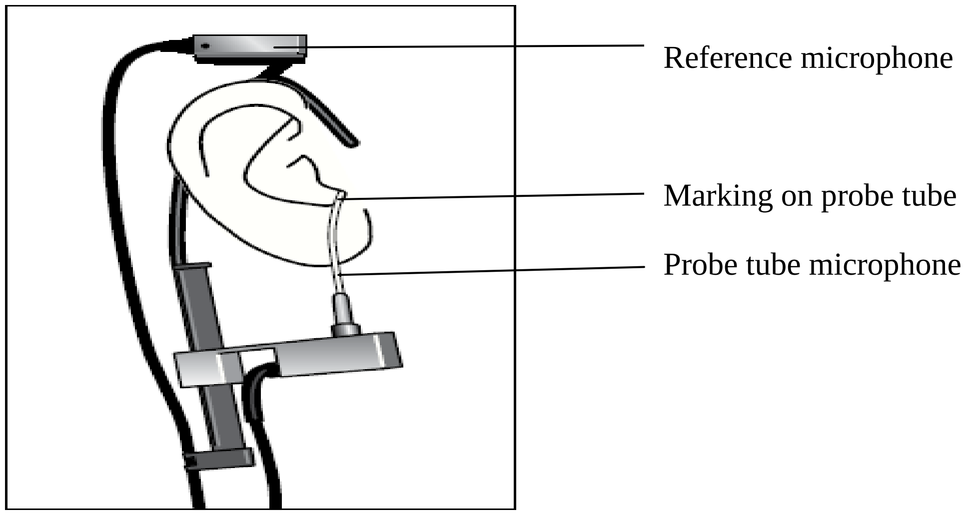

Fig. 3.3: Placement of reference microphone and probe tube microphone.

7) Later, the protocol given in Table 3.2 was used for the IG measurement. After setting up the participant and the instrument for insertion gain measurements, the Win CHAP software icon was selected from the monitor of the personal computer. The WinCHAP is the windows based Computerized Hearing Aid Program (CHAP). This program makes interfacing of Fonix instrument with the computer easy and convenient while providing with the data base of clients. This software enabled storing the participant’s data and hearing aid data. This software also enabled storing the important measurements made with Fonix 7000 analyzer, eliminating the need for paper or pen.

Table 3.2

Protocol for insertion gain measurement

	Measurement type	*Insertion Gain (IG)*
1	Reference microphone	Enabled; Placed over the pinna of the test ear of the participant
2	Placement of probe microphone	Probe microphone extended 3 to 5 mm from the tip of the ear mold in the ear canal of the participant
3	Placement of loudspeaker	12 inches from the participant's head, at an azimuth angle of 45 ° from test ear side
4	Stimuli type	- Pure tone sweep; - ANSI digi speech signal
5	Stimulus levels	50, 65, and 90 dB SPL
6	Formulae used	NAL-R (for linear program mode) ; NAL-NL1 (for non-linear mode)
7	Hearing aid program mode	Linear in P1; Non-linear in P2
8	Output limiting	125 dB SPL
9	Noise tracker	Off

8) The IG was done for pure tone and ANSI digi speech signals for linear and non-linear program modes for each of the participant using the following procedure:

- After enabling Win CHAP measurement in the personal computer, 'Data entry' icon was selected. Then, by clicking on 'Client's icon', the details of each participant were stored.
- The 'Hearing aid icon' was selected to store the details of the hearing aid used for testing.
- Then, from the main menu of WinCHAP, 'Test menu' for the specific participant was accessed.
- In the 'Client's test menu', the details of the audiogram such as air conduction, and bone conduction thresholds were entered.
- Next, from the WinCHAP's, DSL/NAL Test menu, 'NAL' testing i.e., NAL screen gain, Real ear and Real ear aided gain was selected.
- On entering the NAL screen, three target gain curves (for soft, moderate and loud levels) were displayed at different frequencies from 125 Hz to 8000 Hz as shown in the Figure 3.4.

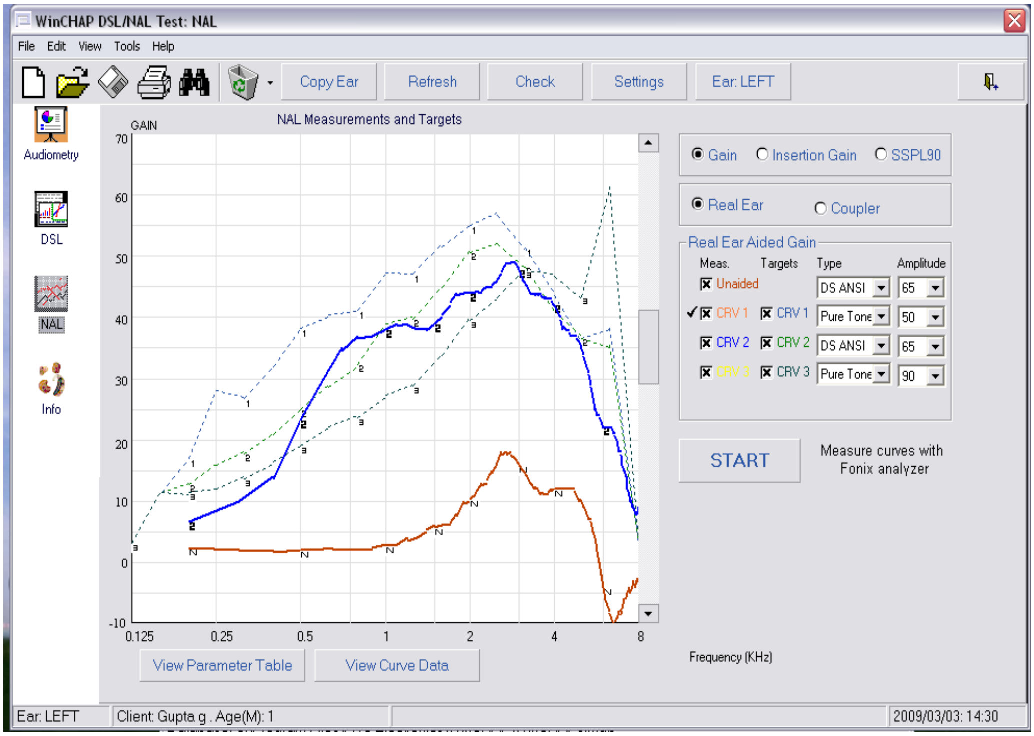

Fig.3.4: Three target aided curves (marked as dashed lines) and aided curve for 65 dB input (marked as solid line), on WinCHAP.

9) By selecting the 'unaided' from the 'Real Ear Aided Gain' option, three unaided measurements were made for pure tone signals (at 50, 65, & 90 dB SPL) and three unaided measurements were made for ANSI digi speech signal (at 50, 65, & 90 dB SPL). For the aided measurements in linear program mode, the CRV1, CRV2 and CRV3 were selected for measurement one after other, from Real Ear Aided Gain sub-menu. This measurement was made for pure tone (at 50, 65, & 90 dB SPL) and for ANSI digi speech signal (at 50, 65, & 90 dB SPL).

For unaided and aided measurements in linear program mode, the data were tabulated from 'View curve data' option. From this option, the following data for each participant were tabulated.

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1) Data tabulated from unaided response for pure tone and ANSI digi speech signal included:

   a) Real ear unaided gain (REUG) for an input level of 50 dB SPL at different frequencies.

   b) Real ear unaided gain (REUG) for an input level of 65 dB SPL at different frequencies.

   c) Real ear unaided gain (REUG) for an input level of 90 dB SPL at different frequencies.

2) Data tabulated from aided response for pure tone and ANSI digi speech signal for linear and non-linear program modes of the hearing aid included:

   a) Real ear aided gain (REAG) for an input level signal of 50 dB SPL at different frequencies.

   b) Real ear aided gain (REAG) for an input level of 65 dB SPL at different frequencies.

   c) Real ear aided gain (REAG) for an input level of 90 dB SPL at different frequencies.

3) Insertion gain was obtained by subtracting the unaided gain from the aided gain at different frequencies, separately for all the three different levels, i.e., at 50, 65, and 90 dB SPL for linear program mode. The different frequencies at which the insertion gain were noted were 250, 500, 1000, 2000, 4000, 6000, and 8000 Hz. The similar

procedure was carried out for non-linear program mode of hearing aid also, for each of the participant.

4) Articulation Index (AI) calculation from insertion gain method

The count-the-dot method for calculating the AI was utilized to convert the REIR into the AI values, as recommended by Mueller and Killion (1990). In this procedure for calculation of articulation index from insertion gain, Real Ear Insertion Response (REIR) values were subtracted from the unaided audiogram of the participant. This provided the real ear insertion thresholds of the participant. The threshold values were plotted across the frequencies from 250 Hz to 5 kHz in Count-the-dot audiogram (Mueller & Killion, 1990). To calculate the articulation index, the number of dots which were below the real ear insertion thresholds across frequencies was counted. By dividing the total number of dots which were below insertion threshold from 100, the AI value was obtained. The AI was calculated for three different levels 50, 65, and 90 dB SPL for linear as well as non-linear program modes. So, for each participant six AI values (three in linear program mode & three in non-linear program mode) were obtained. Table 3.3 and 3.4 depict the calculation of AI using count-the-dot method for one of the participants.

Table 3.3

Calculation of AI using count-the-dot method: An illustration

Parameters	*Frequency (Hz)*					
	250	*500*	*1000*	*2000*	*4000*	*6000*
Unaided threshold (UAT) in dBHL	70	70	70	80	85	85
REIG at 50 dB SPL in dB	29	36	44	52	37	45
REIG at 65 dB SPL in dB	25	30	37	49	35	40
REIG at 90 dB SPL in dB	22	24	24	34	33	37
UAT – REIG at 50 dB SPL	41	34	26	28	48	40
UAT – REIG at 65 dB SPL	45	40	33	31	50	45
UAT REIG at 90 dB SPL	48	46	46	46	52	48

Table 3.4

Computation of AI at three levels

Input level	*AI*
At 50 dB SPL	0.51
At 65 dB SPL	0.40
At 90 dB SPL	0.11

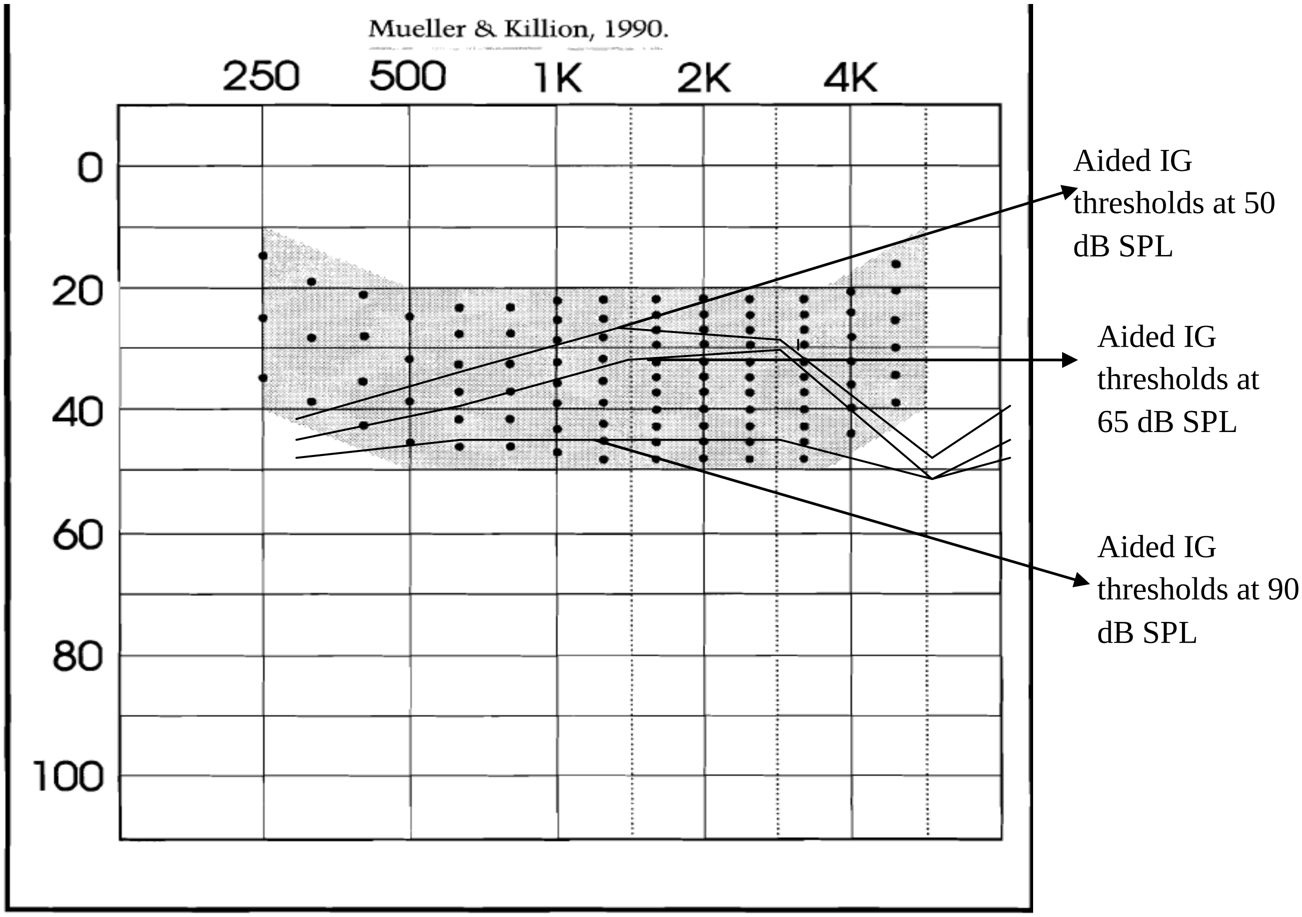

Fig. 3.5: The ‘Count-the-dot’ audiogram format for calculation of Articulation Index (Mueller & Killion, 1990).

Stage III: Functional gain measurement (FG measurement)

The functional gain, using aided thresholds and Speech Identification Scores (SIS), were measured for linear and non-linear program modes of the hearing aid for each participant. Protocol for functional gain measurement procedure is shown in the Table 3.5.

Table 3.5

Protocol for functional gain measurement

	Measurement type	*Functional gain (FG)*
1	Placement of loudspeaker	At an azimuth angle of 45 ° from test ear side, at one meter distance from participant.
2	Stimuli type	- Warble tone at 250, 500, 750, 1000, 1500, 2000, 3000, 4000, 6000 Hz, for unaided and aided thresholds. - Monitored live voice for Speech Identification Scores (SIS).
3	Presentation level for SIS	35, 50, and 75 dB HL (equivalent to 50, 65, & 90 dB SPL).
4	Response mode for SIS	Closed set picture identification
5	Hearing aid program mode	- Linear (P1); - Non-linear (P2)
6	Prescriptive formulae used	- NAL-R (for linear program mode) - NAL-NL1 (for non-linear mode)

a. The FG measurement was carried out with the calibrated sound field audiometer. The loudspeaker was kept at a distance of one meter and 45 ° Azimuth from the test ear of the participant, in the calibrated sound field. For the measurement of FG, the unaided thresholds for warble tone signals were obtained. The aided thresholds were obtained after fitting the hearing aid in linear program mode. Similarly, aided thresholds were measured for the non-linear program mode of the hearing aid also. The thresholds, for the unaided and aided in linear as well as non-linear program modes, were measured at octave and mid-octave intervals from 250 to 6000 Hz. The difference between unaided and aided threshold at each frequency was computed to obtain the functional gain at that frequency. The difference between unaided and aided threshold was obtained separately for linear as well as for non-linear program modes of the hearing aid.

b. The count-the-dot method for calculating the AI was utilized to convert the aided thresholds into the AI values, as recommended by Muller and Killion (1990). The threshold values of individual participant were plotted across the frequencies from 250 Hz to 6 kHz in Count-the-dot audiogram (Mueller & Killion, 1990). To calculate the articulation index, the number of dots which were below the aided thresholds across frequencies was counted. By dividing the total number of dots which were below aided threshold from 100, the AI value was obtained. The AI was calculated for linear as well as non-linear program modes. So, for each participant two AI values (one in linear program mode & one in non-linear program mode) were obtained.

c. Further, the unaided and aided SIS were also obtained, using speech identification test in Kannada (Vandana, 1998), at three levels which will be equivalent to the presentation levels used during insertion gain measurement. The SIS was measured for linear as well as non-linear program modes at 35 dB HL, 50 dB HL, and 75 dB HL (equivalent to 50, 65, & 90 dB SPL).

The participants were given instructions in Kannada in following way “Now, you will hear some words through the loudspeaker. Listen to each word carefully. Look at all the four pictures on the page. You have to point to the picture of the word that you hear". Initially, three practice items were presented at the comfortable level. For the actual testing, a total of 25 words were presented at each of the above mentioned presentation levels. The picture booklet was used to elicit the responses. Thus, the closed set SIS was noted at each level. Both the order of the test material and level of presentations were randomized.

The scoring was done by noting the number of correct pictures being identified. Each word identified correctly was given a score of one and the incorrect identification was given a score of zero. The maximum score was 25 as there were 25 words in the word list. These SIS were not converted into percentage. The same procedure was followed for both linear as well as non-linear program modes of the hearing aid, for each participant.

Thus, for each of the participant, the following data were collected:

6) From the insertion gain (REIG) for different stimuli

Type of stimulus	*Intensity (in dB SPL)*	*Frequency (Hz)*						*AI*
		250	500	1000	2000	4000	6000	
Pure tone	50							
	65							
	90							
ANSI digi speech	50							
	65							
	90							

This was done for each of the participant, with the hearing aid program in linear and non-linear modes.

2) From the functional gain measurement

a) The functional gain

Frequency (in Hz) →	*250*	*500*	*750*	*1000*	*2000*	*3000*	*4000*	*5000*	*6000*	*AI*
Linear program mode										
Non-linear program mode										

a) SIS at 35, 50, and 75 dB SPL

This was done for each participant in linear and non-linear program modes. The data were tabulated and statistically analyzed.

Chapter 4

RESULTS

The current study was conducted to evaluate the effectiveness of the verification procedure, using the functional gain and insertion gain, for linear and non-linear hearing aids. The study, also evaluated whether there was any difference between the two age groups for the insertion gain (IG) measure.

The following measures were analyzed using appropriate statistical tools through Statistical Package for the Social Sciences (SPSS for windows, Version 16) software:

4.1. Insertion gain for pure tone and ANSI digi speech signal, in linear and non-linear program modes:

4.1.1. Descriptive statistics, mean and standard deviation (SD), of the insertion gain for the Group I and Group II across the low-, mid- and high- frequencies were obtained.

4.1.2. The significant difference in the insertion gain between the Group I and Group II across low-, mid- and high- frequencies was analyzed using mixed ANOVA.

4.2. Significant difference for ANSI digi speech and pure tone signals:

4.2.1. The mean insertion gain was analyzed separately for the two types of signals, pure tone and ANSI digi speech signal, in both linear and non-linear program modes

. 4.2.2. The paired t-test was administered to find out whether there was a significant difference between mean IG in the linear and non-linear program modes, for both the signals separately.

4.2.3. The paired t-test was administered to find out whether there was a significant difference between mean insertion gain for low and high intensity signal in linear and non-linear program modes.

4.3. Difference between FG and IG:

4.3.1. The difference between the mean functional gain and insertion gain at different intensity levels (50 dB SPL, 65 dB SPL, & at 90 dB SPL) were compared. The functional gain obtained at one level was compared with insertion gain at all three levels. The difference between them was plotted separately in the graphical form for linear and non-linear program modes for the four octave frequencies (500, 1000, 2000, & 4000 Hz).

4.3.1.1. Mean and SD for the difference between the mean functional gain and insertion gain at three different levels, for linear mode.

4.3.1.2. Mean and SD for the difference between the mean functional gain and insertion gain at three different levels, for non-linear program mode.

4.3. 2. Significance of difference between the mean Functional Gain (FG) and Insertion Gain (IG) for linear and non-linear program modes at different frequencies was measured using paired t-test.

4.4. Relationship between SIS and AI:

4.4.1. Relationship between SIS and AI computed from functional gain measure (AI_{FG}): For linear mode, Pearson's correlation between Speech Identification Scores (SIS) at three levels (35 dB HL, 50 dB HL, & 75 dB HL) and Articulation Index obtained through functional gain measure (AI_{FG}) was analyzed. This was done for both non-linear as well as linear program modes.

4.4.2. Relationship between SIS and AI through insertion gain measure (AI_{IG}): The AI_{IG} at three levels (50 dB SPL, 65 dBSPL, & 90 dB SPL) were correlated with SIS for three equivalent levels (35 dB HL, 50 dB HL, & 75 dB HL) respectively, for linear program mode. Similar analysis was carried out for non-linear program mode also.

4.1. Insertion gain for pure tone and ANSI digi speech signals in linear and non-linear program modes.

4.1.1. The mean insertion gain in participants of the Group I and Group II across low-, mid- and high- frequencies was obtained for pure tone signals. The Tables 4.1 and 4.2 shows the mean and SD, in linear and non-linear program modes, for pure tone signal and for ANSI digi speech signals.

Table 4 .1

Mean and standard deviation (SD) of insertion gain at different frequencies in linear and non-linear modes for pure tone signals, for Group I and Group II.

Frequency (Hz)	*Input intensity (dB SPL)*	*Groups*	*IG for linear (in dB)*		*IG for non-linear (in dB)*	
			Mean	*SD*	*Mean*	*SD*
200	50	Group I	39.36	12.59	39.79	11.81
		Group II	38.99	7.11	37.22	6.65
	65	Group I	34.54	9.90	33.11	7.84
		Group II	34.95	7.76	34.83	6.22
	90	Group I	28.36	9.85	28.93	9.67
		Group II	33.14	6.38	31.67	5.81
500	50	Group I	55.53	6.78	55.48	6.91
		Group II	51.5	6.39	52.26	6.84
	65	Group I	49.29	8.52	47.88	10.28
		Group II	48.21	6.68	50.17	6.70
	90	Group I	39.63	9.36	36.96	9.42
		Group II	40.26	6.58	38.84	4.93
1000	50	Group I	59.57	8.72	59.51	9.26
		Group II	62.45	6.26	63.05	5.61
	65	Group I	53.95	10.73	52.47	10.28
		Group II	57.10	5.64	56.68	5.86
	90	Group I	45.47	9.83	43.27	9.10

Table continues…

Table continued. …

1000	90	Group II	50.21	11.26	49.93	8.69
2000	50	Group I	60.92	7.12	62.54	6.58
		Group II	62.61	5.76	63.49	6.19
	65	Group I	59.33	7.87	67.58	6.81
		Group II	57.51	6.57	58.90	5.84
	90	Group I	55.58	7.96	56.50	7.28
		Group II	53.34	10.72	55.12	8.05
4000	50	Group I	36.97	5.62	38.76	4.91
		Group II	41.28	11.12	41.48	11.46
	65	Group I	33.49	6.10	32.98	7.25
		Group II	33.09	9.58	32.77	9.45
	90	Group I	26.41	8.25	24.1	8.19
		Group II	29.09	8.74	31	7.80
6000	50	Group I	39.83	7.46	42.95	9.89
		Group II	44.12	4.36	45.30	5.52
	65	Group I	38	6.08	37.47	6.85
		Group II	43.38	9.63	44.91	8.93
	90	Group I	35.11	9.07	35.98	9.89
		Group II	35.49	5.10	34.61	6.60

Table 4.2
Comparison of mean and standard deviation (SD) of insertion gain at different frequencies in linear and non-linear modes for ANSI digi speech signals, for Group I and Group II.

Frequency (Hz)	*Input intensity (dB SPL)*	*Groups*	*IG for linear (in dB)*		*IG for non-linear (in dB)*	
			Mean	*SD*	*Mean*	*SD*
200	50	Group I	33.58	9.63	33.96	11.19
		Group II	34.61	5.24	36.57	8.19
	65	Group I	32.98	7.00	31.10	6.31
		Group II	35.26	7.17	34.04	6.59
	90	Group I	33.67	9.72	33.67	9.72
		Group II	32.62	7.14	32.62	7.14
500	50	Group I	48.22	10.41	48.54	8.70
		Group II	48.95	7.74	52.08	6.70
	65	Group I	43.80	12.32	45.49	10.61
		Group II	47.08	6.48	46.53	7.51
	90	Group I	37.52	9.98	37.53	9.98
		Group II	36.00	5.66	36.00	5.66
1000	50	Group I	56.76	7.37	58.75	6.93
		Group II	60.41	4.79	62.46	5.96
	65	Group I	47.75	12.14	50.35	12.35
		Group II	57.44	6.08	57.48	6.36
	90	Group I	43.63	11.74	44.22	10.91

Table continues…

Table continued…

1000	90	Group II	47.85	6.30	48.45	6.35
2000	50	Group I	60.70	6.18	62.05	5.66
		Group II	62.00	6.58	63.29	5.49
	65	Group I	58.23	7.91	60.06	7.51
		Group II	61.75	7.55	62.52	7.56
	90	Group I	52.13	10.85	53.19	10.63
		Group II	56.75	11.29	59.10	9.95
4000	50	Group I	38.31	9.09	39.80	9.61
		Group II	43.76	10.83	47.27	11.56
	65	Group I	29.60	5.18	29.30	7.24
		Group II	35.40	7.76	35.56	7.61
	90	Group I	27.68	11.10	27.87	11.41
		Group II	32.12	6.74	30.61	9.54
6000	50	Group I	42.90	9.09	44.53	8.63
		Group II	47.63	10.41	50.16	9.35
	65	Group I	34.02	6.44	34.41	6.69
		Group II	37.25	8.20	38.41	7.60
	90	Group I	41.53	8.51	40.83	5.48
		Group II	45.16	12.20	43.19	11.38

From Table 4.1 and 4.2, it can be noted that in both linear and non-linear program modes of the hearing aid, the mean IG at majority of the frequencies were higher for pure tones than that of ANSI digi speech signals. This trend was observed for both the groups.

4.1.2. The mean insertion gain difference between the Group I and Group II across frequencies.

From Tables, 4.1 and 4.2, it can be inferred that differences in mean IG between the two age groups were small. In order to know if these differences were significant, mixed ANOVA was done. The frequencies were grouped into low-, mid- and high- frequencies. The low frequencies consisted of 200 and 500 Hz, the mid frequencies consisted of 1000 and 2000 Hz, whereas, the high frequencies consisted of 4000 and 6000 Hz. This was done for the IG at the low-, mid- and high- frequency- regions in linear and non-linear program modes for pure tone signals, as depicted in Table 4.3.

Table 4.3

IG difference between Group I and Group II across frequencies at 50 dB SPL, 65 dB SPL, and 90 dB SPL, in linear and non-linear program modes, for pure tone signals.

Frequencies	*Intensity level (in dB SPL)*	*Significant difference between Group I and Group II*	
		IG for linear	*IG for non-linear*
Low frequencies	50	F(1,18) = 0.42	F(1,18) =0.69
	65	F(1,18) = 0.00	F(1,18) = 0.34
	90	F(1,18) = 0.63	F(1,18) =0 .52
Mid frequencies	50	F(1,18) = 0.65	F(1,18) =0 .60
	65	F(1,18) = 0.04	F(1,18) = 0.94
	90	F(1,18) = .09	F(1,18) =0 .61
High frequencies	50	F(1,18) = 5.18*	F(1,18) = 1.15
	65	F(1,18) = 0.60	F(1,18) = 1.42
	90	F(1,18) = 0.17	F(1,18) =1.27

Note: *** =** significant difference at $p < 0.05$ level

For pure tone signals, in linear as well as for non-linear program modes, there was no significant difference in the mean IG between the two age groups for pure tone signals at all frequencies, with an exception at 50 dB SPL for high frequencies ($p<0.05$) in linear program mode. This was revealed on the mixed ANOVA.

Similarly, for mean ANSI digi speech signal also, mixed ANOVA was carried out in order to find out whether there was any significant difference in the mean IG between the two age groups. This was done for the low-, mid- and high- frequency- regions at different intensities in linear and non-linear program modes, as depicted in Table 4.4.

Table 4.4

IG difference between Group I and Group II across frequencies at 50 dB SPL, 65 dB SPL, and 90 dB SPL for linear and non-linear program modes for ANSI digi speech signal.

Frequency	*Intensity level (in dB SPL)*	*Significant difference between Group I and Group II*	
		IG for linear	*IG for non-linear*
Low frequencies	50	F(1,18) = 0.06	F(1,18) = 0.69
	65	F(1,18) = 0.34	F(1,18) = 3.78
	90	F(1,18) = 0.13	F(1,18) = 0.08
Mid frequencies	50	F(1,18) = 0.86	F(1,18) = 0.98
	65	F(1,18) = 3.58	F(1,18) = 1.76
	90	F(1,18) = 1.15	F(1,18) = 1.63
High frequencies	50	F(1,18) = 2.13	F(1,18) = 3.78
	65	F(1,18) = 4.76	F(1,18) = 4.33
	90	F(1,18) = 1.49	F(1,18) = 0.53

On mixed ANOVA, for IG for ANSI digi speech signal, there was no significant difference observed in IG for ANSI digi speech signal at any of the frequencies between the two age groups. As, there was no significant difference obtained between the two age groups for insertion gain measures (for both pure tone and ANSI digi speech signals), the data from the two groups were combined for further statistical analyses.

4.2. Mean and Significant difference between the IG for pure tones and ANSI digi speech signals in linear and non-linear program modes:

4.2.1. The mean of IG in linear program mode was obtained and paired t-test was administered in order to know whether there is any significant difference between the mean insertion gain at all three levels for pure tone and ANSI digi speech signals, in linear and non-linear program modes, as depicted in Figures 4.1 and 4.2

Figure 4.1 depicts the mean insertion gain for pure tone signals in linear and non-linear program modes. The mean was computed at three different levels in order to know whether there was a different trend observed for linear and non-linear program modes for pure tone signal.

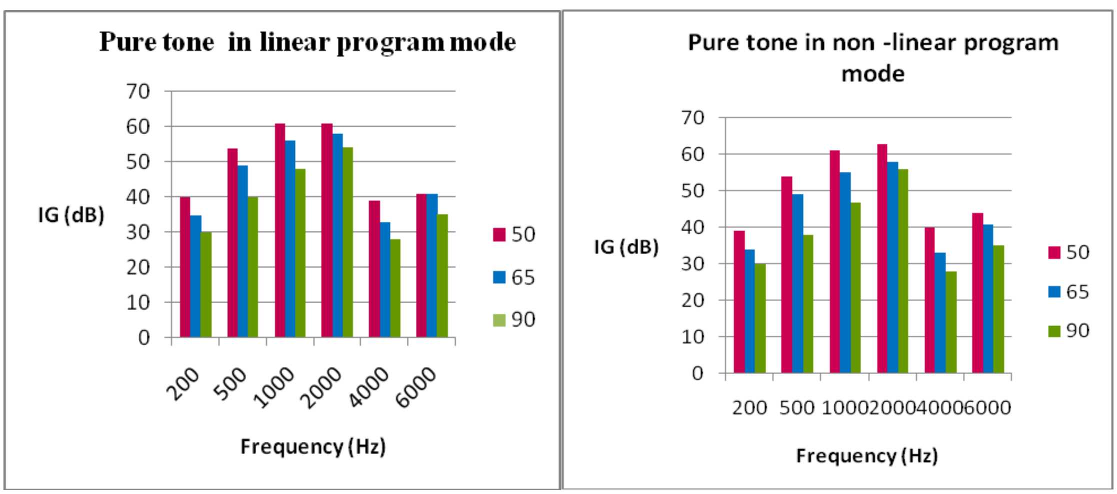

Figure 4.1: Comparison of mean insertion gain mean at different frequencies in linear and non-linear program modes for pure tone signals.

The mean values for IG at lower pure tone levels for non-linear program modes were nearly similar to the linear program mode. A similar trend was observed for moderate and higher signal levels also.

Figure 4.2 depicts the mean insertion gain for ANSI digi speech signals in linear and non-linear program modes. The mean IG was computed at three different levels in order to know whether there was a different trend observed for linear and non-linear program modes for ANSI digi speech signal.

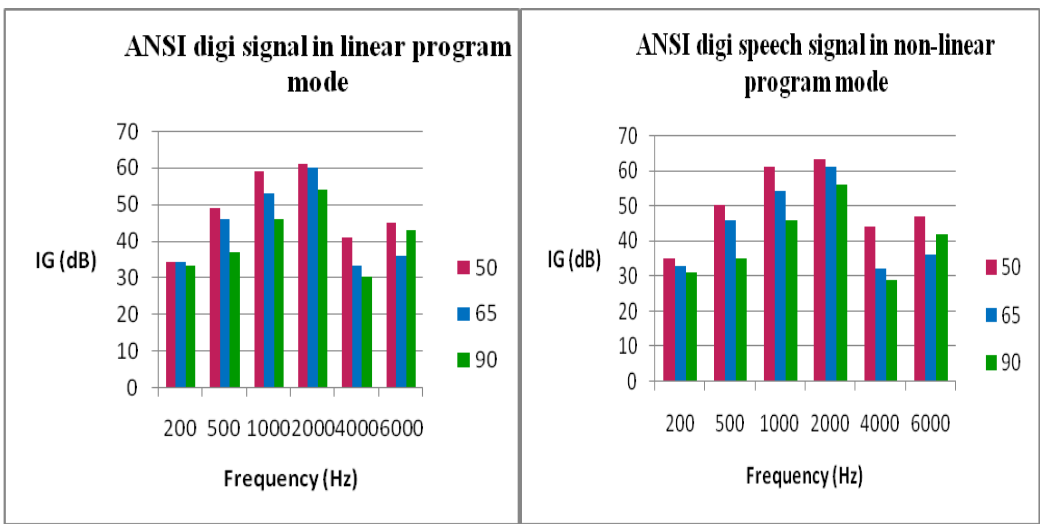

Fig. 4.2: Comparison of mean insertion gain at different frequencies in linear and non-linear program modes for ANSI digi speech signals.

From Figure 4.2, it can be inferred that the mean IG for non-linear program mode is higher than the linear program mode at lower levels of signals. It was a general trend seen across the frequencies. Whereas, the mean IG values were similar for mid and higher levels of ANSI digi speech signals. At low intensity, the insertion gain measure reflects the actual functioning of the non-linear hearing aid, with the low intensity signals provided by more gain.

The paired t-test was administered in order to know whether there was any significant difference obtained for pure tones and ANSI digi speech signals in both the program modes. Table 4.5 shows the significance difference on t-test for mean IG for pure tones and ANSI digi speech signals in linear and non-linear program modes.

Table 4.5

Significant difference on t-test between Insertion Gain (IG) for pure tone signal and ANSI digi speech in linear and non-linear program modes at different frequencies.

Frequency (Hz)	*Input intensity (dBSPL)*	*Significant difference between IG for pure tone and ANSI digi speech*	
		Linear t-value	*Non-linear t-value*
250	50	3.38**	1.46
	65	0.57	1.47
	90	2.08	2.37
500	50	3.25**	2.75*
	65	2.58*	2.91**
	90	2.86**	3.96**
1000	50	2.41*	0.72
	65	2.50*	0.58
	90	1.48	0.24
2000	50	0.38	0.35
	65	1.14	0.96
	90	0.01	0.23
4000	50	1.15	1.78
	65	0.56	0.29
	90	1.08	0.76
6000	50	1.30	1.22
	65	2.66*	2.71*
	90	2.42*	2.58*

Note: *= $p < 0.05$ significant difference, at 0.05 level
**= $p < 0.01$ significant difference, at 0.01 level

In linear program mode, the IG for pure tone signal was significantly different from that of ANSI digi speech signal with 50 dB SPL at 250 and 500 Hz and 1 kHz. The IG for pure tone was different from that of ANSI digi speech signal with 65 dB SPL at 500 Hz, 1 kHz, and 6 kHz. The IG for pure tone was different from that of ANSI digi speech signal with 90 dB SPL at 500 Hz and 6 kHz. At other frequencies there was no significant difference between the pure tone and ANSI digi speech signal even at, 0.05 level of significance.

In non-linear program mode, IG for pure tone was different from that of ANSI digi speech signal with 50 dB SPL at 500 Hz. The IG for pure tone was different from that of ANSI digi speech signal with 65 dB SPL at 500 Hz and 6 kHz. The IG for pure tone was different from that of ANSI digi speech signal with 90 dB SPL at 500 Hz and 6 kHz. At other frequencies, no significant difference between the pure tone and ANSI digi speech signal was found even at 0.05 level of significance. Hence for further analyses, the pure tone and ANSI digi speech signals were analyzed separately.

4.2.2. Paired-t test was administered to find out the difference between mean insertion gain in linear and non-linear program modes for pure tone as well as for the ANSI digi speech signals at the three levels.

Table 4.6 depicts the significant difference between the mean insertion gain in linear and non-linear program modes for pure tone signals. Similarly significant difference for ANSI digi speech signal is also depicted.

Table 4.6

Significant difference on t-test between Insertion Gain (IG) for linear and non-linear program modes for pure tone signal and ANSI digi speech signal at different frequencies.

Frequency (Hz)	*Input intensity (dBSPL)*	*Significant difference between linear and non-linear*	
		for pure tones t value	*for ANSI digi speech t value*
250	50	0.76	0.89
	65	1.01	2.80*
	90	0.73	3.53**
500	50	0.64	1.67
	65	0.32	0.66
	90	2.75*	2.19*
1000	50	0.39	3.85**
	65	0.97	2.16*
	90	1.21	0.68
2000	50	1.93	2.48*
	65	1.06	2.12*
	90	1.49	1.46
4000	50	1.25	3.27**
	65	0.65	0.97
	90	0.41	2.48
6000	50	1.75	3.17**
	65	0.66	0.68
	90	0.66	1.35

Note: *= $p < 0.05$ significant difference, at 0.05 level
**= $p < 0.01$ significant difference, at 0.01 level

For pure tone signal, there was no significant difference observed between insertion gain in non-linear and linear program modes, except at 500 Hz at 90 dB SPL. For ANSI digi speech signals at 50 dB SPL the IG was significantly different for linear and non-linear program modes at 1 kHz, 2 kHz, 4 kHz and 6 kHz. The IG for non-linear was different from that of linear with 65 dB SPL at 250 Hz, 1 kHz, and 2 kHz. At 90 dB SPL the IG for ANSI digi speech signals in linear and non-linear program modes differed only at two frequencies (250 Hz and500 Hz). At other frequencies there was no significant difference between the linear and non-linear even at 0.05 level of significance.

These results suggested that the gain varies for the linear and non-linear program modes with respect to the type of input signal used for the measurement. That is, the difference between the linear and non-linear program modes was higher for ANSI digi speech signals. Whereas, there was no significant difference observed for linear and non-linear program modes for pure tone signals. Thus ANSI digi speech signals are to be used for measurement of insertion gain of hearing aids. For further analyses only ANSI digi speech signal was considered.

4.2.3. The paired t-test was administered to find out whether there was a significant difference between mean insertion gain for low and high intensity signal in linear and non-linear program modes.

The significant difference between the 50 dB SPL and 90 dB SPL was obtained for pure tone signals in linear mode. Similar analysis was carried out for the pure tone signal in non-linear mode, ANSI digi speech signal in linear and non-linear program modes. Table 4.7 depicts the significant difference at softer and louder levels for linear and non-linear program modes for both the signals. Moderate levels of signals were not considered for the analysis because gain provided by the linear and non-linear hearing aid was found to be same at moderate signal levels.

Table 4.7

Significant difference on t-test between insertion gain linear and non-linear program modes at softer and louder levels for pure tone signal and ANSI digi speech signal at different frequencies.

Frequencies (Hz)	*Significant difference for ANSI digi speech for 50 dB SPL and 90 dB SPL*		*Significant difference for pure tones at 50 dB SPL and 90 dB SPL*	
	Linear	*Non-linear*	*Linear*	*Non-linear*
200	5.12***	5.50***	1.13**	3.24**
500	8.27***	10.81***	8.25***	14.67***
1000	10.11***	13.70***	10.97***	13.92***
2000	3.18***	4.01**	3.06***	2.85***
4000	6.24***	5.78***	3.64***	4.79**8
6000	4.13***	4.26***	0.72**	2.26**

Note: ** Significant difference at 0.01 level
*** Significant difference at 0.001 level

Results indicated that in both the program modes (the linear and non-linear program modes), the gain provided at low and high intensity levels were very significantly different for pure tone signals. Whereas, similar kind of trend was observed for ANSI digi speech signal also in linear as well as non-linear program modes.

4.3. Difference between FG and IG:

4.3.1. The difference between the mean FG and IG was analyzed using descriptive statistics and test for significant difference:

The difference between the FG and IG at three intensity levels were computed across the frequencies to find out the mean and SD for linear as well as non-linear program modes. The comparison of mean and SD values for the difference in FG and IG, in linear and non-linear modes, across frequencies are depicted in Figures 4.3 and 4.4, and Table 4.7.

4.3.1.1. Mean and SD for the difference between the mean functional gain and insertion gain at three different levels, for linear mode, is given in Figure 4.1 and Table 4.7.

The difference between FG and IG were analyzed. The mean and standard deviation (SD) of this difference were obtained at 500 Hz, 1000 Hz , 2000 Hz and 4000 Hz at one level of functional gain and all three levels of signal for insertion gain including 50 dB SPL, 65 dB SPL, and 90 dB SPL for pure tone signals in linear program mode. At each of the three input levels of IG, the difference of FG and IG was obtained. The mean and standard deviation was calculated for the difference between FG and IG.

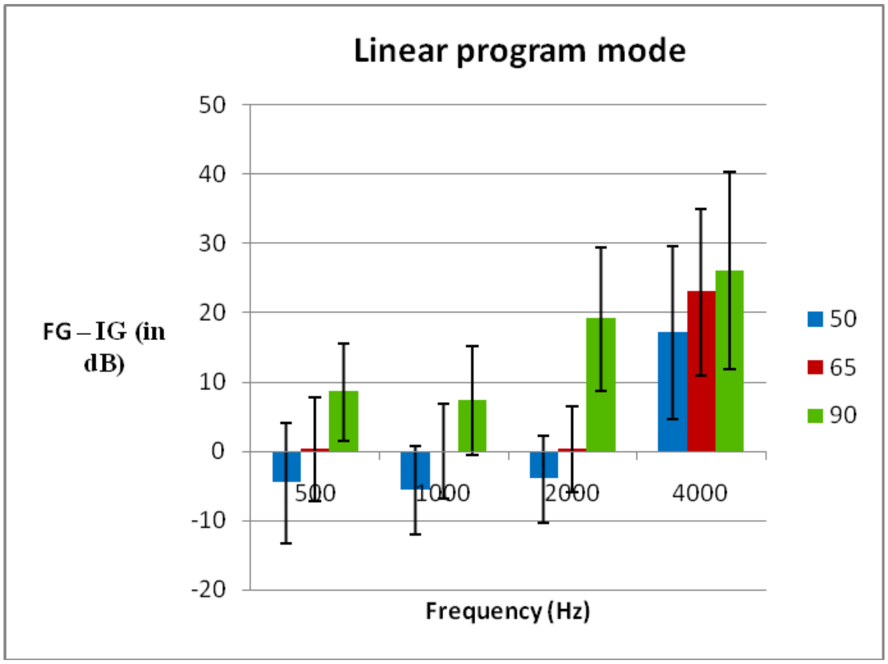

Fig. 4.3: Mean and standard deviation of the difference between FG and IG for pure tone signals at 50 dB SPL, 65 dB SPL, and 90 dB SPL, across frequencies for linear program mode.

Figure 4.3 and Table 4.8 depict the difference between functional gain and insertion gain at 50 dBSPL, 65 dB SPL, and 90 dB SPL at four frequencies 500 Hz, 1000 Hz, 2000 Hz, 4000 Hz. The negative value for non-linear program mode at 50 dB SPL suggests that insertion gain exceeded the functional gain at three frequencies (500 Hz, 1000 Hz, and 2000 Hz) Whereas, at 4000 Hz the mean functional gain value exceeded the mean insertion gain value.

The difference between the functional gain and insertion gain at 65 dB SPL for 500 Hz, 1000 Hz, and 2000 Hz was very minimal for linear program mode. Whereas, the difference between FG and IG was much more at 4000 Hz. The difference between the FG and IG at 90 dB SPL was a positive value, depicting that the FG values exceeded the

IG at all the four frequencies (500 Hz, 1000 Hz, 2000 Hz, & 4000 Hz) for linear program mode. At 4000 Hz., as the intensity increased, the difference between the FG and IG also increased. This implied that FG and IG cannot be used as substitute for one another, especially at higher ferquencies.

Table 4.8

Mean and standard deviation of the difference between the FG and IG for pure tone signals at 50 dB SPL, 65 dB SPL, and 90 dB SPL across the frequencies for linear and non-linear program modes.

Frequency (Hz)	*Intensity level (in dB SPL)*	*FG – IG Mean (SD)*	
		Linear	*Non-linear*
500	50	-4.51 (8.69)	-4.92 (8.41)
	65	0.25 (7.51)	0.46 (7.82)
	90	8.56 (7.02)	12.6 (6.37)
1000	50	-5.06 (6.43)	-6.03 (6.49)
	65	0.03 (6.79)	0.68 (8.02)
	90	7.34 (7.79)	11.44 (8.25)
2000	50	-4.01 (6.23)	-5.26 (7.46)
	65	0.33 (6.19)	8.8 (5.91)
	90	19.10 (10.35)	2.11 (9.76)
4000	50	17.13 (12.47)	16.13 (11.48)
	65	22.96 (12.0)	23.67 (12.89)
	90	26.0 (14.22)	30.23 (15.22)

4.3.1.2. Mean and SD for the difference between the mean functional gain and insertion gain at three different levels, for non-linear program mode.

The difference between the FG and IG at 50 dB SPL, 65 dB SPL, and 90 dB SPL were computed across the frequencies to find out the mean and SD for non-linear program mode. The comparison of mean and SD values for FG – IG non-linear program mode across frequencies are depicted in Figure 4.2 and Table 4.8.

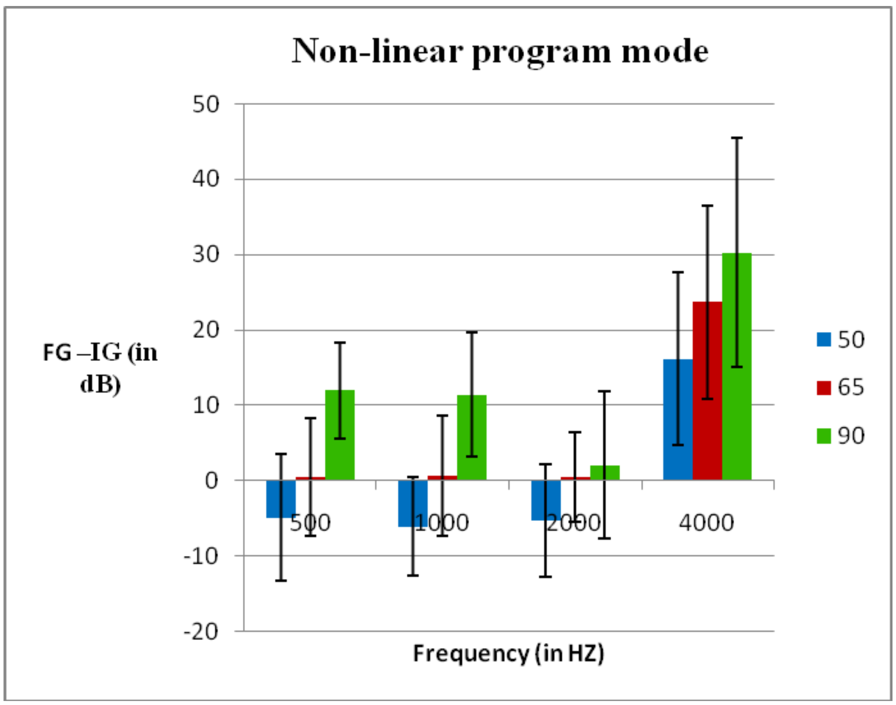

Fig. 4.4: Mean and standard deviation of the difference between the FG and IG for pure tone signals at 50 dB SPL, 65 dB SPL, and 90 dB SPL, across the frequencies, for non-linear program mode.

Figure 4.4 depicts the difference between mean functional gain and mean insertion gain at 50 dB SPL, 65 dB SPL, and 90 dB SPL at four frequencies 500 Hz, 1000 Hz, 2000 Hz, 4000 Hz for non-linear program mode. The negative value of the difference between FG and IG at 50 dB SPL suggests that the insertion gain exceeded the functional gain at three frequencies (500 Hz, 1000 Hz, and 2000 Hz). Whereas, at 4000 Hz, the mean functional gain mean insertion gain values at all the three input values. The reson for this can be squashed the probe tube which results in attenuation of high frequency response. The difference between the functional gain and insertion gain at 65 dB SPL for 500 Hz, 1000 Hz, and 2000 Hz was very minimal for non-linear program mode also. The FG and IG difference exceeded upto 22.96 dB at 4000 Hz which was much higher compare to the other frequencies. For pure tone signals, the difference between the FG and IG at 90 dB SPL was positive, indicating that the FG values exceeded the IG at all the four frequencies (500 Hz, 1000 Hz, 2000 Hz, & 4000 Hz) .

The difference between FG and IG is lesser in linear program mode, this indicates that FG and IG are parallel measures for the linear hearing aid. The difference being more for non-linear program mode indicates that the FG and IG are not similar measures. Further, for non-linear program mode, the IG is a more realistic measure as the IG decreased with increase in input intensity.

The IG measure is an important measure because it reflects the functioning of non-linear mode of the hearing aid. The IG measure reflected more gain for soft level of signal and moderate gain for moderate level of signal and lesser gain for higher level of signal.

4.3.2. Significant difference in the mean values of the difference between FG and IG at different frequencies for linear and non-linear program modes.

To find out if FG and IG difference was significant, paired t-test was administered. Significant difference between Functional Gain (FG) and Insertion Gain (IG), for linear and non-linear program mode at different frequencies is shown in Table 4.9.

Table 4.9

Significant difference between Functional Gain (FG) and Insertion Gain (IG) for linear and non-linear program modes for pure tone signals at different frequencies.

Frequency (Hz)	*Input intensity (dBSPL)*	*Difference in FG and IG (t-values)*	
		Linear	*Non-linear*
250	50	2.49*	3.71**
	65	5.59**	8.23**
	90	6.55**	8.65**
500	50	2.32*	1.59
	65	0.15	0.94
	90	5.76**	9.01**
1000	50	3.89**	2.25*
	65	0.17	1.64
	90	4.25**	6.04**

Table continues….

Table continued….

2000	50	2.85*	2.93**
	65	0.36	0.93
	90	1.42	1.35
4000	50	6.21**	6.5**
	65	8.55**	8.45**
	90	8.96**	8.87**
6000	50	7.0**	5.88**
	65	7.47**	6.18**
	90	7.71**	6.47**

Note: * = Significant difference at p < 0.05 level
** = significant difference at p < 0.01 level

Significant difference in FG and IG for different frequencies in linear and non-linear program modes are given below:

In the present study for linear hearing aids, it was noted that there was a significant difference between the FG and the IG at low and high levels for pure tones. At moderate levels, there was no significant difference between the FG and the IG at 500, 1000 and 2000 Hz. For non-linear hearing aids, it was noted that there was a significant difference between the FG and the IG at low and high levels for pure tones. At moderate levels, there was no significant difference between the FG and the IG at 500, 1000 and 2000 Hz.

4.4. Relationship between SIS and AI

For linear mode, the relationship between Speech Identification Scores (SIS) at three levels (SIS_{35}, SIS_{50} dB, & SIS_{75} dB) with that of articulation index computed from FG measure (AI_{FG}) and articulation index computed from IG measure (AI_{IG}) was investigated. This was done only for ANSI digi speech signals as its relationship with SIS was being analyzed. This was also done for non-linear program mode, as shown in Table 4.9. On Pearson's correlation analysis, the correlation was higher with AI_{FG} than with AI_{IG} . in both the program modes

Table 4.10

Correlation of Articulation index from the FG (AI_{FG}) and IG (AI_{IG}), in linear and non-linear program modes, with Speech Identification Scores (SIS) at different levels.

Pearson Correlation between		*Linear*	*Non-linear*
SIS_{35}	AI_{FG}	r = 0.59*	r= 0.58*
	$AI_{IG,50}$	r = 0.40	r = 0.53*
SIS_{50}	AI_{FG}	r = 0.39	r= 0.42
	$AI_{IG,65}$	r = 0.16	r = 0.26
SIS_{75}	AI_{FG}	r =0.36	r = 0.10
	$AI_{IG,90}$	r = 0.08	r = 0.04

Note: * = Significant correlation at $p < 0.05$ level

4.4.1. Relationship between SIS and AI_{FG} computed from functional gain measure (AI_{FG}):

On performing Pearson's correlation, the results indicated a significant correlation between SIS_{35} and AI_{FG} in linear program mode. Whereas, there was no significant correlation obtained for SIS_{50} with AI_{FG} and SIS $_{75}$ with AI_{FG} in linear program mode.

A significant correlation was obtained between SIS_{35} and AI_{FG} in non-linear program mode. Whereas, there was no significant correlation obtained for SIS at other levels and $AI_{FG.}$ The overall trend was similar in both the program modes.

4.4.2. Relationship between SIS and AI_{IG} through insertion gain measure (AI_{IG}):

On performing Pearson's correlation, in linear program mode, though there was a positive correlation, it was not significant ($p>0.05$). For non-linear mode, a significant correlation between SIS_{35} and $AI_{IG,50}$ was noted. Whereas, there was no significant correlation obtained for SIS with AI_{IG} at other levels.

To summarize the findings of the present study:

1) Mean insertion gain was higher for pure tone signal than ANSI digi speech signals for both the program modes at all the three intensity levels.

2) Insertion gain was not significantly different for 4+ to 5 years age group and 5+ to 6 years age group. Thus, for further analyses of data the groups were analyzed.

3) Mean insertion gain was higher for low signal levels than the moderate and high signal levels for linear program mode for both the input signals. Similar trend was observed for non-linear program mode also.

4) Insertion gain for pure tones and ANSI digi speech signal was found to be significantly different in linear program mode which did not hold good for non-linear program mode.

5) There was a significant difference observed between linear and non-linear program modes for ANSI digi speech signal which was not true for the pure tone signals.

6) At different levels, there was no significant difference found for linear and non-linear program modes for both the signals (pure tone and ANSI digi speech signal).

7) Functional gain and insertion gain were found to have no significant difference if the insertion gain is carried out at moderate signal levels. But function gain and insertion gain were significantly different when the insertion gain was carried out at low and moderate signal levels.

8) A significant positive correlation was obtained between SIS_{35} and AI_{FG} in linear program mode. Similar correlation was obtained for SIS_{35} and AI_{FG} in non-linear program mode also.

9) A significant correlation between SIS_{35} and $AI_{IG,50}$ was obtained only for non-linear program mode

Chapter 5

DISCUSSION

5.1. IG for pure tone and ANSI digi speech signals in linear and non-linear program modes.

5.1.1. The insertion gain between the Group I (4 + to 5 years) and Group II (5+ to 6 years) of participants were not significantly different across frequencies consisting of low-, mid- and high- frequencies for pure tone signals as well as ANSI digi speech signals.

Ching and Dillon (2003) reported that the length of the ear canal determines its resonance properties, which determines the real-ear unaided gain (REUG) of the ear, and hence the real-ear threshold. As, a child grows from 1 month to 3 years, the resonance peak in REUG moves from around 6 kHz to 3 kHz (Kruger, 1987). Therefore, the child's hearing thresholds will deteriorate at 6 kHz, but improves at 3 kHz over this time, assuming that there are no other maturation effects. Bentler (1989) reported that the average external ear resonance characteristics for children (three to thirteen years of age) appeared to be similar as adults, but some small differences were noted above 3000 Hz. The difference in measured SPL between adults and children were 3-5 dB.

Whereas Seewald, Cornelisse, and Ramiji (1997) had reported the age related differences in the SPL in the ear canal for children from the birth to 7-years of age. The current study findings suggested no significant difference in SPL for 4 + to 5 years and

5+ to 6 years. This might be attributed to the maturational changes in the resonance properties of the external ear not being significant during the four to six years of age.

5.2. The insertion gain between the IG of Group I and Group II for the two types of signals, pure tone and ANSI digi speech signal were significantly different for certain frequencies. For ANSI digi speech signals the linear and non-linear program modes insertion gain differed significantly. Whereas, for pure tone signals not much difference was obtained between both the program modes. For ANSI digi speech signal and pure tone signal in linear program mode there was a significant difference obtained which suggested that pure tone and ANSI digi speech cannot be substituted for one another. But in non-linear program mode it was observed that the difference between the ANSI digi speech signal and pure tone signal was significant at very few frequencies.

It can be attributed to the Noise Tracker of the hearing aid which was switched off even during the non-linear program mode measurements, which in turn affects the performance of the hearing aid in the non-linear program mode. The other factor which may have contributed to the less significant difference can be the degree of hearing loss when the degree of the hearing loss is more it’s possible that both the linear and non-linear hearing aids provide equivalent amount of gain.

Leijon (2002) demonstrated that for all WDRC hearing aids, the gain and frequency responses of pure tone signals were clearly different from broad band speech-like signals (at three levels 55, 65, or 75 dBSPL). The differences were of the order of 5 to 10 dB depending on the test signals.

Scollie and Seewald (2002) showed that electroacoustic test signals differ in their ability to match the aided levels of real speech at three levels (50 dB SPL, 70 dB SPL, & 85 dB SPL). In general, the speech-weighted signals provide a closer match to aided speech levels, than constant-level pure tone sweeps, which tend to overestimate aided output. The findings also reported that aided levels of pure tone signals should not be used to estimate the aided levels of real speech in sound pressure level. The inaccuracy is primarily due to the large difference in input levels between conventional pure tone sweep and real speech across frequencies.

The current study also supported the view that the insertion gain for pure tone and ANSI digi speech signal are significantly different because of their different temporal and spectral characteristics.

5.3. The mean of FG and IG difference in linear and non-linear modes was analyzed.

5.3.1. The difference between the mean functional gain and insertion gain at different intensity levels (50 dBSPL, 65 dBSPL, & 90 dBSPL) were compared. The functional gain obtained at one level was compared with insertion gain at three levels.

For the soft level of IG, FG - IG difference was in negative values. This indicated that the IG values were more than the functional gain at low signal levels, which was not the same for moderate and high levels of signal. The difference in the FG and IG at 50 dBSPL for non-linear program mode was more at 500, 1000, and 2000 Hz when compared with linear program mode. This might be because the insertion gain provided more gain in the non-linear than in the linear program mode, at low signal levels. Thus, it indicates that functional gain cannot be used to predict the insertion gain at soft level of signals as the difference between the functinal gain and insertion gain was more at low signal levels.Whereas, for linear as well as non-linear program modes, the FG - IG at 50 dBSPL for 4000 Hz difference was positive indicating that FG at 4000 Hz is always higher than the predicted IG measured. The reason might be underestimation of the gain by insertion gain measurement at 4000 Hz where FG is much higher than the IG.

FG and IG difference at 65 dBSPL in the current study were within 8 dB which was close to 5 dB as reported by Mason and Popelka, (1986). This was observed for three frequencies i.e., 500 Hz, 1000 Hz and 2000 Hz in linear program mode. Cuda, De Benedetto, and Leante, (1992) found similar results. But there was a high variability at lower frequencies found occasionally, which was probably due to sealing problems of ear molds in the ear canal. For non-linear program mode also, results correlated well for FG and IG difference at 65 dBSPL at 500 Hz 1000 Hz, and 2000 Hz. That is FG can be predicted or substituted by IG if the IG measurement is carried out at moderate level of signals. In other words IG and FG provide similar measurements at moderate levels of signals.

For 4000 Hz, the FG and IG difference exceeded and it was found to be above 10 dB, for both linear and non-linear program modes, which conforms to that reported by Harford, (1981). But, for the diference obtained between FG and IG was only 6 dB at 4000 Hz. This might be due to underestimation of the gain by insertion gain measurement at 4000 Hz where FG is much higher than the IG. Dillon and Murray (1987) reported that the probe tube itself can be squashed and its high frequency response can be attenuated.

For higher levels of signals, FG and IG difference at 90 dBSPL was positive depicting that FG value exceeded the IG at all the four frequencies (500 Hz, 1000 Hz, 2000 Hz, 4000 Hz) for linear as well as non-linear program mode indicating that FG and IG measurements interchanged. The difference between FG and IG was found to be less for linear program mode at 500, 1000 Hz and 4000 Hz compared to non-linear program mode. Whereas, for 2000 Hz there was a different trend seen as linear program mode FG-IG exceeded that in the non-linear one.

The difference between FG and IG is less in linear program mode indicating that the FG and IG are similar measures especially if the IG measurement is carried out at moderate signal levels. This difference is more for non-linear program mode indicating that the FG and IG are not similar measures. That is, the difference between the FG and IG in non-linear program mode exceeded that in the linear program mode values even at moderate level of signals.

Further, IG is a more realistic measure as IG decreases with increase in input intensity. This is because, in the non-linear hearing aids, there in decrease in gain with increase in input level. This cannot be measured or reflected through the FG. FG is mainly a measure which predicts the gain at low levels (at threshold) or moderate levels of signal. So, the amount of gain provided at high level of signals cannot be measured through FG. As, the results depicted that the difference between the FG and IG is more at higher signal levels across the frequencies, it is Suggested that, FG and IG measures cannot be substituted for each other at low and high signal levels. It provides insight to the fact that for evaluation of the hearing aid performance at higher signal levels, insertion gain is a more realistic measure, which can reflect the non-linear gain.

5.3.2. Significant difference between Functional Gain (FG) and Insertion Gain (IG) for linear and non-linear program modes at different frequencies.

In the current study, the FG correlated well with the IG at average conversational level as expected, but not with the soft or the louder level of signals for linear as well as non-linear program modes. The trend seen was similar to the trend observed in the FG and IG difference.

Jenstad, Seewald , Cornelisse, and Shantz (1999) reported that speech intelligibility testing and loudness rating carried out for linear as well as WDRC resulted in equivalent comfort and intelligibility for average input levels. The results of the present study support this finding. Whereas, Kuk, Keenan, Lau, and Ludvigsen (2004) showed that the aided sound-field threshold (ASFT) represents the softest sound that the wearer can hear inside the audiometric test booth when using a non-linear hearing aid.

For higher frequencies, it was found that FG was considerably different from IG at three input levels for linear as well as for non-linear program modes. Suggested, that IG at any level (50 dBSPL 65dBSPL, and 90 dBSPL) cannot be used as a substitute for functional gain. Both the measures need to be evaluated independently for high frequency FG or IG measurements.

Stelmachowicz and Lewis (1988) mentioned that validity of FG or IG depends on the circumstances of its measurement. If the hearing loss is profound, then the responses may be vibratory, and functional gain estimates should supplement, or even replace, insertion gain measures. In the case of a non-linear hearing aid or a hearing aid in saturation, insertion-gain measures are more valid than functional-gain measures only because we are forced to obtain these measures at relatively high levels due to the internal noise of measuring equipment and/or undesirable room noise. Fortunately, a 60 dB SPL pure-tone input corresponds fairly well to most estimates of the average long-term speech spectrum.

In the present study, the conversational level insertion gain is the best predictor of the functional gain in linear as well as non-linear program modes, as the correlation of FG and IG was best at conversational level. But, it’s not always true with non-linear hearing aids. Because the gain provided by the non-linear hearing aid is considerably high at soft signal a level which does not hold good for linear hearing aids as linear hearing aids provide equal gain at all input levels.

5.4. Relationship between SIS and AI

5.4.1. Articulation index from functional gain measure (AI_{FG}) in linear program mode correlated with Speech Identification Scores (SIS) at 35 Db HL in linear program mode. Similarly (AI_{IG}) in non-linear program mode correlated with Speech Identification Scores (SIS) at 35 dB HL in non-linear program mode.

Jenstad, Seewald , Cornelisse, and Shantz 1999; Marriage and Moore 2003 reported that the linear hearing aid as well as WDRC (non-linear) hearing aids provided more gain at low input levels soft speech than for speech at moderate level. But the processing type and presentation level were not statistically significant in most of the participants. The reason for this might be that WDRC used in their study was single channel and children with profound hearing loss were using hearing aid with linear amplification strategy so, they were not given time for acclimatization with the non-linear hearing amplification strategy.

Souza, Jenstad, and Folino (2005) reported that there was a significantly poorer performance with three channel WDRC system compared with a linear with compression limiting system for adults with severe hearing impairment. These contradictory results might have been obtained because a single constant compression ratio of 3:1 was used across the channels. Boothroyd, Springer, Smith, and Schulman (1988) also reported similar results for children with profound degree of hearing loss.

The present study indicated that AI from functional gain correlated better with the lower level of SIS in non-linear as well as in linear program modes, indicating that AI at soft levels can be a predictor for SIS at soft levels.

5.4.2 Insertion gain AI (AI_{IG}) at three levels and SIS at three levels in linear and non-linear program modes

In the present study, AI from IG at low levels (50 dBSPL) correlated well with the SIS at 35 dBHL in non-linear program mode. Whereas, there was no significant correlation found for linear program mode between AI_{IG} and SIS at any of the levels.

Scollie and Seewald (2002) suggested that the match between the aided test signal and aided speech was different for high level of signals. For the composite signal, the tests at high intensities tended to underestimate the aided speech levels, primarily in the mid- to high-frequency region for linear as well non-linear hearing aids.

Dillon (1993) reported that speech gain in quiet provided by a hearing aid can be accurately predicted from electroacoustic information comprising of the participant’s thresholds, internal hearing aid noise and, and the hearing aid’s insertion gain for mild to moderate degree of hearing loss. But, as the hearing loss increases, the distortions such as reduced frequency and temporal resolution makes it less likely that audible energy will continue to be equally useful. This might be the reason that the current study findings of the IG did not correlate well with the SIS because as the degree of hearing loss increases, the frequency and the temporal resolution becomes poorer. And also with increase in the degree of hearing loss, more amount of gain is required which in turn induces distortion.

It also suggested that the success of the AI method in prediction of speech gain implies that additional speech audibility afforded by the hearing aid carried the relative importance predicted by the importance function. Results indicated that in children with moderate to profound degree of hearing loss the IG measures are not good predictor of the speech measures.

The results suggest that the IG measures are same for two age groups. But, the insertion gain values differed for different types of signals and for different program modes. As the spectral and temporal characteristic of ANSI digi speech signal is similar to speech, it was a more realistic stimulus to evaluate the hearing aid performance. The ANSI digi speech signal was found to be significantly different in linear and non-linear program modes, indicating that it is able to reveal the difference obtained in different program modes (i.e., linear and non-linear) which was not obtained by use of the pure tone signals.

The FG and the IG differences were significantly larger at low and high signal intensity, indicating that the FG and IG cannot be substituted for each other, if the IG measurement is carried out at these signal levels for linear and non-linear program modes. Whereas, the FG and IG can be substituted for one another if IG measurement is carried out at moderate signal levels for both the program modes. It must however be noted that for predicting benefit for speech identification, the lower level of IG can be a better predictor and not the moderate or the higher signal levels. This finding was consistent only with non-linear program mode.

There was no correlation obtained between the speech measures and IG measures at any of the levels in linear program mode. The functional gain can be used to predict the speech gain only at lower speech levels in for linear as well as non-linear program modes. But this did not hold good for moderate and higher levels of speech signals. Thus, the findings of the study implied that the FG and IG should be used as two separate measures in order to know the hearing aid performance for different types of signals and at different levels of signal.

Chapter 6

SUMMARY AND CONCLUSIONS

Hearing aid selection and fitting is a step-wise procedure involving hearing evaluation, pre-selection of hearing aid, hearing aid fitting, verification of hearing aid, and validation. Methods for verification of hearing aid fitting differ with respect to the type of processing strategy used in the hearing aid. When assessing the suitability of different signal processing strategies for children who have hearing impairment from early life, , it is important to evaluate the strategies using a representative sample of those children, rather than generalizing from results derived from adult subjects with acquired hearing loss. Therefore, there is a need to evaluate the subjective and objective verification measures for the linear and non-linear hearing aids, as they provide different amount of gain with the different input levels. Hence, the present study attempted to compare the functional gain and insertion gain of hearing aids to see if one can be used instead of the other for verification of hearing aid fittings using different technologies (Hawkins, 2004).

Thus, the objectives of the study were -

1) To compare the insertion gain (IG) of hearing aid across the age groups.
2) To compare the different types of signals used for insertion gain measurement.
3) To compare linear and non-linear program modes for insertion gain measurement.

4) To compare insertion gain (IG) and functional gain (FG) for linear and non-linear hearing aids.

5) To investigate the relationship between the speech identification scores and the articulation index derived from the insertion and functional gain measures.

To evaluate the effectiveness of the verification measures, 20 children with hearing impairment using hearing aids participated in the study. The children, in two age groups, Group I with 4+ to 5 years, and Group II with 5+ to 6 years, were included in the study. They used a range of different models of Behind-The-Ear (BTE) hearing aids. The hearing loss ranged from moderately severe to profound hearing loss. These children were attending therapy and were at a stage that they could at least perform word identification task. The test hearing aid was a digital BTE hearing aid which could be programmed for linear as well as non-linear program modes.

The study was conducted in three stages –

Stage I: Optimization of hearing aid parameters was done for non-linear and linear program modes. The NAL-R was used for linear program mode and NAL-NL1 was used for non- linear program mode. The former hearing aid setting was stored in Program 1 (P1) and the latter hearing aid setting was stored in Program 2 (P2) of the test hearing aid.

Stage II: Verification of hearing aid fitting through insertion gain (IG) measurement.

Stage III: Verification of hearing aid fitting through functional gain (FG) measurement

Thus, for each of the participant, the following data were collected after the hearing aid was optimized for NAL-R and NAL-NL1 prescriptive procedures:

1) From the insertion gain measurement:

The REIG for pure tone and ANSI digi speech signal, across the different frequencies at three levels (50 dB SPL, 65 dB SPL & 90 dB SPL) was measured. This was done in linear as well as non-linear program modes. The articulation index (AI) was also calculated from REIG at three levels, in linear and non-linear program modes, using Count-the-dots procedure (Mueller & Killion, 1990).

2) From the functional gain measurement:

a) The functional gain across frequencies for linear as well as non-linear program modes. This was converted to AI for linear as well as non-linear program modes, using Count-the-dots procedure (Mueller & Killion, 1990).

b) The SIS at 35, 50, and 75 dB SPL.

The results of the study indicated that:

1) The insertion gain between the Group I (4 + to 5years) and Group II (5+ to 6 years) of participants was not significantly different across frequencies consisting of low, mid- and high- frequencies, for pure tone signals and ANSI digi speech signals.

2) The mean insertion gain was significantly higher for pure tone signals than that of ANSI digi speech signals. The mean insertion gain between the two types of signals, pure tone and ANSI digi speech signal, were significantly different at 500 Hz, 1000 Hz, 2000 Hz, and 6000 Hz. The mean insertion gain was statistically different for linear and non-linear program modes for pure tone and ANSI digi speech signals.

3) For different levels, the difference in mean IG was not significant for linear as well as non-linear program modes, for pure tone as well as ANSI digi speech signal.

4) There was no significant difference was obtained for functional gain and insertion gain at moderate signal levels for both linear as well as non-linear program mode, which did not hold good for lower and higher levels in both the program modes.

5) The difference between the mean FG and IG was more for low and high levels of signal for non-linear program mode than for linear program mode.

6) The AI from functional gain (AI_{FG}) correlated well with the SIS at low level for both the program modes. Whereas, AI_{FG} did not correlate significantly with the SIS at moderate and high levels.

7) The AI from insertion gain (AI_{IG}) correlated significantly with the SIS at low level, for non-linear program mode. Whereas, AI_{IG} at moderate and high level did not significantly correlate with SIS at moderate and high insertion gain levels for non-linear program mode. For linear program mode none of the AI levels significantly correlated with any of the SIS levels.

Conclusions

The IG and FG can be used as verification measures, for linear hearing aids. This is because both of them provide comparative results for linear hearing aids. However, the values of FG and IG were different for non-linear hearing aids. Moreover, the IG measures can be carried out at different levels which provide a better estimation of gain across the frequencies. This is important for evaluating the performance of non-linear hearing aids as it functions differently at different input levels.

For ANSI digi speech signal, the IG were significantly different for linear and non-linear hearing aid indicating that the insertion gain provided by ANSI digi speech differ significantly across the program modes, whereas, this was not revealed by the pure tone signal. For pure tone signal the insertion gain was similar for linear and non-linear program modes. So, ANSI digi speech signal is a better measure to predict the performance for linear as well as non-linear hearing aid.

For verification of linear and non-linear fitting, the difference between FG and IG was least for moderate level of signals. This suggested that both the measures can be used for verification, if performance of the hearing aid needs to be verified for moderate signal levels. At low and high levels, the difference between FG and IG was more for non-linear program mode compared to linear program mode indicating that the FG and IG should be used as two separate measures. The IG being a better reflector of the hearing aid performance at low and high levels, verification would be effective if performed with IG measure.

The AI from functional gain can be used to predict the SIS for soft signal levels (at 35 dB HL), for linear as well as non-linear program modes. Whereas, AI from functional gain is not a good predictor of SIS at moderate and higher levels (50 dB HL & 75 dB HL).

Clinical Implications:

From the results of the present study the following implications can be derived:

1) The insertion gain measure can be used as an important tool in order to verify the hearing aid fittings, especially for non-linear hearing aids.
2) Insertion gain can be used as a realistic tool for predicting the hearing aid gain at different signal levels (soft, moderate & loud).
3) As the IG for ANSI digi speech provides more realistic information about real speech, this type of signal should be preferred for verification.
4) Functional gain and insertion both can be used to evaluate the children's performance with the hearing aid at moderate signal levels as they are comparable at moderate levels.
5) The AI from FG measure can be used to predict the SIS, if the SIS is done at low intensity level.

Future research

Future research can focus on evaluating the linear and non-linear hearing aids through both the verification measures (functional gain and insertion gain) for children with younger age group (below four years of age). This is because, the present study did not reveal any significant difference in IG for the two age groups i. e., 4+ to 5 and 5+ to 6 years of age. Study can be conducted with more number of participants across different age group in order to know the age related trend seen for insertion gain measure. Along with the insertion gain and SIS the aided thresholds can also be used as a verification measure for hearing aid fitting in children.

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APPENDIX

Picture identification test for Kannada speaking children (Vandana, 1998)

List I	
lota	me:dzu
e:ni	ili
tʃa:ku	su:dzi
bassu	tale
gu:be	kivi
kattu	pennu
la:ri	mara
mane	bale
nalli	ka:lu
me:ke	gante
mola	sara
ka:ge	tʃendu
se:bu	

List II	
ka:ru	ka:su
o:le	su:rya
a:ne	ni:ru
Tatte	Ele
Gini	Railu
ha:vu	bi:ga
na:ji	ko:li
hallu	hu:vu
mu:gu	hasu
Male	dha:ra:
Kappe	tʃ$_h$tri
kannu	tʃi:la
mi:nu	

APPENDIX

List I	List II
lota	me:dzu
e:ni	ili
tʃa:ku	su:dzi
bassu	tale
gu:be	kivi
kattu	pennu
la:ri	mara
mane	bale
nalli	ka:lu
me:ke	gante
mola	sara
ka:ge	tʃendu
se:bu	railu
bi:ga	ka:ru
ko:li	o:le
hu:vu	a:ne
mu:gu	tatte
hasu	gini
male	ha:vu
kappe	na:ji
kannu	hallu
dha:ra:	ka:su
tʃ$_{h}$tri	su:rya
tʃi:la	ni:ru
mi:nu	ele

APPENDIX

List I	
lota	bi:ga
e:ni	ko:li
tʃa:ku	hu:vu
bassu	mu:gu
gu:be	Hasu
Kattu	Male
la:ri	Kappe
mane	Kannu
nalli	dha:ra:
me:ke	tʃ$_{h}$tri
mola	tʃi:la
ka:ge	mi:nu
se:bu	

List II	
me:dzu	ka:ru
ili	o:le
su:dzi	a:ne
tale	tatte
kivi	gini
Pennu	ha:vu
Mara	na:ji
bale	hallu
ka:lu	ka:su
gante	su:rya
Sara	ni:ru
tʃendu	ele
railu	