SENSORINEURAL HEARING LOSS: LINEAR VS NONLINEAR CIRCUITRY FITTING

REGISTER NO. M 9923

A dissertation submitted in part fulfilment of the Final year M.Sc, (Speech & Hearing), Mysore.

ALL INDIA INSTITUTE OF SPEECH AND HEARING, MYSORE - 570 006

MAY, 2001

TO

anna

&

amma

वागर्थाविव संपृक्तौ, वागर्थप्रतिपत्तये ।
जगतः पितरौ वन्दे पार्वतीपरमेश्वरौ ॥

CERTIFICATE

This is to certify that this dissertation entitled **"SENSORINEURAL HEARING LOSS: LINEAR VS NONLINEAR CIRCUITRY FITTING"** is the bonafide work in part fulfilment for the degree of Master of Science (Speech & Hearing) of the student with Register No. **M 9923.**

Mysore.

May, 2001

Dr. M. JAYARAM
Director
All India Institute of Speech &
Hearing, Mysore - 570006

CERTIFICATE

This is to certify that this dissertation entitled **"SENSORINEURAL HEARING LOSS: LINEAR VS NONLINEAR CIRCUITRY FITTING"** has been prepared under my supervision and guidance. It is also certified that this dissertation has not been submitted earlier in any other university for the award of any Diploma or Degree.

Mysore.

May, 2001

Dr. K. RAJALAKSHMI
Lecturer in Audiology
Department of Audiology
All India Institute of Speech & Hearing,
Mysore - 570 006

DECLARATION

This dissertation entitled **"SENSORINEURAL HEARING LOSS: LINEAR VS NONLINEAR CIRCUITRY FITTING" is** the result of my own study under the guidance of Dr. K. Rajalakshmi, Lecturer in Department of Audiology, All India Institute of Speech & Hearing, Mysore and has not been submitted earlier in any other University for the award of any Diploma or Degree.

Mysore.

May, 2001 **Register No. M 9923**

ACKNOWLEDGEMENTS

I would like to extend my heartfelt gratitude to my guide Dr. K. Rajalakshmi, Lecturer, Department of Audiology, for her guidance, patience and immeasurable help. Also I would like to thank her for being a constant source of inspiration and for putting up with all my liables.

I express my sincere gratitude to Dr. M. Jayaram, Director, *AW* India Institute of Speech & Hearing, Mysore, for permitting me to take up this work.

I would like to thank all the faculty members of the Department of Audiology and Speech Pathology for their timely support.

I thank Dr. Venkatesan & Dr. Lancy D'souza for assisting me in doing the statistical analysis.

I thank all my classmates who directly or indirectly provided me a moral support. Thanks to all my seniors and juniors for extending their co-operative hands whenever I needed.

I express my gratitude to the library staff- Mahadev, Lokesh, Shivaprakash & Subramanya for their valuable help.

I sincerely thank M/s Maruthi Computers for metamorphosing the dissertation to its present delightful form.

I sincerely thank *tAr.* Sivappa and his brother for their excellent work in Xeroxing and binding.

Last, but not the least to Anna and Amma for their constant source of inspiration.

I extend my special gratitude to my brother, Dr. 5.N. Navin for showing love & affection on me.

Thanks to the Almighty for every bit of *success* and failure I have experienced in the completion of this dissertation.

TABLE OF CONTENTS

LIST OF TABLES

INTRODUCTION

I Introduction

Sound signals undergo a complex series of transformations in the early stages of auditory processing. Sound processing in the cochlea can be described in three stages: analysis, transduction and reduction. In the first stage, the uni-dimensional sound signal is transformed into a distributed representation along the length of the cochlea. This representation is then converted in the second stage into a pattern of electrical activity on thousands of parallel auditory nerve fibers. Finally, perceptual representations of timbre and pitch are extracted from these patterns in the third stage.

When sound impinges upon the eardrum of the outer ear, it causes vibrations that are transmitted via the ossicles of the middle ear, which in turn produce pressure waves in the fluids of the cochlea of the inner ear. These pressure waves cause mechanical displacements in the membranes of the cochlea, specifically the basilar membrane. Because of the unique spatially distributed geometry and mechanical properties of the basilar membrane, the vibrations acquire distinctive properties that reflect the structure of the sound stimulus.

The mechanical vibrations along the basilar membrane are transduced into electrical activity along a dense, topographically ordered, array of auditory nerve fibers. At each point, membrane displacements cause a local fluid flow that bends small filaments (cilia) that are attached to transduction cells, the inner hair cells (Shamma & Morrish, 1986). The bending of the cilia controls the flow of ionic currents through nonlinear channels into the hair cells. The ionic flow, in turn, generates electrical receptor potentials across the hair cell membranes. The receptor signals are then conveyed by the auditory nerve fibers to the central auditory system.

The functioning of the normal cochlea appears to reflect the operation of an active mechanism that is dependent on the integrity of the outer hair cells within the cochlea. This mechanism may involve the application of forces to the basilar membrane by the outer hair cells, and it plays an important role in producing the high sensitivity of the basilar membrane to weak sounds and sharp tuning on the basilar membrane (Moore, 1996).The normal basilar membrane shows several non-linearities (Rhode and Robles, 1974), including compressive input-output functions (Robles et al, 1986; Sellick et al., 1982), two-tone suppression (Ruggero, 1992) and combination generation also appear to depend on the operation of the active mechanism.

In a normal ear the basilar membrane vibration is distinctly nonlinear; the magnitude of the response does not grow directly in proportion with the magnitude of the input (Rhode, 1971; Rhode and Robles, 1974; Ruggero., 1992; Sellick et al., 1982).

The basilar membrane provides higher gain at low input sound levels and lower gain at higher input levels. The basilar membrane exhibits a nonlinear input-output function. This is illustrated in figure. 1, which shows input -output functions of the basilar membrane for a place with a characteristic frequency of 8 kHz (Robles et al, 1986).

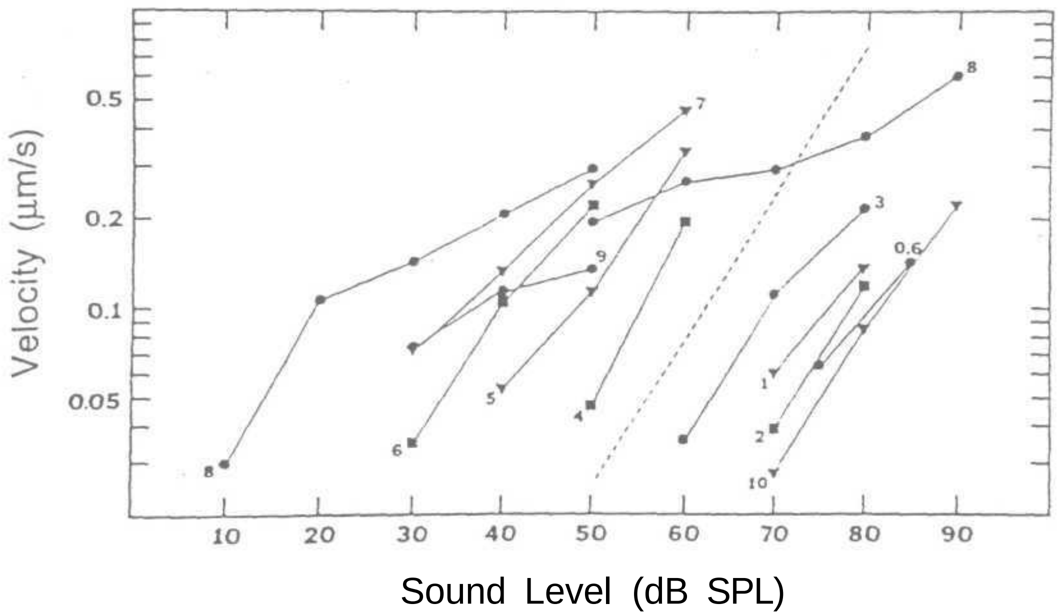

Figure l.Input-output functions for a place on the BM with CF = 8 kHz. The stimulating frequency, in kHz, is indicated by a number close to each curve. The dashed line indicates the slope that would be obtained if the responses were linear (velocfty directly proportional to sound pressure. [adapted from Robles et al. (1986).

The function for the characteristic frequency tone approaches linearity at low input levels (less than 20 dB SPL) and at high levels (above 90 dB) but has a shallow slope at midrange levels. This indicates a compressive non-linearity; a large range of input sound levels is compressed into a smaller range of response in the basilar membrane. At low and medium sound levels, the active mechanism amplifies the response on basilar membrane. The amplification may be as much as 55dB. As the sound level increases, the amplification progressively reduces. Thus, the response grows more slowly than it would in a linear system. When the sound level is sufficiently high, around 90 dB SPL, the active mechanism is unable to contribute any amplification and the response becomes linear.

The non-linearity mainly occurs when the stimulating frequency is close to the characteristic frequency of the point on the basilar membrane that is being monitored. For stimuli with frequencies well away from the characteristic frequency, the responses are more linear. Effectively, the compression occurs only around the peak of the response pattern on the basilar membrane. As a result, the peak in the pattern flattens out at high sound levels.

Depending on the intensity of the input sound signal, the outer hair cells perform some form of differential mechanical or electrical amplification. The low intensity acoustic inputs are amplified as much as 10,000 Xs, but the high intensity sounds are amplified to a much lower extent, viz., 10Xs. (Ruggero et al, 1992). Thus, the outer hair cell mechanics provides amplification of low amplitude acoustic input and compression of high amplitude inputs (Fig 1a).

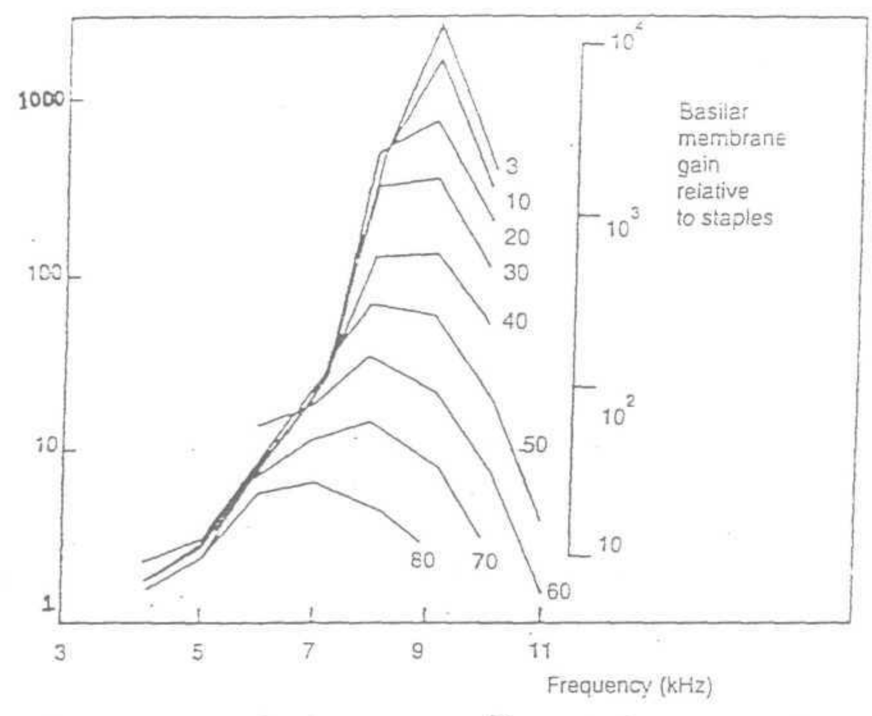

Figure 1a . Mechanical responses of a chinchilla cochlea to tones. Note that at peak, this gain is more than 10.000 for the lowest stimulus level (3 dB SPL) but drops to values between 10 and 100 for sound leveis typical of speech (60 dB SPL to 80 dB SPL). [Ruggero(1992)!]

Sensorineural hearing loss (SNHL) is a term used to describe losses in hearing sensitivity due to either an end organ malfunction (sensory) or an auditory nerve malfunction (neural). Although some sensorineural hearing loss can involve central auditory mechanisms, the term is generally used to describe peripheral auditory pathology that does not involve the outer or middle ear structures.

The most common causes of gradual onset sensorineural hearing loss are presbycusis (age-induced hearing loss) and occupational noise exposure. These pathologies typically arise from abnormalities in the cochlea and only after the condition has existed for some time do they involve the auditory nerve or higher auditory physiology. Some of the nerve fibers supplying the damaged, hair cells may also become damaged, resulting in a neural component to the hearing loss as well".

Cochlear hearing loss often involves damages to the outer hair cells and inner hair cells; the stereo cilia may be distorted or destroyed, or entire hair cells may die. The outer hair cells are generally more vulnerable to damage than the inner hair cells.

When outer hair cells are damaged, the active mechanism tends to be reduced in effectiveness or lost altogether. As a result, several changes occur; the sensitivity to weak sounds is reduced, so sounds need to be more intense to produce a given magnitude of response on the basilar membrane; the tuning curves on the basilar membrane become much broader; and all of the frequency selective nonlinear effects disappear. The most obvious symptom of cochlear hearing loss is a reduced ability to detect weak sounds.

However, cochlear hearing loss is also accompanied by a variety of other changes in the way that sound is perceived. Even if sounds are amplified so that they are well above the threshold for detection, the perception of those sounds is usually abnormal.

Several perceptual consequences occur due to the sensorineural hearing loss. Factors such as the frequency selectivity, loudness judgment and temporal integration are impaired in most of the sensorineural hearing loss patients.

1. FREQUENCY SELECTIVITY:

Frequency selectivity refers to the ability of the auditory system to separate or resolve (to a limited extent) the components in a complex sound. It is often qualified by using masking experiments to measure psychophysical tuning curves or to estimate auditory filter shapes using rippled noise or notched noise (Glasberg & Moore, 1990; Glasberg et al, 1984; Houtgast, 1977; Moore & Glasberg, 1983; Moore & Glasberg, 1987; Patterson, 1976; Patterson & Moore, 1986; Patterson & Nimmo-Smith, 1980). It seems likely that frequency selectivity as measured behaviorally would be poorer than normal in people with cochlear hearing loss.

There have been studies comparing psychophysical tuning curve in normal subjects and subjects with cochlear hearing loss (Bonding, 1979; Carney & Nelson, 1983; Festen & Plomp, 1983; Florentine, et al, 1980; Leshowitz et al, 1975; Nelson, 1991; Stelmachowicz et al, 1985; Tyler et al, 1982; Zwicker & Schorn, 1978). Their results are in general agreement that psychophysical tuning curve is broader than normal in the hearing-impaired subjects (Figure 2). When cochlear damage occurs, the input-output function of the basilar membrane becomes less compressive, having a slope closer to unity', which can be incorporated in the model by making the non-linearity less compressive.

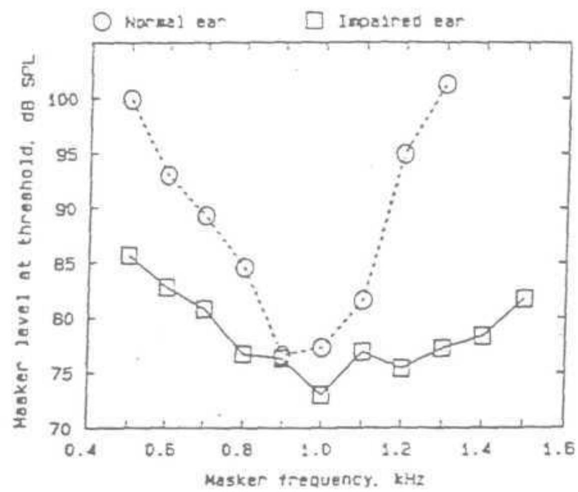

Figure 2. Psychophysical tuning curves (PTCs) determined in simultaneous masking for the normal ear (circles and dashed line) and the impaired ear (squares and continuous line) of a subject with a unilateral cochlear hearing loss. The signal frequency was 1 kHz. The variable masker was a narrowband noise. A fixed notched noise was gated with the variable masker, to restrict off-frequency listening. The signal was presented at a level 10 dB above its masked threshold in the notched noise alone. [from Florentine, et al., 1980].

2. LOUDNESS RECRUITMENT:

Steinberg & Gardner (1937) thought of recruitment as an ameliorating factor in hearing impairment, i.e., with decrease in impairment proportional to increasing intensity there would be a lesser degree of handicap due to the hearing loss. But the expanding action of recruitment when applied to the speech spectrum can present a distorted auditory image of the speech. The energy present in the vowel components is significantly greater than the acoustic energy for many of the consonant sounds. Hence, the intense vowel components can be detected with much greater ease than the weak low intensity consonant sounds when the dynamic range is reduced.

Villchur (1974) carried out an experiment to simulate the effect of recruitment on loudness relations in speech. He reported that recruitment was a sufficient cause for loss of intelligibility in the hearing-impaired, whether or not there are other causes.

Most, if not all people suffering from cochlear damage show loudness recruitment i.e., an abnormally rapid growth in loudness perception (Steinberg & Gardner, 1937). The absolute threshold is higher than normal. However, when a sound is increased in level above the absolute threshold, the rate of growth of loudness level with increasing sound level is greater than normal. When the level is sufficiently high, usually around 90 to 100 dB SPL, the loudness reaches its "normal" value; the sound appears as loud to the person with impaired hearing as it would to a person with normal hearing. With further increases in sound level above 90 to 100 dB SPL, the loudness grows in an almost normal manner (Figure 3).

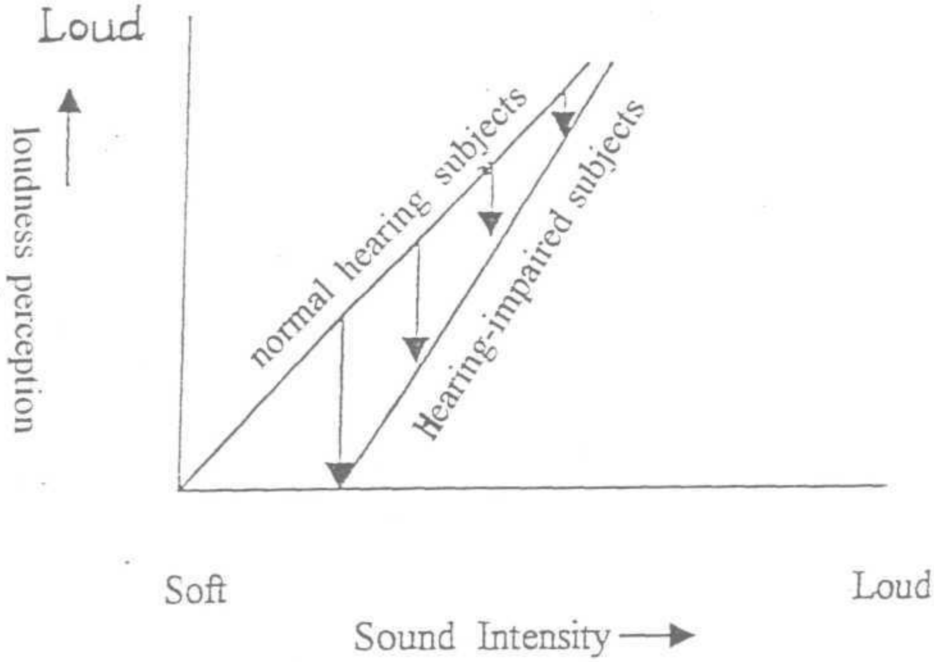

Fig. 3 . Loudness Responses of individual with normal hearing and person with SNHL (with recruitment), [adapted from Smriga,1993)

The patient with an end-organ disorder has recruitment, which can be characterized as an increased sensitivity to increasing increments in the intensity of a stimulus.

By the same definition, the patient with a neural disorder (or a conductive or central disorder) has no recruitment and may even show evidence of recruitment or loudness reversal.

A plausible explanation for loudness recruitment is that it arises from a reduction in or a loss of the compressive non-linearity in the input-output function of the basilar membrane.

If the input-output function on the basilar membrane were steeper (less compressive) than normal in an ear with cochlear damage, it would be expected to lead to an increased rate of growth of loudness with increasing sound level.

However, at high sound levels, around 90 to 100 dB SPL, the input-output function becomes almost linear in both normal and impaired ears. The magnitude of the basilar membrane response at high sound levels is roughly the same in a normal and an impaired ear (Figure 4).

This can explain why the loudness in an impaired ear usually "catches up" with that in a normal ear at sound levels around 90 to 100 dB SPL. Hence recruitment is a common phenomenon observed in most of the sensorineural hearing loss patients. These patients experience increased growth of loudness at higher levels of input intensities.

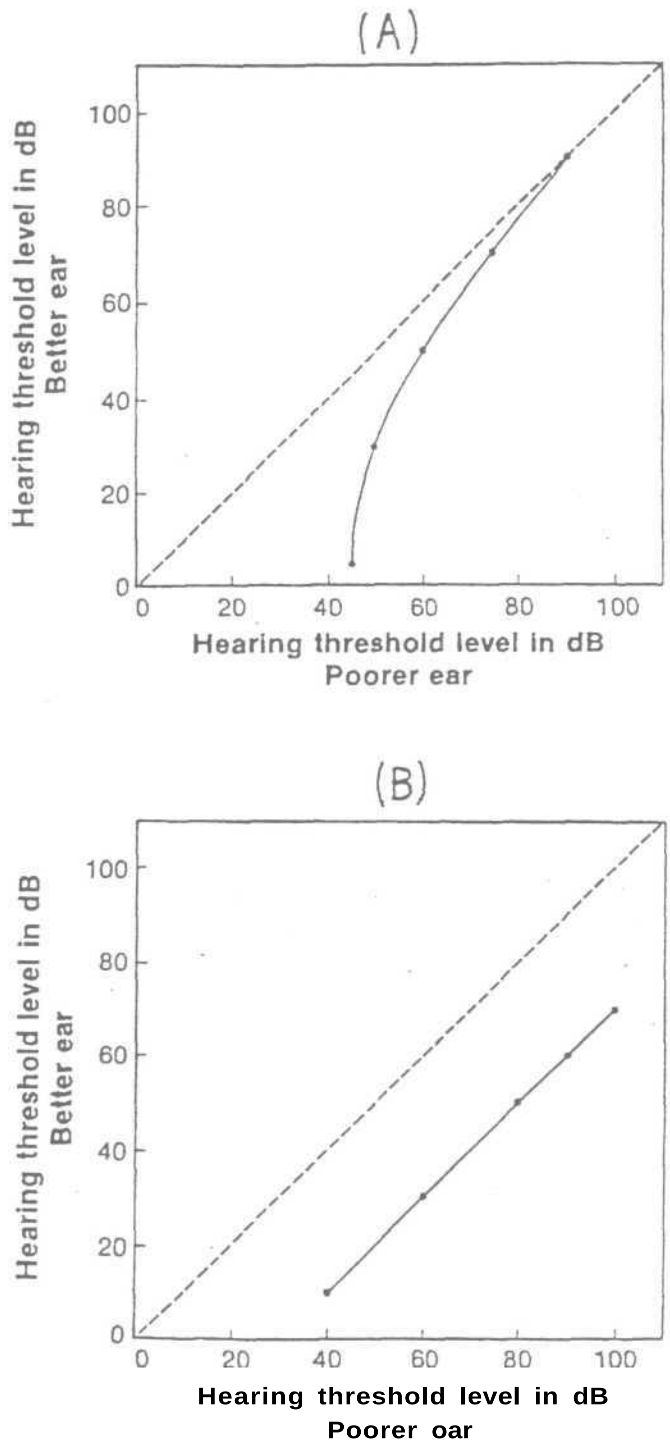

Figure 4 **A,** Complete recruitment. B. No recruitment.

Evans (1975) suggests that reduced frequency selectivity might be the main factor contributing to loudness recruitment. They suggest that, once the level of a sound exceeds threshold, the excitation in an ear with cochlear damage spreads more rapidly than normal across the array of neurons, and this leads to the abnormally rapid growth of loudness with increasing level.

A complementary way of describing this effect is in terms of dynamic range. This refers to the range of sound levels over which sounds are both audible and comfortable. The absolute threshold for detecting sounds determines the lower end of the dynamic range. The upper end is determined by the level at which sounds start to become uncomfortably loud.

Typically, in people with cochlear hearing loss, the absolute threshold is elevated, but the level at which sounds become uncomfortably loud is about the same as normal. Hence, the dynamic range is reduced compared with normal.

On average, the rate at which loudness grows with increasing intensity goes up with increasing absolute threshold at the test frequency (Glasberg & Moore, 1989; Helhnan & Mesielman, 1990, 1993).

This is consistent with the idea that threshold elevation and loudness recruitment are both linked to the loss of the active mechanism in the cochlea. When the absolute threshold is high, the dynamic range can be very small indeed.

Various theories have been put forth to explain this hair cell phenomenon of recruitment. Most of the initial theories presumed the involvement of nerve fibers as the cause of recruitment. (Steven et al., 1936; Salvi et al., 1983; Evans et al., 1976, Kiang et al., 1970). These theories explained loudness recruitment using concepts like steepening of the slope function, which relates the neural discharge rate to intensity, and broadening of tuning curve for high intensity sounds. Though these models seemed logically sound, they did not conform to psychoacoustic and physiologic data.

Later research put forth the view that both, the hair cells and the nerve fibers were involved in the sound processing in a recruiting ear (Lorente de No, 1937; Lurie, 1940; Simmons and Dixon, 1966).

The more recent theories postulate the involvement of hair cell damage in an attempt to explain the phenomenon of recruitment. (Tonndorf., 1980, 1981; and Killion, 1996). These newer theories explain the physiology of the damaged outer hair cell and are more successful in explaining the expanding action of recruitment. Their explanations concur with the psychoacoustics and physiologic data.

3. TEMPORAL ASPECTS:

For most subjects with cochlear damage, recruitment or equivalently, a reduction in the peripheral compressive non-linearity may provide a sufficient explanation for increased gap thresholds.

However, a few subjects show impairment in temporal resolution even using non-fluctuating stimuli (Jesteadt et al, 1976; Moore & Glasberg, 1988b, Moore et al, 1989; Place & Moore, 1991). It is possible that the subjects showing this impaired resolution had damage to both the outer hair cells (affecting the active process and the compressive nonlinearity) and inner hair cells (affecting the transduction process), or that they had a retro cochlear component to their hearing loss.

For people with cochlear damage, the change in absolute threshold with signal duration is often smaller than it is for normally hearing people. If the thresholds are plotted on dB versus log-duration coordinates, the slopes are usually much less in absolute value than the typical value of -3 dB/doubling found for normally hearing people. This is often described as reduced temporal integration (Carlyon et al, 1990; Chung, 1981; Elliott, 1975; Gengel & Watson, 1971; Hall & Fernandes, 1983; Pedersen &Eberling, 1973).

There is a trend for higher absolute thresholds to be associated with flatter slopes. In other words, the greater the hearing loss, the more reduced is the temporal integration. It seems likely that the main cause of reduced temporal integration in people with cochlear damage is a reduction or complete loss of the compressive non-linearity on the basilar membrane. Experiments in which the amplitude fluctuations in bands of noise are either expanded or compressed supports the idea that increased fluctuations result in impaired gap detection (Hall & Fernandes, 1983).

REHABILITATION FOR SENSORINEURAL HEARING LOSS PATIENTS:

In terms of the rehabilitation for the sensorineural hearing loss candidates, the conventional guidelines for amplification developed 20 years ago, no longer suits these patients. Historically, there have been a number of factors contributing to the lack of confidence shared by the public and hearing health care professionals regarding the benefit derived from hearing aids for the group of sensorineural hearing loss patients.

There is also a history of poor results obtained by patients with sensorineural hearing loss wearing conventional amplification. Current theories concerning cochlear mechanics and recent technological advances providing enhanced signal processing however raise serious doubts about the applicability of linear amplification for patients with sensorineural hearing loss.

Majority of patients with sensorineural hearing loss should be fitted with hearing aids providing nonlinear amplification. Issues relating to the exact parameters defining the nonlinear parameters (i.e., optimal number of compression bands, compression knee point, compression ratio, time constants) remain to be resolved.

Compression threshold (or knee point) can be defined as the lowest level at which compression becomes active. For hearing aids with input compression, the knee point is generally between 45 and 75 dB SPL. One advantage of lower compression thresholds is that the listener may not be aware of the system activating and deactivating (pumping and fluttering) since the hearing aid is in the compression mode most of the time. Knee points in the 70-75 dB ranges may result in frequent pumping and fluttering because of activation by normal conversational speech (particularly that generated by the user's own voice). Another advantage of the lower (45-55 dB) knee point is that compression is operating over a wider range (perhaps the fully dynamic range of the listener). Since recruitment begins near threshold, it could be argued that the use of nonlinear amplification should operate over the entire dynamic range.

Normal functioning outer hair cells act as a nonlinear "cochlear amplifier", providing up to 60 dB of gain for low input sounds (e.g., 0 dB). The listener with impaired outer hair cell function has no "cochlear amplifier" for high input or low input sounds. (Figure 5).

Hypothetically, as a result, the normal listener may have a dynamic range on the order of 100 dB while the sensorineural impaired listener's dynamic range may be compressed to 40 dB.

Thus, it can be argued that in order to restore "normal" nonlinearity to the sensorineural impaired ear, a compression ratio of no greater than 2.5:1 should be sufficient. High compression ratios (e.g., 8:1) with multichannel AGC having relatively low knee points may degrade the relative intensity cues required to identify certain speech sounds.

EFFECT OF OUTER HAIR CELL AMPLIFIERS

NORMAL & PARALYZED (RUGGERO & RICH,1991)

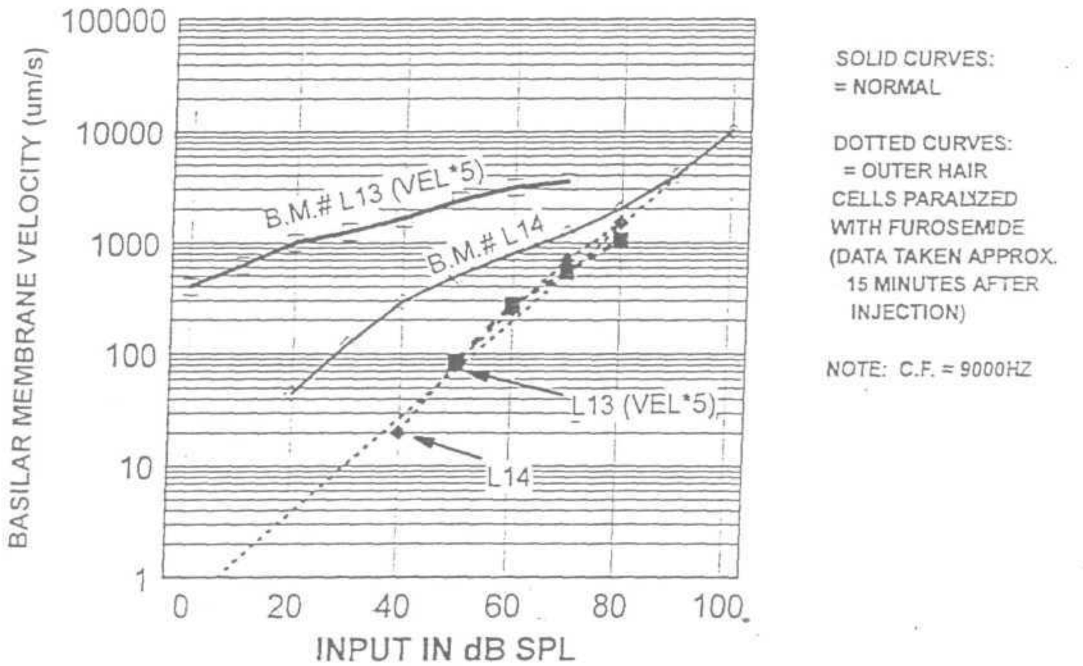

Figure 5 . Ruggero and Rich (1991) data on the operation of Corti's organ as a wide-dynamic-range-compression amplifier, [adapted from Ruggero, M-A.(1392)].

The use of amplification in these patients is to restore audibility via frequency selective amplification. Many hearing aids operate essentially as linear amplifiers; over most of their operating range they apply a gain that is independent of level.

It became apparent very soon after hearing aids first came into use that it was not practical to use linear amplification to compensate fully for the loss of audibility caused by cochlear damage. The major factor preventing this was loudness recruitment and the associated reduced dynamic range.

A person having a sensorineural hearing loss of 60 dB at all frequencies, the highest comfortable level (HCL) for such a person would typically be about 90 to 100 dB HL. A hearing aid that fully compensates for a loss of audibility would apply a gain of 60 dB at all frequencies. However, that would mean that any sound with a level above about 40 dB HL would be amplified to a level exceeding the HCL. In practice, many sounds encountered in everyday life would become unpleasantly loud.

Most hearing aids incorporate a way of limiting output of the aid so as to avoid discomfort to the user. In many hearing aids this is achieved by electronic peak clipping in the output stage of the aid. Such clipping introduces unpleasant sounding distortion (Grain and Van Tasell, 1994) and in practice most users of hearing aids set the volume control to avoid clipping in everyday listening situation.

So even when aids include output limiting, it has been found to be impractical in compensating fully for the loss of audibility.

A related problem with linear hearing aids is that user often finds it necessary to adjust the volume control to deal with different listening situations. The overall level of speech and other sounds can vary considerably from one situation to another and people with cochlear damage do not have sufficient dynamic range to deal with this.

Linear hearing aids amplify speech and noise equally well and do not take into account the phenomenon of loudness growth that is common to many with sensorineural hearing loss. The use of binaural fitting directional microphones moving physically nearer to the speaker and multiple fixed frequency responses in conventional hearing aids, have been shown to be effective nonadaptive processing approaches to noise reduction.

Linear amplifiers have as their defining feature the characteristic of adding the same amount of amplification to all levels of input intensity until the amplified output saturation limit has been exceeded.

Thus, low-level input signals will be amplified with the same amount of gain as high-level input signals. What is evident is that linear amplification systems can only provide adequate amplification for a very limited range of input levels.

For most of the loudness range within which the patient must operate, this amplification system would provide either too little (under-amplification) or too much (over-amplification) performance has the potential of creating a variety of listening problems for the patient wearing such a device.

If the normal cochlea was, as once thought, linear, passive and broadly tuned, then linear amplification would be appropriate for most hearing losses. However, recent findings indicate that this is certainly not an accurate description of the cochlea. The input - output function of the cochlea is nonlinear. At input sound levels less than 60 dB sound pressure level (SPL), the active mechanical process of the outer hair cells "amplifies" these sounds while sharpening frequency selectivity.

Normal functioning outer hair cells act as a nonlinear "cochlear amplifier", providing up to 60 dB of gain for low input sounds and no gain for high input sounds. The listener with impaired outer hair cell function has no "cochlear amplifier for high input or low input sounds. Hypothetically, as a result, the normal listener may have a dynamic range of the order of 100 dB where as a sensorineural hearing loss patient will have it around 40 dB.

Among the most illuminating observations in this area are the Mossbauer studies (Ruggero, 1992), which conclude that outer hair cells play an important part in whatever compression mechanisms operate in the normal ear. The gain functions of the K-amp and Re-sound hearing aids (figure. 5a) are qualitatively similar to the gain functions that Ruggero observed.

Thus it is no longer appropriate to fit a linear hearing aid to patients with hair-cell-based losses from 20 to 75 dB HL. High fidelity dynamic compression aids that make low-level signals uniformly and smoothly audible will be a great aid for these patients.

Idealized Gain Curves for K-Amp or Dynamic Compression Programmable Aids

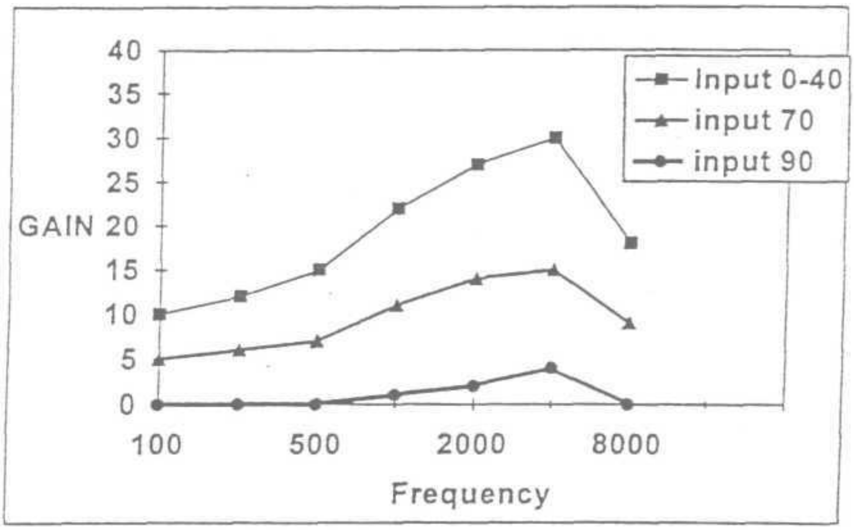

Figure 5a,- An idealized gain curve for K-Amp or dynamic compression programmable aids.

[adapteci from Ruggero M. A(1992)]

To more appropriately meet the needs of the recruiting sensorineural hearing loss patients, the audiologist must consider non-linear mode of compression amplification. Compression amplification has its defining feature the characteristic of decreasing (compressing) gain as the input level increases.

If fit properly, the non-linear performance structure of the compression amplifier can deliver a more natural loudness growth perception throughout the patients entire auditory listening range without under-amplification or over-amplification compromises. The type of compression system utilized can play an important role in delivering the desired performance result.

Compression of some sort is now used in all of the more advanced hearing aids in the market. The rationale for including compression varies widely, and, consequently, the manner in which it is implemented also varies widely. If the output of a hearing aid is not limited in some way, output sounds will sometimes exceed the loudness discomfort level of the aid wearer.

The primary advantage expected for compression limiting is that even if the aid wearer selects a high volume control setting to amplify weak input signals, the compressor will prevent discomfort from occurring, without distortion, if a high level wanted or unwanted signal occurs.

Compression amplification must be considered as an effort to provide the amplification performance suitable for the sensorineural hearing loss condition. First, by automatically reducing gain as the input level mcreases, the compression aid will be capable of keeping the output level within the wearer's more restricted dynamic range without the use of peak clipping, thus minimizing cochlear overload. Second, since gain is automatically reduced, the likelihood that amplification saturation will be reached is also reduced.

With good fitting strategies and with good compression aid designs, peak clipping can effectively be avoided altogether. Thus, compression systems can reproduce the input signal to noise ratio without compromise more readily than linear amplifiers.

For a successful compression processing the electronic compression circuit of the hearing aid should accurately match with that of the compression characteristics of the lost outer hair cells in a sensorineural hearing loss patient.

Compensation for recruitment by compression in a sensorineural hearing loss patient turns out to be an electronic substitute for the physiological compression of the outer hair cells.

To date, empirical investigations of the K-Amp have yet to be reported, although numerous reports have indicated that the circuit is being used successfully in the field (Knight, 1992; Kruger and Kruger, 1993). After formal research on the K-Amp has been completed, it should become possible to determine the true benefits and limitations of the algorithm.

Thus the present study was carried out -

1. To compare performance of K-Amp and Linear hearing aids using real ear insertion gain response in sensorineural hearing loss subjects.

2. To compare the performance of K-Amp versus the linear hearing aids on tasks involving questions, paired word repetition and tolerance level in the sensorineural hearing loss patients was evaluated.

3. To compare the subjective preference for either the K-Amp or the linear hearing aids by the sensorineural hearing loss patients was evaluated.

REVIEW
OF
LITERA TURE

II. REVIEW OF LITERATURE

Psychoacoustics and neurophysiologic research indicates that there are several conditions of the sensorineural hearing loss that are not adequately addressed when fitting hearing aids of conventional linear hearing aid technology. These conditions include recruitment, cochlear overload (the "busy line" effect) and brain enhancement of signal to noise perception. Non-linear circuit designs appear more suitable for meeting the needs of sensorineural hearing loss, yet even the most current non-linear designs can be improved if we are to truly meet the needs of the sensorineural hearing loss patient.

Hearing-impaired persons with sensorineural involvement typically experience a reduction in speech intelligibility in noise - a reduction in number of usable cochlear hair cell density can predispose the impaired cochlea to processing saturation. This may leave few, if any, remaining hair cells available to process signal information in the presence of noise. This busy line' effect further deteriorates the impaired person's ability to understand speech in noise (Smriga, 1993). Also, there appears to be an efferent neural mechanism in the brain that can assist the cochlea in perceiving signal in the presence of background noise. This natural signal processing advantage is triggered only when the brain is aware of a speech-in-noise' input condition.

If the impaired cochlear function is unable to transmit this key data to the brain due to the "busy line' effect, such natural enhancement may not be triggered in the sensorineural impaired ear (Smriga et al, 1993).

When fitting patients with sensorineural hearing loss, it is important that the amplifier selected is designed to keep the output level within the user's restricted dynamic range, thus minimizing the busy line effect. He also suggests that the amplifier must maintain the input signal-to-noise ratio available for the normal hearing ear to process, thus triggering the olivocochlear bundle suppression whenever possible.

In recent years, many efforts have been made to manufacture adaptive processing circuitry that efficiently addresses the signal-to-noise problem that interferes with the speech intelligibility. Crandell (1991) indicated that for normal listeners, even a 1 dB enhancement in signal-to-noise (S/N) ratio could result in a 6%-8% improvement in speech recognition when evaluated with sentences from the Speech in Noise (SPIN) test. In other words, a 5 dB improvement in S/N theoretically could provide as much a 35% increase in speech recognition for sensorineural hearing-impaired listeners having good speech recognition abilities. One means of providing a perceptual, albeit not a physical, S/N improvement is to use a hearing aid that automatically reduces low frequency gain in response to the level of the input signal. That is, at low input levels the hearing aid provides a broad, flattened frequency response. However, as the input level increases, the signal processing provides progressively less low frequency gain without altering the high frequency gain.

Tillman et al., (1970) demonstrated that the hearing aid user required better S-N ratio to understand speech with a hearing aid than without the aid. This finding conforms to the results of Villchur (1973) study. He compared the speech recognition scores of six hearing impaired subjects under three conditions; linear hearing aid with voice interference at 10 dB; linear hearing aid with interference removed and compression amplification with voice interference at - 10 dB. He found that the speech recognition scores for the linear amplification in quiet was higher than the scores in noise. Hence, it seems logical to try to develop an electronic circuit, designed to suppress the noise relative to the speech.

Research findings on hearing aids with automatic low frequency reduction have been mixed. The abundance of studies suggest that they do not usually result in significant improvement in speech recognition in comparison to a conventional linear hearing unless (1) the conventional aid provides an inappropriate frequency response or excessive maximum output for the subject's needs or (2) the background noise is limited to the low frequencies. It is likely, therefore, that patients with sensorineural hearing loss who indicate a preference for aids with automatic noise reduction circuitry are reacting in a positive manner to reduced listening efforts in noise and or improved sound quality in comparison to listening through linear hearing aids.

Automatic signal processing (ASP) circuits have been described as a means to sort speech from noise. A common problem occurs when compression results in the reduction of gain at all frequencies so that the intelligibility of soft speech sounds is reduced.

A common hearing loss configuration results in normal or near normal hearing in the low frequencies, but a significant decrease in hearing for the higher frequencies.

ASP circuits that reduce the gain of low frequency signals may not be maximally beneficial for this type of configuration.

It was suggested many years ago that problems associated with reduced dynamic range could be alleviated by the use of automatic gain control (Steinberg & Gardner, 1937).

With AGC it is possible to amplify weak sounds more than stronger ones, which results in the wide dynamic range of the input signal being compressed into smaller dynamic ranges at the output. Hence, AGC systems are also called "compressors".

Although this idea sounds simple, in practice there are many ways of implementing AGC and there is still no clear consensus as to the "best" method, if there is such a thing. There is also considerable controversy about the efficacy of AGC systems.

AGC systems have been designed in many different forms, mostly on the basis of different rationales or design goals. Some systems are intended to adjust the gain automatically for different listening situations.

The idea is to relieve the user of the need to adjust the volume control to deal with these situations. Usually such systems change their gain slowly with changes in sound level; this is achieved by masking the recovery time of the AGC circuit rather long (greater than a few hundred milliseconds).

These systems are often referred to as "automatic volume control" (AVC). Although it is generally accepted that AVC can be useful, relatively few commercial hearing aids incorporate AVC. One reason is that, after a brief intense sound such as a door slamming, the gain drops and stays low for some time; the aid effectively goes "dead".

Various studies have been carried out comparing linear and nonlinear circuitry fitting (Burchfield, 1970; Dreschler, 1988; Fikret - Pasa, 1993; and Hickson, et al, 1995). These studies investigated the effects of different amounts of limiter action vs. peak clipping. Some reported that, compressed speech discrimination scores were much better than those obtained under linear (1:1 ratio) amplification for subject with recruiting ears (Burchfield, 1970).

The results of the more recent studies (Dreschler et al., 1988; Fikret-Pasa 1993; and Hickson et al., 1995) differ from the earlier studies in that they report no significant differences in performance for linear vs. compression amplification. In fact, in the Fikret-Pasa (1993) study, some of the subjects showed superior performance with linear type circuit (compression ratio of 8:1 after 80 dB input) for inputs ranging from 65 to 85 dB SPL.

Linear gain hearing aids have traditionally been fitted using targets to optimize audibility and intelligibility of speech from a distance of 1 m; however, it is unrealistic to expect that the hearing aid user will be listening only to average speech inputs.

Pascoe (1975) measured speech level in different environments and with different talker effort and showed that the average level of speech can range form 52 to 85 dB SPL. With these changes in speech level, the shape of the speech spectrum also changes. Ideally, a hearing aid should be able to accommodate many different speech inputs: low-level speech originating from further away than 1 m, a raised voice (such as in a classroom setting), a shouted voice, or a listener's own speech productions. To truly quantify the benefit of hearing aid fitting strategies and specific circuit types, research must therefore include a wide range of speech conditions.

Extensive research has compared linear and WDRC circuitry (Benson et al, 1992; Dempsey, 1987; King and Martin, 1984; Lawrence et al, 1983 ; Moore et al, 1992; Neuman et al, 1994; SUIT et al, 1997; Troscianko and Gregory, 1984; Van Harten - de Bruijin et al,1997; Yund et al, 1987). Some controversy continues to surround this comparison because of many potential variations between studies, which include; compression type, stimulus type, range of input levels, subject characteristics (such as age and degree and type of hearing loss), and appropriateness of the frequency-gain characteristics of the hearing aid (Dillon, 1996).

Dillon (1996) describes no advantages for compression for speech in quiet at a comfortable listening level. However, compression has been shown to provide benefit for speech perception at reduced levels in quiet (Leaurence et al., 1983). This effect appears to be independent of the time constants of compression; i.e., it occurs for both slow and fast-acting compression. Dillon (1996) suggests further research in this area needs to concentrate on examining the compression advantage for each type of hearing loss and also suggests that compression processing needs to be examined for any advantages at levels that range between the most comfortable level (MCL) and hearing aid limiting, particularly for the possible increased ease of listening provided by compression.

Studies by Laurence et al., (1983) and Mare et al., (1992) both compared linear and compression circuitry using a 2 -channel fast - acting compression and that could also function as a linear aid by turning off the compression. In the Laurence et al., (1983) study, the aid were fitted using a method that ensured that speech at 70 dB SPL was comfortable and speech at 50 dB SPL was audible. Speech intelligibility was measured in quiet at three different levels and in noise. The compression aid maintained high speech perception scores for all conditions had lower scores for speech at low input levels.

Mare et al., (1992) showed a similar result, using aids fitted to individual loudness growth measures using the Loudness Growth in 1/2 -Octave Bands test. They also found that subjects with smaller dynamic ranges received greater benefit from the compression circuit. In addition, a questionnaire showed that most subjects preferred the compression aid to the linear in all situations except for the quality of their own voices. For both studies, speech intelligibility was tested at various input levels, but the spectrum did not change among listening conditions.

For compression systems such as syllabic compressors where the compression is normally activated, it is important that the compressor be input controlled. This is especially true if a high compresssion ratio is used, or else the user has little control over the output level of the signal.

Several laboratory-based experiments have produced some evidence in favor of syllabic compression. Lynn and Carhart (1963) obtained results showing that compression was better for some subjects (depending on the etiology of the hearing loss).

Studies using wearable aids again produced mixed evidence. Brink et al., (1975) found no advantage for commercial compression aid in quiet or in noise.

Lipmann et al., (1981) conducted experiment on five listeners with sensorineural hearing impairments using two 16-channel, computer-controlled, amplitude compression systems and four linear systems. One of the compression systems was designed to restore normal equal loudness contours; the other employed reduced high-frequency emphasis and reduced compression ratios. The linear systems differed only in their frequency-gain characteristics (orthotelephonic plus three characteristics with high-frequency emphasis that were expected to produce better results than orthotelephonic).

In the main experiment, all systems were compared for each listener using nonsense CVC monosyllables and sentence materials spoken by male and female talkers and presented in quiet/anechoic and noisy/reverberant environments at the most comfortable level for each listener. The linear systems with high-frequency emphasis performed substantially better than the orthotelephonic system.

Performance with compression was generally slightly worse than with linear amplification, compression was superior to linear amplification only when speech materials with significant item-to-item level variation were used in quiet with subjects with more severe losses and when reduced input speech levels were used. To the extent that these two conditions represent real-life communication conditions, these results suggest that compression is preferable to linear amplification in a wearable hearing aid.

Nabelek (1983) compared low level, wide dynamic range and high-level compression with each other and with linear amplification, using laboratory equipment, for 13 subjects with sloping moderate sensorineural hearing losses. In quiet, and in speech-shaped noise added after compression, at both 0 dB and 5 dB Signal to Noise Ratio, there were no significant differences between any of the processing schemes.

In a similar experiment, Mare et al, (1992) compared low amplitude compression (LAC) processing, processing similar to compression limiting (but with a 2:1 compression ratio), and linear amplification for 10 subjects with mild to moderate sensorineural hearing losses. Speech reception thresholds (SRTs) differed by less than 0.5 dB for sentence materials and by 2 percentage points for the intelligibility of CVC words.

Fabry and Olsen (1991) carried out a study to judge subject preference for WDRC versus linear compression limiter over a period of one month. They reported no subject preference either objectively or subjectively. Caraway & Carhart (1967) had reached a similar conclusion after they had attempted to improve speech understanding by using amplitude compression.

Dreschler (1988) studied the effects of specific compression threshold settings on phoneme identification and found a small but significant increase in identification scores for the lowest setting of compression threshold. More recently. Barker & Dillon(1999) have reported that a higher compression threshold was preferred by a majority of his subjects. From these studies it is apparent that compression limiting does not provide much benefit from point of view of improved speech perception. But, limiting the output power by means of compression amplification is both feasible and desirable, as it not only protects the ear, but at the same time reduces distortion to a minimum, thus maintaining in most cases, the maximum level of performance over a wide range of speech input levels (Hudgins, et al., 1948).

Hickson et al., 1995 reported that compression amplification (1.3 and 1.8 ratio) was not found superior to linear amplification in any of the test conditions (quiet condition and two different noise conditions) and was significantly worse than linear amplification in the babble background noise condition.

Study by Lawrence et al, (1983) comparing linear vs. two-channel compression aids, the compression aids proved to be substantially better than the linear aids. The compression aids allowed good speech discrimination over a wide range of sound levels.

Fabry and Stypulkowski (1993) studied two-band processors utilizing compression or linear processing in either band. For noisy backgrounds, the linear processing in high band (which contains low intensity speech sounds) was judged to be superior to the compression.

Thus, multiband compression aids provides a better fit for those patients with reduced dynamic range and loudness recruitment. In this context, the study by Souza and Turner (1999) provides much insight. They examined the effects of alternation of temporal information and audibility of speech cues through multichannel compression system. They reported that multichannel compression improved speech recognition under conditions where superior audibility was provided by the two -channel compression systems over linear amplification.

Dreschler (1988b) measured speech identification using CVC nonsense words for compression limiting versus peak clipping for hearing aids in various degrees of saturation. Scores for compression limiting were 15 percentage points higher than for peak clipping, although the levels at which the two types of processed stimuli were presented may also have affected the true differences in scores.

Dawson et al., (1990) investigated whether clients changed over from a peak clipping hearing aid to a compression limiting aid with otherwise similar electroacoustic properties reacted positively or negatively to the change. Twenty-eight clients with a moderate or severe loss reacted positively to the change, but the reactions of 32 clients with a profound loss were divided. In a detailed study of 14 subjects with a profound loss, it seemed that those subjects who favored and / or who performed better with peak clipping were those who had the lowest speech identification abilities and who also used their hearing aids at the maximum volume and power settings.

Hawkins and Naidoo (1993) asked 12 subjects with mild to moderate, sloping sensorineural losses to rate the relative clarity and sound quality of continuous discourse amplified by two circuits employing peak clipping and compression limiting but which were otherwise identical. The degree of saturation of the output was related significantly higher than the peak clipping circuit for both judgments.

Moore et al, (1991) compared speech intelligibility scores obtained with a dual time constant compression limiter against those obtained with a linear amplifier and a compression limiter employing adaptive release time. Several combinations of release time for the fast time constant detector were compared. Best results were achieved with the release time of the fast compressor equal to 80 m sec (the attack time was fixed at 7.6 sec for three experiments and 2 sec for one experiment). A transient noise, with a peak level 18 dB above the speech RMS level, preceded the sentence material for three experiments; for two of these experiments, the dual time constant system produced significantly better speech intelligibility than did the linear system.

Moore et al., (1991) suggest that this level difference in the speech caused the higher scores for the dual time constant system. Unfortunately, a single time-constant compression limiter was not included in the experimental conditions, so the performance of the dual time constant system relative to a single time constant compressor is unknown. In the fourth experiment, speech at a variety of levels was used (without variation of the volume control). For the lowest speech level (55 dB SPL), intelligibility for the dual time constant compressor was far higher than that for the linear amplifier. At the highest speech level (85 dB SPL), the linear system produced the highest scores (significance not stated). Scores for the dual time constant system were lower for the 85 dB SPL input level than for those at the lower speech levels. In summary, whenever the speech was below MCL for the linear system (either because intense transient maskers required a volume control reduction to maintain comfort or because a low input level was used), the dual time constant compressor allowed speech to be presented closer to MCL and produced significantly better speech intelligibility. Reasons for the superiority of the dual time constant compressor over the adaptive time constant compressor (with apparently similar rationale and compression effects), in two out of four experiments, are not clear. The six subjects who participated in each experiment had moderate sensorineural loss.

A total of 42 selected patients with hearing impairment of purely perceptive type and with definite recruitment by Metz's test compared a behind-the-ear hearing aid with amplitude compression amplification and a behind-the-ear hearing aid with linear amplification over a trial period of at least 2 months, the instruments being tested by alternating use. Not quite one third of those studied (13 patients) chose the compression amplifier hearing aid, while the remainder preferred the conventional amplifier. The subjective evaluation revealed only minor differences between the two types of apparatus (Killion, 1996).

Until the 1990s, hearing aids fitted to most patients with high-frequency hearing losses have contained Linear Class A circuits. Two circuits, Linear Class D and nonlinear K-Amp (Killion, 1990; Killion, 1993, Killion et al, 1990; Longwell & Gawinski, 1992; Preves, 1992) instruments have become available in recent years.

Both may have advantages over Linear Class A circuits and each represents a replacement option for the Linear Class A instruments (Longwell & Gawinski, 1992)

The acoustic gain of a hearing instrument is the ratio, expressed in decibels, between the instrument's output and input. An instrument with 40 dB of acoustic again (at 1000 Hz) will increase a 60-dB SPL input signal by 40 dB and present a 100-dB SPL signal to the wearer's ear canal.

Amplification systems achieve their gain through two or more "stages of gain" (amplifier segments). However, the simple design uses a two-stage amplifier, while the more complex design uses a three-stage amplifier.

The simple design includes a maximum power output (MPO) control, while the complex design includes both an MPO control and a gain trim adjustment.

Separate gain stages also allow the designer to adjust the instrument's frequency response while optimizing the circuit's noise performance. Four types (or classes) of power amplifiers are now available. Class A, Class B, Class C and Class D.

Amplifiers were originally defined according to the classes of operation more than five decades ago. The first Amplifier developed, the class A, consumes a constant average value of current that is modulated (changed about this average in conformance with the desired signal). The battery power consumed is present. When this power is not being fed to the load, it has to be disputed by the active device, which results in a very inefficient use of battery power. The maximum efficiency possible with a Class A is 25%. This type of amplifier is generally inexpensive and produces low distortion at medium gain levels.

The second amplifier developed, the Class B, has two, rather than one, active devices. When an input signal is present, each device conducts current on alternate conduction cycles. In the absence of an input signal, neither device draws much current, which means that very little energy is drawn from the battery and dissipated as heat. This results in a more efficient use of battery power. In actual applications, a small DC bias current is designed into the amplifier to reduce "crossover distortion, "Which can occur as the signal swings from one half of the sine wave to the other. Maximum efficiency possible with a Class B is 78%. Class B amplifiers are capable of high gain and require less current than Class A Amplifiers in actual applications.

The Class C amplifier, which is used in radio transmitters, is not suitable for audio amplification because it is useful only at high frequencies. This amplifier is very efficient because in a non-conducting state there are no current losses.

The class D amplifier is similar to the Class C in that its operation is based on modulation of a high -frequency carrier, and the active devices are either fully on or fully off. However, in the Class D, the times when the devices are on and the times when they are off are varied so that the averaged effect can be made to replicate a low-frequency (audio) signal. Because the transistors are not dissipative, very little power is highly efficient. The theoretical maximum efficiency is 100%. The most promising benefit of Class D hearing instruments is higher sound quality is a very subjective parameter and difficult to quantify, three recent studies have led us to conclude that the perceptible differences in sound between hearing instruments with Class B amplifiers are a significant factor in the increased use of Class Integrated Receivers in hearing instruments.

In one study by Gawinski & Longwell (1992) listeners (110 self selected hearing instrument dispensers) clearly ranked Class D sound best in both In- the -ear (ITE) (70%) and behind -the- ear (BTE) (55%) implications, followed in order by Class B (27% in ITEs, 42% in BTEs) and Class A (3% in both ITEs and BTEs) along with ranking the listening modules, subjects were invited to comment on sound quality. Then-qualitative comments included descriptions of Class D sound as "Clear", "more natural," "of richer quality," and as having less distortion.

Manufacturers have suggested that the improved sound quality of Class D hearing instruments, compared to that of Class A - driven ; instruments, is a result on improved headroom and consequent reduction of clipping and other types of nonlinear distortion at high input levels.

The difference in perceived sound quality between Class A and Class D amplifiers is understandable, especially at relatively high output SPL. However, the difference between Class D and Class B perceived sound quality is not well understood and deserves further investigation.

A secondary benefit of Class D hearing instruments is more efficient use of current, which translates into longer battery life. The Class

A amplifier is the least efficient to use with moderate losses in normal listening situations and especially in quiet listening conditions. Although Class B amplifiers are more efficient than Class A, they are less efficient than Class D in handling intermediate to full signals.

The wide variety of sizes and power levels of Class D integrated receivers enables hearing instrument dispensers to fit a broader population of patients with Class D sound.

Most of the increased use of Class D amplification is accounted for by its availability in integrated receivers, which are popular because of their size and performance.

The Class D amplifier has a design fundamentally different form that of either the Class A or Class B. Its use in hearing instruments can result in higher perceived sound quality and longer battery life.

The advantage of the Linear Class D over the Linear Class A circuit is that the Class D does not saturate until higher input levels. Consequently, class D instruments produce less audible distortion, especially at relatively high output levels. This improvement has been described as increased "headroom" (Longwell & Gawinski, 1992). According to the results of a series of studies reported by Longwell and Gawinski (1992), the Class D circuit has been rated as having better sound quality characteristics, such as "clearer", "more natural", and "richer", in comparison with the Class A amplifier by both normal-hearing and hearing-impaired listeners.

The K-Amp, which also incorporates a Class D amplifier, is a nonlinear, automatic signal processing circuit. It provides relatively more gain in the higher frequencies with low input levels and conversely, less gain at high input levels.

The first K-AMP hearing aids (Version 1) were designed with two trimpots and a volume control. The on-off switch on the volume control did not turn the hearing and on and off, that was accomplished by opening the battery drawer. Instead, this switch was utilized to change the instrument from "manual" to "automatic".

In the manual mode, the volume control operated in a more or less conventional manner to control the gain of the input amplifier operating as a linear amplifier. Switched into automatic mode, the internal circuit perated electronic volume and tone controls, providing a maximum gain of approximately 25 dB for inputs below 40 dB SPL, gradually reducing gain with increasing input level down to a minimum gain of 0 dB (without gain or loss) for all inputs above approximately 90 dB SPL.

Three problems arose with the Version 1 design. First, while some wearers thought the new instruments sounded fine right away, others complained of the "pumping" or "breathing" sound caused by the rapid action of the automatic control circuit. Adding the Adaptive Compression circuit alleviated this problem. This circuit almost completely eliminates the "breathing" sound. It also improves the instrument's ability to ignore extremely short, intense transient sounds. Second, most people judge tonal balance while listening to loud sounds, for which the tone trimpot had no effect, leading to complaints that "the tone control doesn't work".

In Version 2, eliminating the tone trimpot and letting the manufacturer choose the appropriate frequency response behavior have corrected that. Third, some experienced hearing aid wearers seem to have become "gain hungry", possibly because of their need for extra gain for soft sounds. When the output gain trimpot was increased until such experienced the average gain they were used to hearing for conversational speech, they sometimes ended up with 10 to 20 dB of gain they could not turn down in Version 2, this was corrected by the volume control as the output gain control, leaving it up to the individual wearer to choose the amount of gain, if any, desired for loud sounds. Finally, the on-off switch in Version-2 does turn the hearing aid on and off.

A TILL-type algorithm, exemplified by the K-Amp, is based on assumptions that differ from those associated with BILL-type instruments (Killion, 1993). The K-Amp rationale assumes that most hearing-impaired listeners have greater hearing loss in the high frequencies than in the low frequencies and, therefore, require high-frequency emphasis amplification. The rationale also assumes that maximum high-frequency gain is required for low-level sounds, and that less gain is necessary when input levels increase. Although technically a single-band compressor, the algorithm provides significant high-frequency gain at low input levels and progressively less high-frequency gain as input levels increase.

Research findings are contradictory on whether it is necessary, or even desirable, however, to extend the high frequency response of wearable amplification for listeners. Pascoe et al., (1975) demonstrated, under controlled laboratory conditions, the advantage of extending bandwidth beyond 3000 Hz for improved speech reception in hearing impaired subjects.

Murray and Byrne (1986) varied the upper cutoff frequency between 1500, 2500, 3500, and 4500 Hz and had normal hearing as well as hearing-impaired subjects judge intelligibility and pleasantness. Normal listeners judged the wider bandwidths (e.g., upper cut-off frequencies at 4500 Hz) as being better on both intelligibility and pleasantness. However, the hearing-impaired subjects seldom reported additional benefit when bandwidth extended beyond 2500 Hz. Unfortunately, it was difficult to ascertain from this experiment whether the high frequency gain was sufficient above 2500 Hz to allow sounds to be audible for the impaired listeners.

Sullivan et al., (1980) indicated, on the other hand, that hearing impaired subjects experienced improved speech recognition with the addition of spectral information above 2000 Hz. They did not divide the spectral information above 2000 Hz into smaller bands, however, as did Murray and Byrne (1986). For the majority of sensorineural subjects, speech recognition increases as the high frequency region was amplified; however for subjects with sloping high frequency losses the conditions that produced the greatest high frequency gain results in decreased aided performance.

Skinner (1980) found that increasing high frequency gain by more than 20 dB above low frequency gain caused a decrement in performance for some; though not all, listeners. She speculated that spectral balance is needed to maintain optimal performance.

Sullivan et al., (1980) found at least three factors affecting performance of listeners with steeply sloping losses: (1) the overall gain of the aid, (2) the presence of background noise, and (3) the type of performance measured. In quiet, amplification with the broadest response was reported to provide the best performance. This finding is similar to that reported by Punch and Beck (1986), 27 who demonstrated that both normal and hearing-impaired subjects associate better sound quality with speech containing low frequency energy.

Sullivan et al., (1980) also indicated that they could obtain the same benefits by increasing the gain within a restricted bandwidth. They found that providing more high frequency energy resulted in improved scores for fricatives and affricatives. However, eliminating the high frequencies, while increasing the sensation level in the low and middle frequencies resulted in better recognition of plosives. They also reported that high frequency energy was more critical for syllable recognition in noise than in quiet. They concluded that, as audible bandwidth increases above 2000 Hz, speech recognition performance improves, but subjective judgments of speech intelligibility do not. This is in contrast with normal listeners, who report improvement in sound quality with increased bandwidth.

In an attempt to ascertain the minimum high frequency characteristics necessary to provide access to the important spectral cues for the sounds in English, Boothroyd and Medwetsk (1992) analyzed the /s/ sound because of its high frequency content and importance to speech recognition. They found that the lowest prominent spectral peak for the /s/ sound was 4300 Hz for males and 7200 Hz for females. Furthermore, they found that the frequency of the lowest prominent spectral peak varied by as much as 1000 Hz, depending on co-articulation effects. For example, the /s/ sound in the /I/ context was 1000 Hz higher than it was in the /u/ context. Thus, if a hearing aid has an upper frequency limit that is lower than the lowest prominent spectral peak, (i.e., 4300 Hz for male speakers and 7200 Hz for female speakers), there may be occasions when the listener will fail to hear the /s/ sound or may confuse it with the /f/ or /th/ sounds. Some individuals could make these distinctions simply on the basis of the cues provided by formant transitions, but Zeng and Turner (1990) found that hearing impaired subjects did not use these formant transition cues as efficiently as normal listeners and needed to hear the specific fricative spectra in order to provide correct identification.

Given these findings, combined with the difficulty generating audible high frequencies without creating feedback or distortion, it may be prudent to concentrate efforts on obtaining a better middle frequency response.

The treble increase at low levels is intended to provide or improve the audibility of weak high-frequency consonants. As the input level is raised, decreasing high-frequency gain serves to prevent loudness discomfort, minimize hearing aid saturation, and possibly compensate for abnormal loudness growth. The K-Amp circuit also applies Adaptive Compression, and has recently been designed with an adjustable compression ratio.

The K-AMP hearing aid is designed for hearing instrument wearers with mild - moderate and/or sharply sloping high frequency losses who require greater gain for quiet sounds than they do for loud sounds. It is also intended for those desiring good sound quality. The K-AMP amplifier provides high frequency gain for lower input levels, since the hearing loss for soft sounds is typically greater at high frequencies.

The K-AMP circuitry is good for patients with tolerance problems for patients whose listening environments vary constantly or for patients who are often in listening environments where loud noise is present. Overall gain is automatically reduced for high-level inputs, preventing audible distortion under all listening conditions (Killion; 1990).

The frequency range of the K-AMP instrument is from 100 to 14,000Hz depending on the specifications of the manufacturer. This ability to choose wide bandwidth, in combination with reduced distortion, is intended to enhance speech intelligibility in many listening situations (Killion 1990). Providing an accurate frequency response, however, is only part of the challenge in producing a high fidelity hearing aid. Reducing audible distortion also is a design factor.

Many hearing aid circuits, however, are designed to operate without distortion only up to 90 dB SPL even at minimum volume control settings. To handle loud sounds without distortion (distortion typically makes them sound even louder - and thus more annoying - than they would otherwise be), the input circuit of the K-AMP was designed to operate without distortion up to 110-115 dB SPL inputs.

For a successful compression processing the electronic compression circuit of the hearing aid should accurately match with that of the compression characteristics of the lost outer hair cells in a sensorineural hearing loss patient. Compensation for recruitment by compression in a sensorineural hearing loss patient turns out to be an electronic substitute for the physiological compression of the outer hair cells.

Adaptive amplification can be obtained using hearing aids containing K-Amp circuitry. For these aids, the high frequencies are emphasized for low input intensities, but, as input increases, the frequency response flattens, approximating the REUR.

Studies have been conducted using K-Amp hearing aids regarding its efficacy in sensorineural hearing loss subjects.

Marshall chasin (2000) compared performance of digital aids with K-Amp hearing aids on 65 well-trained musicians. Eighty-two percent preferred K-Amp circuitry, have these; twelve percent noticed no difference, but chose the K-Amp aids based on cost. All other aids (analog or digital) clip or limit at high levels. The resulting output appears distorted to the trained ear (and to many untrained ears). But K-Amp hearing aids provide reduced gain for high-level inputs, preventing audible distortion under all listening conditions (Killion; 1990).

Hayes & Cormier (2000) conducted a double blind comparison of three types of hearing aid circuits: Class A linear peak cupping, Class D compression limiting and K-Amp wide dynamic range compression. Subjective ratings, speech perception tests, real ear measurements and questionnaire data were obtained from a group of 17 new hearing aid users with mild to moderate sensorineural hearing loss. The results indicated a similar performance of all three circuits. They saw no evidence of performance degradation due to saturation distortion, even in the presence of high levels of speech and noise. Their primary conclusions included recommending K-Amps to new hearing aid users with mild to moderate hearing loss, mostly on the basis of battery life, while cautioning about the use of compression knee-point controls and recognizing that Class A and Class D amplifiers are virtually equivalent in every performance measurement.

To examine the benefits of a new non-linear amplification circuit, Nilsson et al (1997) conducted a double blind crossover study. Two new hearing aids were developed; they were identical in external appearance and differed only in that one involved ordinary linear amplification while the other employed compressive amplification (the K-Amp circuit). Forty-five experienced users with sensorineural hearing loss, aged 60-80 years, used each of the aids for ten weeks, in balanced order. The subjects' need for hearing aid ranged from listening to radio and television to extensive use in all kinds of demanding listening situations. The results, using a structured questionnaire concerning real-life settings, speech reception tests and subject preferences for a particular hearing aid, showed little difference between the two hearing aids. Twenty-three subjects selected the non-linear amplification circuit, 20 subjects preferred the linear hearing aid and two chose to return to their previous aid. No consistent differences between those preferring the linear circuit and those preferring compression were found.

Hawkins & Naidoo (1997) compared monaural and binaural hearing aid preferences of 15 adults with mild-to-moderately-severe bilaterally symmetrical sensorineural hearing losses. Subjects listened to connected discourse in quiet and background noise at 70 and 80 dB SPL with K-Amp, linear Class D, linear output limiting compression (OLC), Manhattan II and linear asymmetrical peak clipping circuit (APC). In experiment 1, subjects made judgments of sound quality and speech intelligibility in a modified paired-comparison paradigm during which they compared the monaural and binaural fittings of each circuit. In experiment 2, subjects engaged in subjective ratings on a scale of 0 to 10. Subjects benefited from improved sound quality and speech intelligibility in high-noise conditions when fit with binaural K-Amp, linear Class D, and linear OLC and Manhattan II circuits. Monaural listening was preferred with the APC circuit. Results indicated that improved sound quality and speech intelligibility might be obtained with binaural fittings of circuits that include high fidelity, low distortion or increased headroom.

Painton (1993) measured total amount of "ampclusion" (low-frequency amplification + occlusion) and its reduction in two identically fitting canals hearing aids with three circuitry capabilities: 1) K-Amp, 2) linear and 3) noise reduction (ANP II). The insertion gains of all three circuits were adjusted to closely match and the amounts of occlusion for both aids were almost identical. Real-ear measurements were obtained as the author vocalized the phoneme /i/ at 65 dB SPL at 2 ft in three conditions: 1) unoccluded, 2) occluded, aid off, and 3) occluded aid on. The results indicated a reduction in the low-frequency amplification component of ampclusion for both adaptive circuits when compared to the linear circuit, with the K-Amp showing slightly greater reduction than the ANP II. These findings suggest that in addition to noise-reduction circuitry, K-Amp circuitry will provide a more pleasant experience to the hearing aid wearer when listening to his/her own voice.

To date, the assumptions underlying TILL-type circuits such as the K-Amp have not been thoroughly evaluated. Skinner (1980) measured word recognition scores for hearing-impaired listeners with sloping high-frequency hearing losses and reported that performance improved as the amount of high-frequency emphasis amplification increased. In a subsequent investigation, Skinner et al, (1982) reported that word recognition performance of 2 hearing-impaired listeners increased with the overall bandwidth of the test material. The bandwidth associated with typical hearing aids of the time was associated with generally poor performance.

Killion and Tillman (1982) reported that experimental hearing aids built with high-fidelity microphones and receivers similar to those now used in the K-Amp were associated with subjective fidelity ratings that were comparable to those associated with high-fidelity loudspeakers. These results, however, were based on the impressions of normal-hearing listeners. These investigations support the rationale for the K-Amp circuit, although all were conducted before the K-Amp was introduced.

Surr et al., (1997) studied eighteen subjects, experienced with Class A hearing aid use. They were given a choice of binaural hearing aids with either Linear Class D circuits or Class D with K-Amp circuits after consecutive 30-day trial periods with each set of instruments. The patients also rated the benefit obtained from each circuit using the Profile of Hearing Aid Benefit (PHAB). There was no significant difference in the number of subjects who chose one or the other of the circuits. Further, the PHAB scores showed no statistically significant differences between the two circuits. In most cases, the instruments rated highest on each of the subscales by an individual subject were also the ones preferred based on the 30 day trial.

Hearing aids with either Class D Linear or Class D with K-Amp circuits provided significant benefit in many everyday listening environments for individuals with a mild to moderate degree of hearing loss. Subjective choice between the Linear Class D and the K-Amp circuits was relatively evenly divided (Surr et al., 1997).

Real-ear measurements typically are used in clinical hearing aid evaluations (HAE) to select the appropriate frequency/gain response for a given audiometric configuration. Formulae based on hearing threshold values have been developed that prescribe a target response (Hecox, 1988). When comparing linear versus nonlinear instruments, the frequency/gain patterns can be matched fairly closely for a specific signal level, but large differences in output occur across different input levels. As a result, the two types of instruments provide different gain in varying daily listening environments independent of volitional volume adjustments.

METHODOLOGY

II METHODOLOGY:

1. Subjects:

Sixteen patients with bilateral mild to moderate sensorineural hearing loss their age ranging from 26 to 47 years served as subjects for the present study. The mean age of the patients was 36 years. All patients had good speech discrimination scores.

The audiometric criteria for these patients were hearing levels of 20 dB HL or poorer at 1000 Hz and 55 dB HL or better at 2000 Hz by air and bone conduction and normal immittance test results indicative of sensorineural type of hearing impairment. Consequently, the audiometric profile for each patient was within the low risk fitting ranges recommended by Killion for the K-Amp hearing aids (Killion, 1993).

2. Instrumentation:

K-Amp and Linear hearing aids were used for the present study. The FONTX 6500 C hearing aid test system was used to carryout the real ear insertion response measurement with the above hearing aids. The instrument meets the ANSI requirements (ANSI S3.22 -1987) for test box measures. All measurements were followed by probe and instrument calibration.

3. Procedure:

a) Objective measures:

The real ear insertion response (REIR) measures for K-Amp & linear hearing aids were carried out separately on the subjects aided with K-Amp and linear hearing aids, to verify the hearing aid gain performance. REAR is the SPL, as a function of frequency at a specified measurement point in the ear canal for a specified sound field with the hearing aid in place and turned on. This is expressed either in SPL or as gain in decibels relative to the stimulus level. The real ear unaided response (REUR) is the SPL as a function of frequency, at a specified point in the unoccluded ear canal for a specified sound field. This can be expressed either in SPL or a gain in decibels relative to the stimulus level.

The real ear insertion response (REIR) is the difference in decibels as a function of frequency between the REUR and the REAR measurements taken at the same measurement point in the same sound field.

The FONIX 6500C hearing aid test system was utilized to record the REIR measurement with the hearing aids. The following steps were involved in the REAR measurement.

- Patients underwent a routine otoscopic examination to assure that there was no middle ear pathology in them.
- Probe calibration was done by holding the opening of the probe tube near the reference microphone. There was a flat response across

frequencies at the input level chosen for the calibration check. This was because the acoustical transmission effects of the probe tube will already be stored in the equipment memory and thus the measured response should be equal to the response of the reference microphone.

- Following the probe calibration, the patient was positioned at 45° azimuth with the loudspeaker placed 0.5 meters from the patient.

- The REUR was measured by inserting the probe tube to a depth of 25-28 mm from the tragal notch. During the REAR measurement, when sliding the hearing aid or earmould into the ear, the probe tube was held with one hand while pushing the hearing aid or the mold with the other. This will help keep the probe tube in place as *it* was during the REUR measurements.

- Once the hearing aid was placed in the ear, and switched on, the intensity level of input to hearing aid was kept at 70dB and the volume control was set to comfortable listening level of the patient.

- POGO formula was used to establish the target gain curve.

- Finally, the printout of the REIR graph was taken for both the groups of hearing aids in the subjects.

B. Subjective measures:

The performance of the two hearing aids was compared on the following tasks:

1. Questions
2. Paired - Words repetition
3. Tolerance level
4. Subjective preferences for the two groups of hearing aids.

Five standardized questions and 5-paired words were asked to each patient. One point was given for each correct answer for both, questions and correct paired word repetition.

The tolerance level was rated into 5 categories:

1. No tolerance problem, even at very loud sounds.
2. Tolerance problem for loud sounds.
3. Tolerance problem for moderate level sounds.
4. Tolerance problem for moderately soft sounds.
5. Tolerance problem for soft sounds.

Statistical Test: - (Paired t-test)

The results were analyzed using the t-test to calculate the statistical significant differences among the two groups of hearing aids.

RESUL TS AND DISCUSSIONS

IV. RESULTS & DISCUSSION

a. Objective measures.

REERs were conducted on both K-Amp and the linear hearing aids. From the figure.6 illustrated below, it can be inferred that the K-Amp hearing aids closely approximated the target gain curve than the linear hearing aids. The K-Amp hearing aids provide more gain at the higher frequencies where most of the speech frequency components are present, thus providing increased speech perception through these aids.

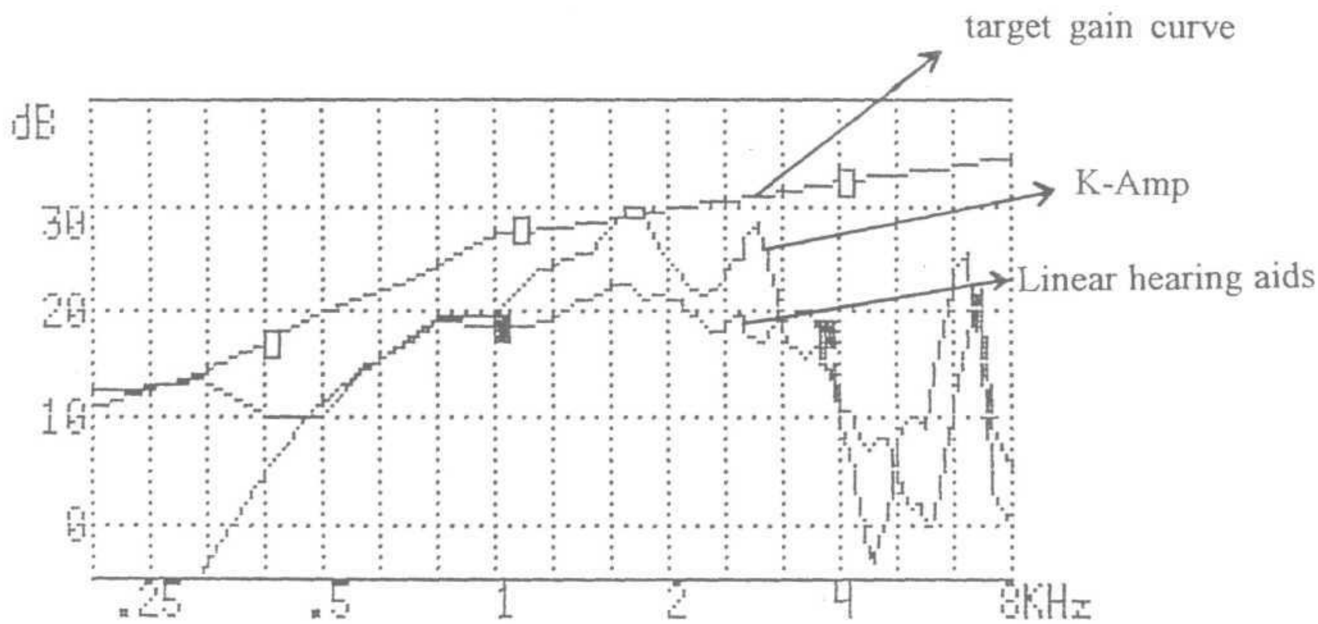

Figure 6. REIRs conducted on both K-Amp and the linear hearing aids.

SUIT et al., (1997) inferred that the mean REIRs for the Linear and the K-Amp hearing instruments with four input levels are different from each other.

The difference in the gain between the two circuits at lower input levels of 65 dB indicated no large differences among them in contrast. The REIRs obtained at other higher input levels indicated notable differences between the two hearing aids.

The linear hearing aids remained linear from 55 to 75 dB SPL input levels but showed evidence of saturation, and consequently distortion with the *85* dB SPL input level.

The results for the K-Amp hearing aids showed gradually changing REIRs function across the input levels tested, that is, mean gain decreased systematically with increasing input levels over the 55-to-85 dB SPL range.

These results reflect the compression effects of the K-Amp hearing aids, with increasing input levels, which is essential for a sensori neural hearing loss patient with recruitment.

b. Subjective measures.

1. Question task:

The subjects were required to answer the questions asked by the clinician. Each correct answer was given a score of 1. Only one request for repetition of the question was allowed.

Table 1 indicating mean, standard deviation and t-value for Linear and K- Amp instruments on question tasks.

SI. No.	Hearing aids	Questions answered(Mean)	Standard Deviation	t-value
1.	Linear	4.281	0.672	8.34
2.	K-Amp	4.593	0.551	

From table 1, we can infer that the K-Amp hearing aids are significantly better than the linear hearing aids in terms of performance on question tasks ($t = 8.34$; $p<0.01$). This may be due to the fact that the K-Amp hearing aids provide significantly more gain at higher frequencies, which enhances the speech intelligibility for these patients. Contrasting studies conclude that hearing aids with either class D Linear or Class D with K-Amp circuits provide significant benefit in many everyday listening environments for individuals with a mild to moderate degree of hearing loss. (Surr et al, 1997).

The use of amplification in these patients is to restore audibility via frequency selective amplification. Many hearing aids operate essentially as linear amplifiers; over most of their operating range they apply a gain that is independent of level. It is not practical to use linear amplification to compensate fully for the loss of audibility caused by cochlear damage. The major factor preventing this is the loudness recruitment and the associated reduced dynamic range.

A person having a sensorineural hearing loss of 60 dB at all frequencies, the highest comfortable level (HCL) for such a person would typically be about 90 to 100 dB HL. A hearing aid that fully compensates for loss of audibility would apply a gain of 60 dB at all frequencies. However, that would mean that any sound with a level above about 40 dB HL would be amplified to a level exceeding the HCL. In practice, many sounds encountered in everyday life would become unpleasantly loud.

Most hearing aids incorporate a way of limiting output of the aid so as to avoid discomfort to the user. In many hearing aids this is achieved by electronic peak clipping in the output stage of the aid. Such clipping introduces unpleasant sounding distortion (Crain and Van Tasell, 1994) and in practice most users of hearing aids set the volume control to avoid clipping in everyday listening situation. So even when aids include output limiting, it has been found to be impractical to compensate fully for loss of audibility.

Many studies have been carried out to compare the subject performance on linear peak clipping aids and compression aids, with equivocal results. Dreschler, (1988), Barker et al., (1999), Jenstad et al., (1999), report that compression hearing aids give much better speech intelligibility scores than peak clipping linear amplifiers. But other researchers (Caraway and Carhart, 1967; Blegvad, 1974; Dreschler et al., 1984 and Biering- Sorensen et al., 1995) found no significant improvement in speech intelligibility for compression aids as compared to peak clipping aids.

It must also be noted that most of the studies (Blegvad, 1974; and others) used compressor aids with extreme compression settings. This may have distorted the loudness relationships among speech sounds to such an extent that though the loudness of the output sound never reached discomfort levels the output signal was unintelligible to a large extent.

Hence in most studies, the subjects preferred the linear amplification with intact loudness relationships and uncomfortable loudness levels to the distorted speech output of the compression aids.

A related problem with linear hearing aids is that the user often finds it necessary to adjust the volume control to deal with different listening situations. The overall level of speech and other sounds can vary considerably from one situation to another and people with cochlear damage do not have sufficient dynamic range to deal with this.

Linear hearing aids amplify speech and noise equally well and do not take into account the phenomenon of loudness growth that is common to many people with sensorineural hearing loss. The use of binaurally fitted directional microphones, moving physically nearer to the speaker and multiple fixed frequency responses in conventional hearing aids, have been shown to be one of the effective nonadaptive processing approaches to noise reduction.

Linear amplifiers have as their defining feature the characteristic of adding the same amount of amplification to all levels of input intensity until the amplified output saturation limit has been exceeded. Thus, low-level input signals will be amplified with the same amount of gain as high-level input signals. What is evident is that linear amplification systems can only provide adequate amplification for a very limited range of input levels.

For most of the loudness range within which the patient must operate, this amplification system would provide either too little (under-amplification) or too much (over-amplification) performance has the potential of creating a variety of listening problems for the patient wearing such a device.

If the normal cochlea was, as once thought, linear, passive and broadly tuned, then linear amplification would be appropriate for most hearing losses. However, recent findings indicate that this is certainly not an accurate description of the cochlea. The input - output function of the cochlea is nonlinear. At input sound levels less than 60 dB sound pressure level (SPL), the active mechanical process of the outer hair cell "amplifies" these sounds while sharpening frequency selectivity.

Normally functioning outer hair cells act as a nonlinear "cochlear amplifier", providing up to 60 dB of gain for low input sounds and no gain for high input sounds.

The listener with impaired outer hair cell function has no "cochlear amplifier for high input or low input sounds. Hypothetically, as a result, the normal listener may have a dynamic range of the order of 100 dB where as a sensorineural hearing loss patient will have it around 40 dB.

Among the most illuminating observations in this area are the Mossbauer studies (Ruggero, 1992) which concludes that outer hair cells play an important part in whatever compression mechanisms operate in the normal ear.

Thus it is no longer appropriate to fit a linear hearing aids to sensorineural hearing loss patients. A high fidelity dynamic compression aid such as the K-Amp hearing aid that make low-level signals uniformly and smoothly audible will be a great aid to the sharp and uneven loudness growth of segments of the speech code for people with recruitment.

2. Paired word tasks:

In this task the subjects were required to repeat the paired words that were presented by the clinician. Each correct repetition given a score of 1.

Table 2 indicating mean, standard deviation and t-value for Linear and K- Amp instruments on paired word repetition task.

Sl.No	Hearing Aids	Paired words Repitition (Mean)	Standard Deviation	t-value
1.	Linear	4.531	0.612	5.4744
2.	K-Amp	4.718	0.4754	

From the above table, it is evident that K-Amp hearing aids perform significantly better than the linear hearing aids on tasks involving paired word repetition ($t = 5.47$; $p < 0.01$). K-Amp hearing aids have been suggested as a solution to the universal problem of noise interfering with speech intelligibility. The effect of background noise on speech intelligibility is as follows: First, background noise is assumed to cause the greatest problem when its level is greatest. Second, the noise is assumed to have a predominantly low frequency emphasis, and specifically, to have a greater low frequency emphasis than the signal of interest, which is assumed to be speech. Third, speech intelligibility is greatest when the overall signal to noise ratio (SNR) or the received signal is maximized. Based upon these assumptions, it can be concluded that in the presence of high levels of background noise, the hearing aid should exhibit a greater low frequency cut than it does in the presence of low background noise levels. K-Amp aids have been effective in improving the speech perception in presence of background noise (Killion, 1990).

The better performance obtained using the K-Amp hearing aids may be attributed to the fact that these aids have decreasing gain function with increase in the stimulus input level thus preventing these aids from being driven into saturation.

The K-Amp hearing aid is designed for patients with mild to moderate and /or sharply sloping high frequency losses that require greater gain at the higher frequencies. It also is intended for those demanding good sound quality. The frequency range of the K-Amp instrument is from 100 to 14000 Hz depending upon the specifications of the manufacturer. This ability to choose wide bandwidth, in combination with reduced distortion, is intended to enhance speech intelligibility in many listening situations (Killion; 1990).

Hearing aids have long been thought of as devices that compensate for a hearing loss. The early schemes for selecting amplification attempted to provide amount of gain equal to the amount of loss at each frequency. After it became apparent that this was not appropriate because of the reduced dynamic range of those with a sensorineural hearing loss, subsequent schemes applied different amount of gain at different frequencies.

Frequency- dependent K-Amp hearing aids modify the speech input and thus change the output of the preliminary auditory analysis by producing new patterns. Therefore, it takes considerable time for the user to adapt to the new pattern and to learn new "recognition" cues.

The major benefit of K-Amp hearing aids over linear hearing aids is that the output is limited without generating distortion components. This advantage is likely to be perceived by the mildly and moderately impaired as a quality improvement in sound reproduction. Those severally and profoundly impaired hearing aid users who have sufficient residual frequency selectivity to make use of spectral shape information are also likely to benefit from the reduced distortion. Those who make little or no use of spectral shape information may notice no difference between linear and K-Amp hearing aids. If the hearing loss is less severe at low frequencies, overall speech intelligibility may even improve because of the distortion. One disadvantage of compression systems is that it is not technically possible to produce as high an SSPL as it is with peak clipping. This is more marked with a speech input signal than with a pure tone test signal. The disadvantage will be greatest for those requiring the highest possible output levels.

Several studies using well -controlled laboratory based equipment have shown that compression limiting results in better speech discrimination scores than limiting by peak clipping (Davis et al., 1947; Hudgins, et al., 1948). This conclusion is consistent with the reduction in intelligibility caused by the high level of harmonic and inter-modulation distortion that can be generated by peak clipping (Gioannini and Franzen 1978; Young et al, 1979). An additional benefit is the improvement in reproduction quality of compression limiting over peak clipping. Compression limiting such as those utilized in the K-Amp hearing aids are used routinely in the broadcast industry to achieve high average signal levels without serious quality degradation.

Contrasting studies using wearable hearing aids have produced less convincing evidence of the value of compression limiting (Blegrad, 1974; Brink, et al., 1975). Many of the studies comparing compression hearing aids to linear hearing aids with peak clipping have not specified all the basic compression parameters so that the type of compression employed and the appropriateness of the parameters are not clear. Further difficulties of interpretation arise because the results can be biased toward compression if an inappropriate reference condition is chosen, and away from compression because of the effect of technical deficiencies in the particular compression aids employed (Nabalek and Robinette, 1975). Also, the linear behind-the-ear hearing aid produced distortion at higher input intensity levels (Vinay & Rajalakshmi, 1999).

Also, it has long been noted that there are large differences between the intensities of different speech syllables, even for continuous speech at a fixed overall long-term level. One rationale for the use of K-Amp hearing aids is that without it, for a hearing- impaired overall level for amplified speech, such that this wide range of individual syllable intensities can all be made audible and comfortably loud. With the linear hearing aids, the most intense syllables are as loud as is comfortable, the weakest syllables may be either inaudible or at a sensation level insufficient to allow the user to identify them. It is likely that because of the effects of recruitment the loudness differences between syllables will appear greater to the hearing impaired than to normal hearing people. Even if the weak syllables are sufficiently audible when presented as isolated syllables, there is some possibility that they will be masked by adjacent intense syllables when they are part of continuous speech. Although this process of forward and backward masking of speech by speech has not been specifically investigated, the general phenomenon of masking by temporarily separated sounds does occur in both normal and hearing-impaired listeners. The severity of temporal masking is known to increase in some hearing - impaired people (Festen and Plomp, 1983; Moore etal., 1985).

Thus, when fitting the patients with sensorineural hearing loss it is important to provide compression amplification with emphasis of the high frequencies for the listener. Since many patients with sensorineural hearing loss lack sufficient motivation to treat it, they often demand to be convinced concerning the improvement that a hearing aid can provide. Since the main goal of amplification is to facilitate the case of communication, some patients may be disappointed when they experience only minimal benefit during the initial evaluation of amplification.

Proper counseling can alleviate this difficulty. Patients must be educated that prediction of long-term benefit from amplification is tenuous at best because of the initial adjustment and learning process that takes place. Most hearing aid users require several weeks before the analysis consisting of converting incoming acoustic signals into neural impulses is followed by a "recognition device" that matches these neural impulses to previously learned information to recognize phonemes properly.

3. Tolerance Level:

The subjects were asked to report any tolerance problem that they experienced while wearing either of the two hearing aids. Subjects were asked to rate the level of tolerance experienced either in one of the five categories rated for tolerance level.

Table 3 indicating mean, standard deviation and t- value for Linear and K- Amp instruments for Tolerance level.

SI. No.	Hearing Aids	Mean Tolerance level	Std - Deviation	t-value
1	Linear	1.59375	0.6547	10.96
2	K-Amp	1.21875	0.4134	

From table 3, we can infer that the K-Amp hearing aid users did not report of any tolerance problem for higher input sound levels. The K-Amp hearing aids perform significantly better than the linear hearing aids ($t = 10.96$; $p < 0.01$). These results are due to the fact that the K-Amp hearing aids provide decreasing gain as the input level increases (SUIT et al.,1997). These subjects when tried using the linear hearing aids complained of intolerance to sounds at higher levels. For a variety of reasons, it is essential that the maximum output that the hearing aid can deliver, be limited in some way such that output levels capable of damaging the residual hearing of the aid user must be avoided, as most levels capable of causing loudness discomfort to the user. Also, the hearing aid must be prevented from going into overload in an uncontrolled manner, or else the aid may amplify in an unpredictable way.

Most, if not all, people suffering from cochlear damage show loudness recruitment (Steinberg & Gardner, 1937). The absolute threshold is higher than normal. However, when a sound is increased in level above the absolute threshold, the rate of growth of loudness with increasing sound level is greater than normal. When the level is sufficiently high, usually around 90 to 100 dB SPL, the loudness reaches its "normal" value; the sound appears as loud to the person with impaired hearing as it would to a person with normal hearing. With further increases in sound level above 90 to 100 dB SPL, the loudness grows in an almost normal manner.

The patient with an end-organ disorder has recruitment (i.e., an abnormally rapid growth in loudness), that can be characterized as an increased sensitivity to increasing increments in the intensity of a stimulus. By the same definition, the patient with a neural disorder (or a conductive or central disorder) has no recruitment and may even show evidence of recruitment or loudness reversal

A plausible explanation for loudness recruitment is that, it arises from a reduction in or loss of the compressive non-linearity in the input-output function of the basilar membrane. If the input-output function on the basilar membrane is steeper (less compressive) than normal in an ear with cochlear damage, it would be expected to lead to an increased rate of growth of loudness with increasing sound levels. However, at high sound levels, around 90 to 100 dB SPL, the input-output function becomes almost linear in both normal and impaired ears. The magnitude of the basilar membrane response at high sound levels is roughly the same in a normal and an impaired ear. This can explain why the loudness in an impaired ear usually catches up with that in a normal ear at sound level around 90 to 100 dB SPL.

Evans (1975) suggests that reduced frequency selectivity might be the main factor contributing to loudness recruitment. They suggest that, once the level of a sound exceeds threshold, the excitation in an ear with cochlear damage spreads more rapidly than normal across the array of neurons, and this leads to the abnormally rapid growth of loudness with increasing levels.

A complementary way of describing this effect is in terms of dynamic range. This refers to the range of sound levels over which sounds are both audible and comfortable. The absolute threshold for detecting sounds determines the lower end of the dynamic range. The upper end is determined by the level at which sounds start to become uncomfortably loud.

Typically, in people with cochlear hearing loss, the absolute threshold is elevated, but the level at which sounds become uncomfortably loud is about the same as normal. Hence, the dynamic range is reduced compared with normal.

On average, the rate at which loudness grows with increasing intensity goes up with increasing absolute threshold at the test frequency (Glasberg & Moore, 1989; Hellman & Meiselman, 1990 and 1993). This is consistent with the idea that threshold elevation and loudness recruitment are both linked to the loss of the active mechanism in the cochlea. When the absolute threshold is high, the dynamic range can be very small indeed.

Recruitment is a perplexing problem both for the physician and for the patient. The disproportionate growth in loudness, when compared to normal ear function presents a serious problem to the proper selection and fitting of hearing aid devices (Schiff and Sandlin, 1982). The variation in the presenting picture of problems, with respect to degree of recruitment and its frequency distribution further complicates the issue.

As early as in 1937, Steinberg and Gardner understood the implications of recruitment for amplification for hearing impaired persons. They suggested that owing to the expanding action of the hearing loss following hair cell damage, it would be necessary to introduce a corresponding compression in the amplifier.

Villchur (1974) also emphasized that compensation for the loudness recruitment is a necessary although possibly insufficient condition for restoring speech intelligibility. This compensation should be in the form of taking the larger dynamic range of speech and fitting it into the smaller dynamic range of the subject. This will require a non-distorting, decreasing gain system with increasing input, i.e., an aid that subjects the speech signals to amplitude compression, more at high frequency than at low frequencies.

The linear amplification devices are inefficient in meeting these demands. The linear aid gives uniform amount of amplification across all the frequencies. If we give sufficient gain so as to raise the low frequency sounds (vowels) into the region of audibility, this amount of gain is insufficient to reach even the threshold of audibility at the higher frequencies, where the meaning bearing consonants are situated. If the gain is increased such that the amplified speech in the high frequency low intensity region is within the region of audibility/threshold, a corresponding gain in the low frequency, high intensity region will result in intensities beyond the discomfort levels.

*

This expansion of the loudness of vowel peaks and loss of perceived consonant vowel relationships in speech should be countered by better compensatory circuitry of the hearing aid (Villchur, 1974).

Peak clipping may be used as an output limiting strategy by adjusting the amplifier output limit to within the user's restricted dynamic range. Thus, the peak clipping may be used to control the output dynamic range of the linear amplifier.

As a result of peak clipping, there is a change in the wave morphology. After peak clipping, the input waveform is flattened or squared. This results in a physical distortion of the signal because its high amplitude elements are now restricted relative to its low amplitude components. This results in a reduction in the sound quality. This disturbs the S-N ratio and leads to distortion.

In a peak clipping aid, there is a one-to-one relationship between the dB change in input and the resultant dB change in output until the aid reaches saturation. Once the input plus gain exceeds the maximum output limit of the amplifier, peak clipping occurs. This results in harmonic distortion. In addition, some of the information contained in the input signal is not present in the output signal as a result of this peak clipping. Thus, signal fidelity is not maintained. In order to minimize the distortions in signal resulting from peak clipping, a non-linear amplification such as the K-Amp circuitry should be employed.

Fit properly, the K-Amp hearing aids can deliver a more natural loudness growth perception throughout the patients entire auditory listening range without under- amplification or over amplification compromises. K-Amp aids that make low-level signals uniformly and smoothly audible will be a great aid to the sharp and uneven loudness growth of segments of the speech code for people with recruitment.

4. Subjective Preference:

The Subjects were asked to give their subjective preference for either of the hearing aids after the performance on the above three tasks.

Table 4 indicating the patient preferences for the two type of hearing instruments.

SI. No.	Hearing Aids	Subjective Preference
1.	Linear	3
2.	K-Amp	13

Regarding the subjective preference for either of the hearing aids, the subjects were asked to rate the hearing aids in terms of loudness judgment and degree of speech understanding.

The above table depicts a clear preference for K-Amp hearing aids by thirteen patients. The K-Amp circuits are good for patients with tolerance problem or for patients who are often involved in listening environments where loud noise is present.

Contrasting study by SUIT et al., (1997) indicated that the subjective choice between the linear class D and the K-Amp circuits were relatively evenly divided

The K-Amp is a nonlinear circuit, which provides relatively more gain in the higher frequencies with low input levels and conversely, less gain at high input levels. These factors play a major role in the patient's preference for K-Amp hearing aids than to the linear hearing aids.

The variables such as the knowledge of the functioning of two hearing aids were controlled. The patients were not provided any information about the hearing aid functioning. The two instruments were tested randomly by changing their order. The patients were tested on tasks such as questions, paired words and tolerance level for both the types of hearing aids.

The overall preference for K-Amp circuits indicates the better performance with these circuits for sensorineural hearing loss patients. Linear hearing aids on the other hand, provides the same amount of gain to all levels of input intensity thus creating a variety of listening problems for the patients wearing such a device.

SUMMARY AND CONCL USIONS

V. SUMMARY & CONCLUSION:

The present study was carried out to delineate the perceptual consequences of sensorineural hearing loss and their implications on hearing aid designs. Most hearing aids incorporate a way of limiting the output of the aid so as to avoid discomfort to the user. In many hearing aids this is achieved by electronic peak clipping in the output stage of the aid. However such clipping introduces unpleasant sounding distortion.

A related problem with linear hearing aids is that user often finds it necessary to adjust the volume control to deal with different listening situations. The overall level of speech and other sounds can vary considerably from one situation to another and people with cochlear damage do not have sufficient dynamic range to deal with this. Linear hearing aids amplify speech and noise equally well and do not take into account the phenomenon of loudness growth that is common to many with sensorineural hearing loss. Linear amplifiers provide same amount of amplification at all levels of input intensities thus resulting in under amplification at low levels and over amplification at higher levels.

If the normal cochlea was, as once thought, linear, passive and broadly tuned, then linear amplification would be appropriate for most hearing losses. However, recent findings indicate that this is certainly not an accurate description of the cochlea. To more appropriately meet the needs of the recruiting sensorineural hearing loss patients, the fitter must consider non-linear form of compression amplification. The K-Amp is a nonlinear hearing aid, which provides decreasing gain at increasing levels of input intensity. The K-Amp hearing aid provides high frequency gain for lower input levels, since the hearing loss for soft sounds is typically greater at higher frequencies.

It is true, unfortunately, that over the past several decades during which electronic bearing aid devices have been available to the acoustically impaired there has not been developed one single consensus on that amplification system best meeting the needs of the hearing-impaired patient. Neither the otological, audiological, nor the hearing aid dispensing community has presented a compelling hearing assessment procedure which, when applied to hearing assessment procedure which, when applied to hearing losses, will provide unqualified parameters for hearing aid selection. In the medical community, for example, there is no consensus regarding candidacy for hearing aid amplification.

There remains that vestige of the professional discipline which feels that patients presenting with confirmed sensorineural impairment are not candidates for hearing aid amplification, even in view of irrefutable evidence suggesting that the vast majority of patients who have a sensorineural deficit use hearing aids successfully.

Thus compression amplification must be considered in an effort to provide the amplification performance suitable for the sensorineural hearing loss patients. Thus, compression systems can reproduce the input signal to noise ratio without compromise more readily than the linear amplifiers.

REFERENCES

REFERENCES

Barford J. (1979): Speech perception processes and fitting of hearing aids. *Audiology,* 18:430-441.

Barker, C, & Dillon H. (1999). Client preference for compression threshold in single -channel wide dynamic range compression hearing aids. ***Ear and*** *Hearing,* 20, 127-139.

Beiring - Sorenson, M., Riess, H., Boisen G, & Parving, A (1995). A clinical comparative investigation of a non-linear vs. linear hearing aid. ***Scandinavian*** *Audiology,* 24, 125-132.

Benson, D., Clark, T.M., & Johnson, J.S (1992). Patient experiences with multiband full dynamic range compression. ***Ear and*** *Hearing,* 13, 320-330.

Blegrad, B. (1974). Clinical evaluation of behind-the-ear hearing aids with compression amplification **.*Scandinavian Audiology,*** 3, 57-60.

Bonding, P. (1979). Frequency selectivity and speech discrimination in sensorineural hearing loss. *Scandinavian Audiology,* 8, 205-216.

Boothroyd A., Medwetsky L.(1992). Spectral distribution of /s/ and the frequency response of hearing aids. *Ear **and** Hearing :* 12: 150-157.

Brink, B., Borrild, K., & Oving, K. (1975). Body worn stereophonic compression aids used by profoundly hearing impaired children. ***Scandinavian*** *Audiology,* 4, 199-205.

Buerkli-Halevy O. (1986). Hearing aid design parameters affecting performance in noise. *Hearing* ***Journal:*** 39(2): 18-22.

Burchfield, S.B. (1970). Cited in R.W. Rintelmann, (1972). Effect of amplitude compression upon speech perception: A review of research. *Scandinavian Audiology,* 1(3), 127-134.

Caraway B.J., & **Carhart R (1967).** Influence of Compressor action on speech intelligibility. *Journal of Acoustical society of America,* 41, 1424-1433.

Carlyon, R.P., Buss, S & Florentine, M (1990). Temporal integration of trains of tone pulses by normal and by Cochlear impaired listeners. ***Journal*** *of the Acoustical Society of America,* 87, 260-268.

Carney, A.E., & Nelson, D.A. (1983). An analysis of psychophysical tuning curves in normal and pathological ears. *Journal of the Acoustical Society of America,* 73, 268-278.

Chung, D.Y. (1981). Masking, temporal integration and sensorineural hearing loss. ***Journal*** *of Speech and Hearing Research,* 24, 514-520.

Crain, T.R., & Van Tasell, D.J. (1994). Effect of peak clipping on speech recognition threshold, ***Ear and*** *Hearing,* 15, 443-453.

Crandell, C. (1991). Individual differences in speech recognition ability: Implications for hearing aid selection. *Ear and Hearing,* 12(suppl 6): 100S-106S.

Dawson, P., Dillion, H., & Battaglia, J. (1991). Output limiting compression for the severely-profoundly deaf. *Australian Journal of Audiology,* 13, 1-12.

Dempsey, J.J. (1987). Effect of automatic signal - processing amplification on speech recognition in noise for persons with sensorineural hearing loss. ***Annals***

of Otology, Rhinology, and Laryngology, 96, 251-253.

Dillon, H. (1996). Tutorial: Compression? Yes, but for low or high frequencies, for low or high intensities and with what response time? ***Ear and Hearing,*** 17, 287-307.

Dreschler W.A. (1998). The effect of specific compression setting on phoneme identification in hearing impaired subjects. ***Scandinavian Audiology,*** **17,** 35-43.

Dreschler, W.A. (1988a). Dynamic range reduction by peak clipping or compression and its effects on phoneme perception in hearing-impaired listeners. ***Scandinavian Audiology,*** **17,45-51.**

Elliott, L.L. (1975). Temporal and masking phenomena in persons with sensorineural hearing loss. *Audiology,* 14,336-353.

Evans E.F. (1975). The sharpening of frequency selectivity in the normal and abnormal cochlea. *Audiology,* 14, 419-442.

Evans, E.F. (1976). Cited in Silman, S and Silverman C.A., ***Auditory Diagnosis:Principles and Applications,*** **(p.l88).San** Deigo:AcademicPress. **Inc.**

Evans, E.F., Pratt, S. R., & Cooper, N.P. (1989). Correspondence between behavioural and physiological frequency selectivity in the guinea pig. ***British Journal*** *of Audiology*, 23, 151-152.

Fabry, D.A & Olsen, W.O (1991). Cited in Van Vliet D. (1994). Debating Linear vs. non- linear fittings. ***Hearing Instruments,*** 45 (3), 33-55.

Fabry, D.A. & Styupulkowsi, P. (1993). Cited in D. **Van** Vliet (1994) Debating linear vs. non-linear fittings. ***Hearing Instruments,*** 45(3), 33-35.

Festen, J. M., & **Plomp, R. (1983).** Relations between auditory functions in impaired hearing. *Journal of the Acoustical Society of America,* 88, 1725-1736.

Festen, J. M., & **Plomp, R. (1990).** Effects of fluctuating noise and interfering speech on the speech - reception threshold for impaired and normal hearing. *Journal of the Acoustical Society of America,* 88, 1725-1736.

Festen, J.M., & **Plomp, R. (1983).** Relations between auditory functions in impaired hearing. *Journal of the Acoustical Society of America,* 73, 652-662.

Fikret-pasa, S. **(1993).** Cited in Van Vliet D.(1994). Debating linear Vs. Non - linear fittings. *Hearing Instruments,* 45(3), 33-35.

Florentine, M., Buiis, S., Scharf, B., & Zwicker, E (1980). Frequency selectivity in normally hearing and hearing-impaired observers. *Journal of Speech and Hearing Research,* 23, 643-741.

Furui, S. **(1986).** On the role of spectral transition for speech perception. *Journal of the Acoustical Society of America,* 80, 1016-1025.

Gawinski, M.J., & **Longwell, T. F. (1992).** Fitting strategies for the 90s. Class D amplification. *Hearing Journal,* 45 (9), 26-31.

Gengel, R. W., & **Watson,** C.S **(1971).** Temporal integration; I. Clinical implications of a laboratory study. H Additional data from hearing-impaired subjects. *Journal of Speech and Hearing Disorders,* 36, 213-224.

Gioannini. L., & Franzen. R. (1978). Comparison of the effects of hearing aid harmonic distortion on performance scores for the MRHT and a PB-50 test. *Journal of Auditory Research,* 18, 203- 208.

Glasberg, B.R., & Moore, B.C.J. (1989). Psychoacoustic abilities of subjects with unilateral and bilateral cochlear impairments and their relationship to the ability to understand speech. *Scandinavian Audiology,* (Suppl.32), 1-25.

Glasberg, B.R., & Moore, B.C.J., & Nimmo-Smith, I. (1984). Comparison of auditory filter shapes derived with three different maskers. *Journal of the Acoustical Society of America, 15,* 536-544.

Glassberg B.R., & Moore, B.C.J. (1990). Deviation of auditory filter shapes from notched-noise data *Hearing research,* 47', 103-138.

Hall, J.W., & Fernandes, M.A. (1983). Temporal integration, frequency resolution, and off-frequency listening in normal-hearing and cochlear-impaired listeners. *Journal of the Acoustical Society of America, 1A,* 1172-1177.

HaIligan)K,B., & Mercer, A. (1987). A new noise suppression technique. *Hearing Instruments,* 38(8): 29-32.

Hawkins, D., & Naidoo, S. (1997). Comparison of sound quality and clarity with asymmetrical peak clipping and output limiting compression. *Journal of the American Academy of Audiology,* 4, 221-228.

Hayes, D.E., & Cormier, K.L. (2000). Double blind comparison of three hearing aid circuits with new hearing aid users. *Scandinavian Audiology,* 29 (1), 10-19.

Hecox, K.E (1988). Evaluation of hearing aid performance. *Seminars in Hearing,* (9), 239-251.

Heide,J. (1987). Testing electro acoustic performance of ASP and nonlinear hearing aids. *Hearing Journal:* 40(4): 33-35, 1987.

Hellman, R.P., & Mesielman, C. H. (1993). Rate of loudness growth for pure tones in normal and impaired hearing. *Journal of the Acoustical Society* *of America,* 93, 966-975.

Hellman, R.P., & Mesielman, C. H. (1990). Loudness relations for individuals and groups in normal and impaired hearing. ***Journal** of the Acoustical Society of America,* 88, 2596-2606.

Hickson, L. M. (1994). Short course: Compression amplification in hearing aids. *American Journal of Audiology,* 3, 51-65.

Hickson, L., Dodd, B., & Byrne, D. (1995). Consonant perception with linear and compression amplification. *Scandinavian Audiology,* 24, 175-182.

Houtgast, T. (1977). Auditory-filter characteristics derived from direct-masking data and pulsation -threshold data with a rippled-noise masker. *Journal of the Acoustical Society of America,* 62, 409-415

Hudgins, C, Marquis. R., Nichols, R., Peterson, G., & Ross, D. (1984). The comparative performance of an experimental hearing aid and two commercial instruments. *Journal of the Acoustical Society of America,* 20, 241-258.

Jenstad, L.M., Seewald, R.C., Cornelisse, L.E., & Shartz. J. (1999). Comparisons of linear gain and wide dynamic range compression hearing aid circuits: Aided speech perception measure. *Ear & Hearing,* 20, 117-126.

Jesteadt, W., Bilger, R.C., Green, D.M., & Patterson, J.H. (1976). Temporal acuity in listeners with sensorineural hearing loss. ***Journal of*** *Speech and Hearing Research,* 19, 357-370.

Kiang, N.Y.S., Moxon,E.L., & Levine, R.A. (1970). Cited in Silman, S., & Silverman, C.A. *Auditory Diagnosis. Principles* ***and*** *Applications,* p. 187, San Deigo, Academic Press, Inc.

Killion, M.C (1993). The K-Amp hearing aid: An attempt to present high fidelity for persons with impaired hearing. ***American Journal Of** Audiology,* 2 (2): 52-74

Killion, M.C.(1996). Talking hair cells: What they have to say about Hearing aids. In Berlin, C.I. (Ed). *"**Hair** cells **and hearing aids".*** San Diego. Singular Publishing Group, Inc.

Killion, M.C. (1990). Technological report: A high fidelity hearing aid. ***Hearing Instruments,*** 41 (8), 38-39.

Killion, M.C., & Tillman T.W. (1982). Evaluation of high-fidelity hearing aids. ***Journal** Of Speech & **Hearing Research,*** 25:15-25.

Killion, M.C., Staab, W. J., & Preves, D.A. (1990). Classifying automatic signal processors. ***Hearing Instruments,*** 41 -24-25.

King, A.B., & Martin, M.C. (1984). Is AGC beneficial in hearing aids? ***British Journal** of **Audiology,*** 18,31-38.

Knight, J.K. (1992). A subjective evaluation of K-Amp vs. linear hearing aids. ***Hearing Instruments,* 43 (10): 8-11.**

Kruger B., & Kruger F.M. (1993). The K-Amp hearing aid: Clinical impressions with fittings. ***Hearing Instruments,*** 44 (2): 30-35.

Lawrence, R.F., Moore, B.C.J., & Glasberg, B.R. (1983). A comparison of behind the ear high-fidelity linear hearing aids and two-channel compression aids, in the laboratory and in everyday Life. ***British Journal of Audiology,* 17, 31-48.**

Lawrence, R.F., Moore, B.C.J., & Glasberg, B.R. (1983). Comparison of behind - the ear high fidelity linear hearing aid and two - channel compression aids, in the laboratory and in every day life. ***British Journal of Audiology,* 28,**

31-48.

Leshowitz, B., Linstrom, R., & Zurek, P. (1975). Psychophysical tuning curves in normal and impaired ears. *Journal of Acoustical Society of America,* 1,58-71.

Lippman, R.P, Broida, L.D., & Durlach N.I. (1981). Study of Multichannel amplitude compression and linear amplification for persons with sensorineural hearing loss. *Journal of the Acoustical Society of America,* 69 (2), 524-534.

Lorente de No, R. (1993). Cited in Silman,S and Silverman, C.A . *Auditory Diagnosis: Principles and Applications,* p. 187, San Deigo, Academic Press, Inc.

Lurie, M.H. (1940). Cited in Silman,S and Silverman, C.A . *Auditory Diagnosis. Principles and Applications,* p. 187, San Deigo, Academic Press, Inc.

Lynn, G., & **Carhart, R. (1963).** Influence of attack and release in compression amplification on understanding of speech by hyperacusics. *Journal of Speech & Hearing Disorders,* 28, 124-140.

Mare, M., Dreschler, W., & Verschuure, H. (1992). The effects of input-output configuration in syllabic compression on speech perception. *Journal of Speech and Hearing Research,* 35, 675-685.

Marshall, C. (2000). When given the choice between digital and K-Amp hearing aids, the best-trained ears in the world have chosen K-Amps. *Hearing Journal,* 53(9), 19.

Miskolezy-Fodor, F. (1960). Relation between loudness and duration of tonal pulses. HI, Response in cases of abnormal loudness function, ***Journal*** *of the*

Acoustical Society of America, 32, 486-492.

Moore, B. C.J., Laurence, R. F., & Wright, D. (1985). Improvements in speech intelligibility in quiet and in noise produced by two channel compression hearing aids. *British Journal of Audiology,* 19, 175-187.

Moore, B., Glasberg, B., & Stone, M. (1991). Optimization of a slow-acting automatic gain control system for use in hearing aids. *British Journal of Audiology,* 25, 171-182.

Moore, B.C.J,, Glasberg, B. R., Donaldson, E., McPherson, T., & Plack, C.J. (1989). Detection of temporal gaps in sinusoids by normally hearing and hearing-impaired subjects. *Journal of the Acoustical Society of America,* 85, 1266-1276.

Moore, B.C.J. (1990). How much do we gain by gain control hearing aids? *Acta Oto-Laryngologica.* Supplement (Stockholm), 469, 250-256.

Moore, B.C.J., & Glasberg, B.R. (1987). Formulae describing frequency selectivity as a function of frequency and level and their use in calculating excitation patterns. *Hearing Research,* 28, 209-225.

Moore, B.C.J., & Glasberg, B.R. (1983). Suggested formulae for calculating auditory-filter bandwidths and excitation patterns. ***Journal*** *of the Acoustical Society of America,* 74, 750-753.

Moore, B.C.J., & Glasberg, B.R. (1988b). Gap detection with sinusoids and noise in normal, impaired and electrically stimulated ears. ***Journal*** *of the Acoustical Society of America,* 83, 1093-1101.

Moore,B.C.J. (1996). Perceptual consequences of cochlear hearing loss and their implications for the design of hearing aids. *Ear and Hearing,* 17, 133-161.

Murray, N., & Byrne D. (1986). Performance of Hearing impaired and normal hearing aids. ***Australian Journal*** *of* ***Audiology,*** 8,21-28.

Nabelek, I. V. (1983). Performance of hearing -impaired listeners under various types of amplitude compression. ***Journal of the Acoustical Society of*** *America, 14,* 76-791.

Nabelek, I., & Robinette, L. (1975). Comparison of hearing aids with amplitude compression. ***Journal*** *of Acoustical Society ofAmerica,57,* 519 (A).

Nelson, D.A. (1991). High-level psychophysical tuning curves: Forward masking in normal hearing and hearing-impaired listeners. ***Journal of Speech and Hearing Research,*** 34, 1233-1249.

Neuman, A.C., Bakke, M.H., Hellman, S., & Levitt, H. (1994). Effect of compression ratio in a slow-acting compression hearing aid, Paired compression judgements of quality. ***Journal*** *of* ***the Acoustical*** *Society* ***of America,*** 96, 1471-1478.

Neuman, A.C., Johnson, J.S., Clark, T.M., & Pluvinage, V. (1992). Evaluation of a dual channel full dynamic range compression system for people with sensorineural hearing loss. ***Ear and Hearing,*** 13, 349-370.

Nillson, P., Vesterager, V., Sibelle, P., Sieck, L., & Christensen, B. (1997). A double blind cross-over study of a non-linear hearing aid. *Audiology,* 36 (6), 325-338.

Painton, S.W. (1993). Objective measure of low-frequency amplification reduction in canal hearing aids with adaptive circuitry. ***Journal of American*** *Academy of Audiology,* 4(3), 152-156.

Pascoe, D. (1975). Frequency responses of hearing aids and their effects on the speech perception of hearing impaired subjects. ***Annals** of Otology,* ***Rhinology** & Laryngology,* 84 (Suppl 23): 1-40.

Patterson, R.D. (1976). Auditory filter shapes derived with noise stimuli, ***Journal** of the Acoustical Society of **America,*** 59, 640-654.

Patterson, R.D., & Moore, B.C.J. (1986). Auditory filters and excitation patterns as representations of frequency resolution. In B.C. J. Moore (Ed.), ***Frequency** selectivity **in hearing*** (pp. 123-177). London. Academic Press.

Patterson, R.D., & Nimmo-smith, I. (1980). Off-frequency listening and auditory filter asymmetry. ***Journal of the Acoustical Society of America,*** *61,* 229-245.

Pederson, C.B., & Eberling, C. (1973). Temporal integration of acoustic energy in-patients with presbyacusis. *Acta Oto-Laryngologica* (Stockholm), 75, 32-37.

Plack, C.J., & Moore, B.C.J. (1991). Decrement detection in normal and impaired ears. ***Journal of the Acoustical Society of America,*** 90, 3069-3076.

Plomp R (1987). The negative effect to amplitude compression in multichannel hearing aids in the light of the modulation - transfer function. ***Journal of Acoustical Society of America,*** **81:148- 154.**

Preves, D.A. (1992). The K-Amp circuit. ***American Journal** of Audiology,* 1, 15-16.

Punch J., & Beck L(1986). Relative effects of low frequency amplification on syllable recognition and speech quality. ***Ear and Hearing,*** 7:57-62.

Rhode, W. S., & Robles, L. (1974). Evidence from Mossabauer experiments for non-linear vibration in the cochlea ***Journal of the Acoustical** Society **of***

America, 55, 588-596.

Rhode, W.S. (1971). Observations of the vibration of the basilar membrane in squirrel monkeys using the Mossbauer technique. *Journal of the Acoustical Society of America,* 49, 121-131.

Robles, L., Ruggero, M.A., & Rich, N.C (1986). Basilar membrane mechanics at the base of the chinchilla cochlea. Input-output functions, tuning curves and response phases. *Journal of the Acoustical Society of America,* 80, 1364-1374.

Ruggero, M.A. (1992). Cited in C.I Berlin, L.I. Hood, A Herley, H., Wen (1996). *Hearing Aids: Only for Hearing Impaired patients with abnormal otoacoustic emission.* In Berlin, C.I (ed.), hair Cells and hearing aids, p. 108, San Deigo: Singular Publishing Group, Inc.

Ruggero, M.A., Robles, L., & Rich, N.C. **(1992).** Two-tone suppression in the basilar membrane of the cochlea: Mechanical basis of auditory nerve rate suppression. *Journal of Neurophysiology,* 68, 1087-1099.

Salvi, R.J., Henderson, D., Flamernik, R., & Ahroon, W.A. (1983). Neural correlates of sensorineural hearing loss. *Ear and **Hearing,*** 4(3), 115-129.

Sammath CA,. & Ochs, M.T (1991). A review of current "noise reduction" hearing aids: Rationale, assumptions and Efficacy, *Ear and Hearing,* 12(6), 116S.

Sataloff, R.T and Sataloff, R (1987). Occupational Hearing Loss. Marcel Dekker, Inc., pp. 200-201.

Schiff, M., & Sandlin, R.E. (1982). What can be done for the recruiting ear? *Annals of otology. Rhinology, Laryngology,* 91, 521-525.

Sellick, P.M., PatuzzL R., and Johnstone, B.M. (1982). Measurement of basilar membrane morion in the guinea pig using the Mossabauer technique. *Journal of the Acoustical Society of America,* 72, 131-141.

Shamma, S., & Morrish, K. (1986). Synchrony in complex stimulus responses of a biophysical model of the cochlea. *Journal of the Acoustical Society of America,* 81, 1486-1498.

Simmons, F.B. and Dixon, R.F. (1996). Cited in Silman,S & Silverman, C.A. *Auditory Diagnosis. Principles and Applications,* p. 187, San Deigo, Academic Press, Inc.

Skinner, M.W. (1980). Speech intelligibility in noise-induced hearing loss: Effects of high-frequency compensation. *Journal of the Acoustical Society Of America,* 67:306-317.

Skinner, M.W., Karstaedt, M.M., & Miller, J.D. (1982a). Amplification bandwidth and speech intelligibility for two listeners with sensorineural hearing loss. *Audiology,* 21:251-268.

Smriga, D.J. (1993). Rationale for non-linear circuitry: part 2 formulating problem solving strategies for selecting amplification. ***Hearing Instruments,*** 44 (11), 26-31.

Soucek, S., & **Michaels, L. (1990).** *Hearing Loss in the Elderly.* Springer-Verlag, London, Preface.

Souza, P.E., Turner, C.W. (1999). Quantifying the contribution of audibility to recognition of compression - amplified speech. *Ear & Hearing,* 20, 12-20.

Stein L., & Dempsey-Hart D. (1984). Listener assessed intelligibility of a hearing aid self-adaptive noise filter. *Ear **and** Hearing,* 5,199-204.

Steinberg, J.C., & Gardner, M.B. (1937). Cited in E. Villchur, Multichannel compression in hearing aids. In C.I. Berlin, (Ed.) ***Hair cells and hearing aids,*** San Diego, Singular Publishing Group, Inc.

Stelmachowicz, PG. Jesteadt, W., Gorga, M.P., & Moti, J. (1985). Speech perception ability and psychophysical tuning curves in hearing -impaired listeners. ***Journal of the Acoustical Society of America,*** *11,* 621-627.

Steven, S.S. (1936). Cited in Salvi, R.J., Henderson, D., Hamernik, R., & Ahroon, W.A. (1983). Neural correlates of sensorineural hearing loss. ***Ear and Hearing,*** *4(3),* 115-129.

Sullivan, J.A., Levitt H., Hwang J., & Hennessey, A. (1980): An experimental comparison of four hearing aid prescription. ***Journal of the Acoustical Society of America,*** 67: 306-317.

Surr, R.K., Cord M.T., & Waldr, B.E. (1997). Comparison of, linear and K-Amp circuits. ***Ear and Hearing,*** 18 (2), 140-146.

Taylor, R.R (1988). Sorting through the capabilities of noise reduction approaches. ***Hearing Instruments,*** 39(10): 22-27, 1988.

Tillman, T.W., Carhart, R., & Olsen, W.O. (1970). Cited in E. Villchur multichannel compression in hearing aids. In C.I. Berlin, (e.). ***Hair cells and hearing aids,*** San Diego: publishing Group, Inc.

Tonndorf, J. (1980). Cited in Silman, S., & Silverman, C.A. ***Auditory Diagnosis: Principles and Applications,*** **pi 88,** San Diego: Academic press, Inc.

Tonndorf, J. (1981). Cited in Silman, S., & Silverman, C.A. ***Auditory Diagnosis: Principles and Applications.*** P. 188, San Diego: Academic **press,**

Inc.

Trocianko, T., & Gregory, R.L. (1984). An assessment of two amplitude-compression hearing aid systems, especially in high ambient noise. ***British*** *Journal of Audiology*, 18, 89- 96.

Tyler, R.S., Wood, E.J., & Fernandes, M.A. (1982). Frequency resolution and hearing loss. *British Journal of Audiology*, 16, 45-63.

Van Harten -de Bruijin., H.E., Van Kreveld -Bos, S.G.M., Dreschler, W.A., & Verschuure, H. (1997). Design of two syllabic nonlinear multichannel signal processors and the results of speech tests in noise. *Ear and Hearing,* 18, 26- 33.

Van Tasell D., Larsen S, Fabry D (1988). Effects of an adaptive filter hearing on speech reception in noise by hearing impaired subjects. *Ear* ***and*** *Hearing,* 9:15-21.

Villchur, E. (1973). Signal processing to improve speech intelligibility in perceptive deafness. *Journal of the Acoustical Society of America,* 53 (6), 1646-1657.

Villchur, E. (1974). Simulation of the effect of recruitment on loudness relationship in speech. *Journal of Acoustical society of America*, 56(5) 1601-1611.

Vinay, S.N., & Rajalakshmi, K. (1999). Volume control taper characteristics of Indian behind-the-ear hearing aids, (in press of Journal of Indian Speech & Hearing Association) presented at the *31st national conference of* ***Indian*** *Speech & Hearing Association.*

Young, L. L., Goodman, J. T. G., & Carhart, R- C. (1979). Effects of whitening and peakclipping on speech intelligibility in the presence of a competing message. *Audiology*, 18, 72-79.

Yund, E.W., and Buckles, KM. (1995). Enhanced speech perception at low signal to noise ratios with multichannel compression hearing aids. *Journal of the Acoustical Society of America*, 97, 1224-1240.

Zeng F., & Turner C.(1990). Recognition of fricatives by normal and hearing-impaired subjects. *Journal of Speech & Hearing Research*, 33: 440-449.

Zwicker, E., & Schorn, K. (1978). Psycho-acoustical tuning curves in audiology, *Audiology*, 17, 120-140.