NON-LINEAR HEARING AIDS - A REVIEW

Register No. M2K15

A Independent Project submitted in part fulfillment for the
First year M.Sc. (Speech and Hearing),
University of Mysore, Mysore

All India Institute of Speech and Hearing,
Manasagangothri, Mysore-570 006

May 2001

Dedicated to
My Guide
Mrs. Manjula

Dedicated to
My Lord
Sri Guruvayurappan
Achan and Amma

CERTIFICATE

This is to certify that the Independent Project entitled **"NON-LINEAR HEARING AIDS - A REVIEW"** is the bonafide work in part fulfillment for the degree of Master of Science (Speech and Hearing) of the student with register number M2K15.

DIRECTOR

All India Institute of Speech and Hearing
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Mysore
May 2001

CERTIFICATE

This is to certify that the Independent Project entitled **"NON-LINEAR HEARING AIDS - A REVIEW"** has been prepared under my supervision and guidance. It is also certified that this has not been submitted earlier in any other University for the award of any Diploma or Degree.

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DECLARATION

This Independent Project entitled **"NON-LINEAR HEARING AIDS - A REVIEW"** is the result of my own study under the guidance of Mrs. P. Manjula, Lecturer, Department of Audiology, AIISH, Mysore and has not been submitted at any other university for the award of any Diploma or Degree.

Register No. M2K15

Mysore
May 2001

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INTRODUCTION

The ear is an extraordinary sound detecting device, so sensitive that it can almost hear the random Brownian movements of the air-particles as they strike the eardrum. Yet it will be amazing to know that such a sensitive organ is able to tolerate the sound waves generated by an entire symphony orchestra! The ear can also respond to a wide frequency and intensity range. The range of audibility is often stated to be from 15 to 16 Hz to about 20,000 Hz and the dynamic range is from 0 to 120dB SPL.

The production of sound is caused by vibrations. The vibrations of sound in the air travel from external to middle ear and then to the inner ear. The vibrations produced can be linear or nonlinear. At very high intensities, the middle ear does become asymmetrical in its behaviour, with the stapes moving outward during rarefaction more than it moves inward during condensation. Asymmetrical systems, of course are always nonlinear (Gullick, 1989).

Sellick, Patuzzi, Johnstone (1982) found that basilar membrane response is nonlinear at and above the frequency of the sharply tuned peak. This non linear response is also seen in sharply tuned intracellular tuning curves of inner hair cells (IHC). They also reported non-linearities in the intracellular receptor potentials of both inner and outer hair cells.

In the inner ear, the organ of Corti, the seat of hearing, has the inner hair cells (IHC) and outer hair cells (OHC), OHCs probably act as amplifier to IHCs for low amplitude acoustic signals. But for high amplitude sounds they act as "compressors". In the ears with recruitment this "compressor action" or "expanding action" of OHC is impaired. So providing amplification, which compensates for normal physiological compressor action is essential. Thus non-linear circuit in hearing aid which has a frequency shaping and output limiting is a method of suitable option, without sacrificing the speech intelligibility. The dynamic range of hearing impaired person gives a valuable criteria for successful hearing aid fitting. The dynamic range in persons with recruitment will considerably be low so that the person finds it difficult to tolerate sounds, i.e., tolerance problem. Non-linear hearing aids are widely used for persons with reduced dynamic range and poor discrimination in noise.

Since the information on nonlinear hearing aids are scattered, an orderly compact compilation of the facts would bring in a handy reference for the audiologists and help them for a successful choice of hearing aid fitting and guide for further modifications to be done in future. Further, it would also help as a study material for students for better understanding about non-linear hearing aids. It could also help in further research of hair cell physiology and nonlinearity of cochlea, which is lost when a person develops hearing loss. The nonlinear hearing aids would bring in a satisfiable answer for their wide range of listening needs.

Investigators have contributed to the nonlinear hearing aids since 1930s and since then there has been a lot of research in improving a nonlinear technology and fitting procedures. The information is scattered in different journals and books. It would be easier to comprehend and remember when all the information are compiled together. Hence, the objective of the present effort is to collect selected articles / book reviews on nonlinear hearing aids so that it is made available for easy reference.

REVIEW OF LITERATURE -

NONLINEAR HEARING AIDS

The goal of fitting hearing aids is to make the speech signal audible and to transfer the long-term average speech spectrum into the residual dynamic range of the individual. If the residual dynamic range is reduced the input signal has to be shaped by output limiting to avoid the output levels exceeding the uncomfortable loudness levels i.e., the hearing instrument should be fitted in such a way that the normal dynamic range of sounds in transformed into the narrow dynamic range of the hearing impaired. This should be accomplished without altering the loudness relationships between sounds.

The line of treatment for the hearing impairment may be medical or surgical or audiological. For those for whom the medical or surgical line of treatment is not useful/effective, audiological management should be instituted. The audiological management includes fitting the hearing impaired with hearing aids, assistive listening devices, cochlear inplants, etc., of which use of hearing aids is common.

A hearing aid is an electronic device that brings sounds more effectively into the ear. From the time hearing aids have been in use, considerable advancements have been made and special provisions are made in hearing aids for those with specific amplification needs.

The extensive modification in the hearing aid performance for a sensori neural hearing loss (SNHL) and the failure of hearing aid to compensate entirely for these changes suggests that there must be fundamental changes in the way impaired ear processes acoustic information.

Nature does not normally provide us with linear hearing, but hearing provides increased sensitivity for quiet sounds and a gradual reduction of sensitivity with increasing level. Thus, some form of non-linearity is provided to increase the gain for low level sounds. The only time the ear shows a linear input-output function is when the outer hair cells are impaired. Linear amplification is thus "pathological". Other important effects of SNHL include broadening of cochlear filtering, impaired temporal resolution, impairment of the efferent neural mechanism that assists the cochlea in processing signal in the presence of noice, etc.

Linear amplification in hearing aids does not provide an effective way of dealing with the reduced dynamic range of SNHL. With linear amplification it is not possible to restore audibility of weak sounds without intense sounds being overamplified and being uncomfortably loud. It was suggested many years ago that this problem could be alleviated by the use of automatic gain control (AGC) (Steinberg and Gardener, 1937). With AGC, it is possible to amplify weak sounds more than strong ones, with the result that the wide dynamic range of the input signal is compressed into a smaller dynamic range at the output. A sincere effort is made to review various advances since late 1930s and a compilation of literature on non-linear hearing aids thus follows.

The review of literature of selected articles / books has been discussed under the following heads :

1. Hearing, non linearity of hearing and hearing loss.
2. Non linearity in hearing aids
3. Selection procedures for non-linear hearing aids.

1. Hearing, non-linearity of hearing and hearing loss : Hearing is an extraordinary sense that it can hear the random Brownian movements of the air-particles as they strike the eardrum. The ear responds to a wide range of intensity and frequency range.

The vibrations of sound in the air travel from the external ear to middle ear to inner ear and through the auditory pathway reaches the auditory cortex for interpretation of sound. At very high intensities of sound, the ear does become asymmetrical / non-linear in its behaviour. (Gullick, 1989).

Normal hearing is based on the conversion of an incoming sound into a nerve code in the sensory cells of the organ of Corti in the inner ear. This transformation is made by the interaction of the two types of sensory cells - the inner hair cells (IHC) and the outer hair cells (OHC).

The OHCs function, in many aspects, like a servo-control. This servo-control mechanism has the effect that a low level or soft sound is able to evoke sufficiently large vibrations of the basilar membrane, which in turn will excite the IHCs. The OHCs function non-linearly, i.e., there is a significant effect of OHC function, whereas more intense signals generate almost no effect in function.

During the acoustico -mechanical excitation, the IHCs transform the mechanical movement of the sound into a complicated pattern of nerve impulses. These nerve impulses are transmitted to the auditory cortex via the auditory pathway.

A hearing loss can be either conductive (damage in the outer and /or middle ear), sensori neural (damage in the inner ear and beyond) or mixed (combination of conductive and sensorineural) types. The dynamic range of hearing remains almost unchanged in an individual with conductive hearing loss. Whereas, the dynamic range is reduced in inner ear pathologies. Thus hearing aids that amplify linearly will pose little problem to the conductive hearing loss and more with hearing loss associated with inner ear damage. Hence non-linear amplification is recommended for the sensorineural hearing loss.

The other consequences of a hearing loss include loss of impaired audibility, recruitment (abnormal growth of loudness), masking (spread of masking), impaired frequency resolution, impaired temporal resolution, loss of non-linearity, etc.

Hearing loss can be classified also as : Type I, II and III, these types are illustrated as follows (Killion,1996):

Type I Hearing Loss - In this type of hearing loss the loudness growth for a typical cochlear loss of 40 dB is illustrated (Fig. 1). Type I loss shows complete recruitment. Loudness sensation for intense sounds is the same as normal, but sounds below 40 dB HL are inaudible. This finding is consistent with a loss of outer hair cell function with normal inner hair cell function. With increasing level above 40 dB HL, loudness gradually returns to normal.

An individual with Type I hearing loss has a loss of sensitivity for quiet sounds, but may have little or no loss of hearing for loud sounds. The hearing loss is restricted to low-level sounds. An individual with a Type I loss does not need loud sounds to be made any louder than they already are; what is needed is gain for low-level sounds in order to make them audible and clear.

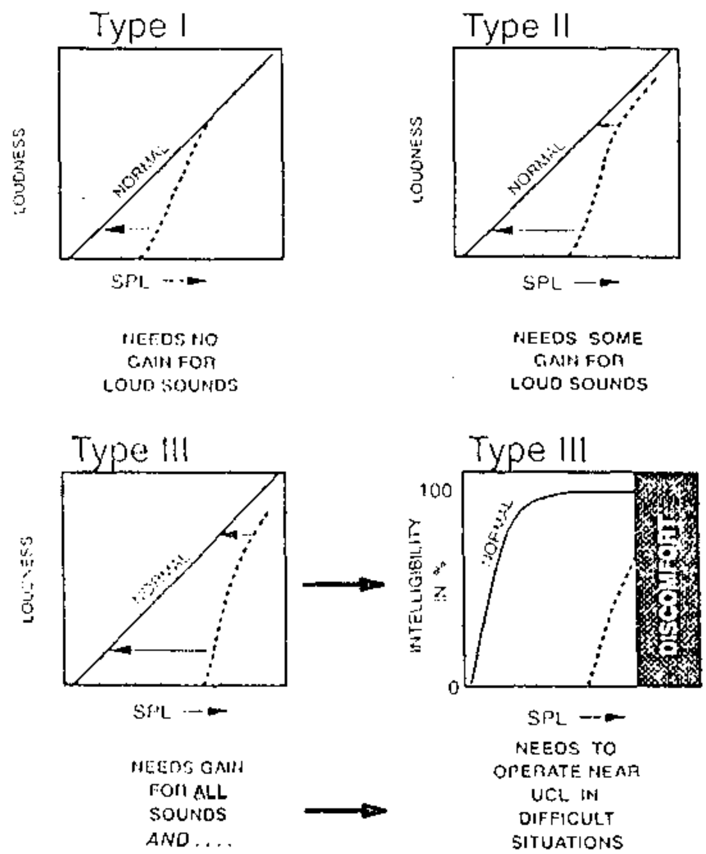

Figure 1. Types of hearing loss

Source : 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 Digo : Singular Publishing Group, Inc.

Type II hearing loss - Here the loudness growth for a Type II loss of 60 dB is illustrated (Fig. 1). A loss of 60 dB is probably too great to explain solely on the basis of a loss of OHC function, and requires that we assume some inner hair cell loss as well. With a Type II loss, there is not only a loss of sensitivity for quiet sounds, but also a loss of some speech cues as well. A loss of IHCs means there is less information available to be transmitted to the brain, even for intense sounds. So not only is more gain required for low-level sounds with a Type II loss, but some gain will be required to restore even loud sounds to normal loudness.

Type III loss - Here both the loudness growth and the intelligibility function for a Type III loss of perhaps 75 dB are shown (Fig. 1). When the hearing loss has progressed into the 70 to 80 dB region, loudness ceases to be primary concern; the IHC loss is so great that one concern dominates; intelligibility.

In hearing loss, it is the OHC that are usually lost first causing recruitment. Electronic compression in hearing aid replaces to some extent the physiological compression that has been lost. Non-linear amplification is recommended for sensori neural hearing loss since the psychoacoustic and neurophysiologic research indicate that several conditions are not addressed adequately with linear amplification.

2. Non-linear aids : All the hearing aids incorporate some means of limiting the maximum output as it is not practical to use linear amplification to compensate fully for the loss of audibility caused by cochlear hearing loss. The function of every hearing aid is to amplify soft sounds strong enough to make them audible, but not to overamplify them to produce uncomfortable listening level. Every hearing aid has a maximum deliverable pressure (saturation SPL) determined by its components (receiver, battery voltage, amplifier). In practice, it is within the amplifier that most limiting occurs, at the saturation of the amplifier. The maximum deliverable pressure the hearing aid can produce, or its limiting level, however can be adjusted below the level of saturation.

Amplifier controls are important part of a hearing aid and are used to modify its basic performance. The output from the amplifier can be controlled electronically or acoustically. The most commonly used control is volume control, which is a variable resistor, to select the most effective listening levels. Master gain adjustments is also possible through trimmer controls.
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Acousticaly, a damper can be used to smoothen the peaks in the frequency response of the hearing aid. Research has implied improved speech intelligibility, greater clarity of speech, increased user satisfaction with dampers in the amplification system (Killion, 1982). Earmolds with different types of venting and tubing could also produce desirable acoustic effects in hearing aids.

Linear amplification means that output changes always bear a direct and constant relationship to the input signal changes i.e., as the input SPL is increased the output SPL increases by the same number of decibels upto a point where SSPL is reached, after which further increases in input do not increase the output. The input-output characteristic is typically 1:1 dB ratio, with a constant gain (Fig. 2). Any deviation from this would no longer be linear.

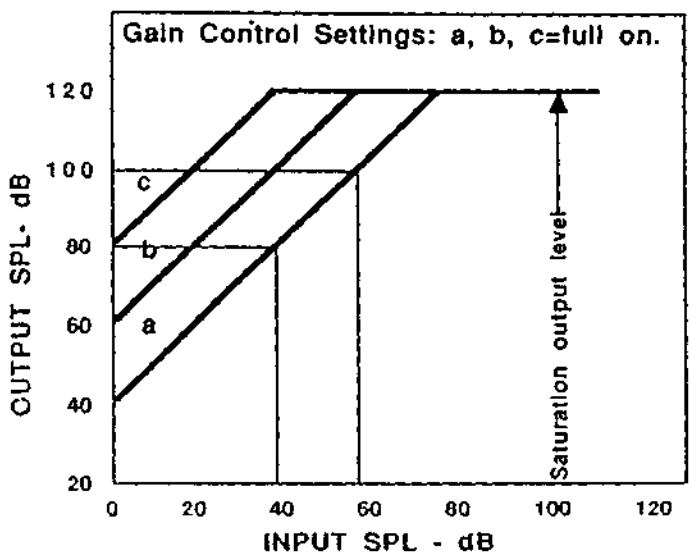

Figure 2 : Input/output curves for a linear hearing aid. As the gain control is advanced (from a toc), the gain in the linear portion of the curve increases, but when the saturation level is reached, no further increase occurs.

Source : Statjb, Lybarger (1994)

Most linear hearing aids reach saturation when the input is above 90 dB SPL (Fig. 2). At saturation levels, considerable harmonic distortion results but most cases presents in problem for the wearer if the aid has been selected the user's Loudness Discomfort Level (LDL). Recent research suggests that linear hearing aids having a higher SSPL90, thus high headroom, are less likely to exceed the user's LDL than do aids having lower SSPL90 but higher distortion (Fortune, Preves, Woodruff, 1991; cited in Katz, 1994). The aid with higher headroom and less distortion will saturate less readily, sounds clearer at high levels, and will likely to be worn at gain settings high enough to provide the user greater benefit. With respect to aided Uncomfortable Level (UCL), hearing aid users equate distortion to loudness.

Linear hearing aids preserve the temporal nuances of the input signals that are especially critical for these hearing aid wearers, its use is limited at both low-input and high input levels without adjustment of the volume control. At low input levels, perception of soft speech, listening from a distance, is a problem, more so for children who are unable to adjust the volume control on their own. At high input levels, there is less specificity in stimulation, increased temporal distortion and increased risk of overamplification.

However, not all types of hearing impairment would be helped with such devices. The conductive hearing impairment or Type I cohlear hearing impairment may be helped. The other types of hearing losses, especially with recruitment, may not be helped with such devices, and would require some form of output limiting, a nonlinear functioning.

Nonlinear hearing aids are hearing aids whose gain at some or all frequencies depends on input loudness even when the input level is not sufficient to cause the output to peak clip or compression limit.

The advent of wearable nonlinear hearing aid since 1970s made a step forward in hearing aid technology. Research has confirmed both that listener prefer nonlinear hearing aids (Hawkins and Naidoo, 1993) and that speech recognition is improved with non-linear hearing aids, particularly for those with narrower dynamic range (Dillon, 1996).

For patients with sensori neural hearing loss, perceived loudness increases disproportionately as the signal SPL increases, i.e., the loss of sensitivity to sounds varies with frequency in a different manner depending on the sound intensity. This abnormally rapid growth in loudness results in degradation of the incoming speech signal. Near threshold, the sensitivity is well described by the audiogram. Above threshold, however, the sensitivity loss is progressively reduced and the shape of the optimum frequency gain characteristic changes.

Limiting Systems in Hearing Aids : Part of the function of every hearing aid is to amplify soft sounds strong enough to make them audible but not to over-amplify them to produce an uncomfortable listening level. It is this upper level of amplification that limiting systems address. Every hearing aid has a maximum deliverable pressure (saturation, overload) determined by the receiver, battery voltage, and amplifier. In practice, however, it is within the amplifier that most limiting occurs, at the saturation of the amplifier. The maximum deliverable pressure the hearing aid can produce, or its limiting level, however, can be adjusted below the level of saturation.

A. Limiting as a Result of Instant Output Regulation

The limiting systems include peak clipping and peak rounding. Both involve limiting amplitude at a certain point but in some what different manner.

Peak clipping : Peak clipping for output limiting usually is implemented in power output stage of the amplifier, it may also occur unintentionally in other sections of the amplifier. Due to inadequate headroom, many linear hearing aids saturate via peak clipping at levels significantly lower than the loudness discomfortable level of the hearing aid user. However, peak clipping has undesirable distortions which causes poor sound quality and lack of clarity (Agnew, 1988). It makes the listening experience annoying and fatiguing that, in noisy environments, hearing aids are turned off or even removed. It is this that may lead to outright rejections of hearing aids (Fig. 3).

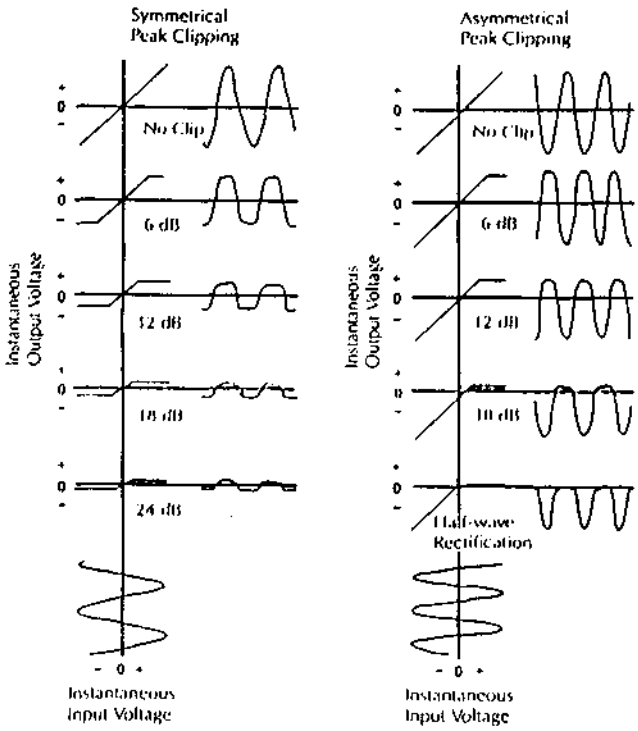

Figure 3 : Examples of asymmetrical and symmetrical peak clipping (after Licklider 1946). Source : Preves, D.A. (1991).

The type of distortion introduced by peak clipping depends to a large extent on how clipping is performed. The levels at which peak clipping occurs can be made variable in most amplifiers by introducing diodes, resistors, and / or transistors (Keller, 1984). Variable peak clipping is implemented with different resistor values in series with the receiver in a class A output stage. This results in asymmetrical peak clipping with attendant gain reduction and produces both even and odd harmonies. Whereas resistor peak clipping produces a strident sound, diode peak clipping generally produces symmetrical peak clipping with little or no accompanying gain reduction. Diode peak clipping has been called soft peak clipping because it usually sounds less harsh than resistor peak clipping.

The literature is mixed on how asymmetrical and symmetrical peak clipping affect intelligibility. Licklider (1946) stated that neither type of peak clipping resulted in a significant reduction in intelligibility even with an 'infinite' amount of clipping, whereas, Krebs (1972) and Keller (1984) opine that even harmonics are more detrimental to speech intelligibility than odd harmonics.

Peak Rounding : (soft peak clipping, curvilinear compression, diode compression control, diode clipping, modified peak clipping) is a form of non-linear amplification that is evidenced by a gradual, ever-diminishing increase in output with each successive increase in input. Limiting is achieved by adding a minimal number of components (two diodes, a capacitor and variable resistor) to hard peak clipping circuitry to form a negative feedback loop that creates a nonlinear resistance and which most often (newer circuits now allow for appreciable reduction of $SSPL_{90}$ without reducing gain) reduces both the gain and the output power. The variable resistor allows for the amount of negative feedback to be adjusted. The results resemble hard peak clipping in many respects, including the creation of harmonic distortion, except that the onset of clipping is gradual and the distortion arising from symmetrical peak clipping, while beginning at lower levels, is not as severe.

Both hearing impaired and normal subjects showed progressively higher SRT increasing levels of peak clipping with significant threshold shifts occuring for clipping levels greater than 18-24 dB (Thomas, et al.,1944). Olsen (1971) reviewed earlier literature and found suggestions that speech intelligibility is not disrupted appreciably by peak clipping speech intelligibility is not disrupted appreciably by peak clipping and its resultant harmonic and intermodulation distortion. Staab (1972) indicated that even excessive peak clipping does not significantly decrease speech discrimination ability. Both these reports indicate that there is no great reduction speech intelligibility when peak clipping is used as a output limiting device though speech quality may determinate. The result of Young Goodman and Carhort (1979) study on effects of whitening and peak clipping on speech intelligibility is reduced by whitening and peak clipping when more than one talker is present. Such a finding has inplications on wearable amplification.

B. Limiting by Time-Dependent Gain Regulation; Feed-back Circuits, Rectifier, Circuits, Adaptive Hearing Aids, Automatic Gain Control (AGC) : These systems have a built-in monitoring circuit that automatically reduces the electronic gain of the hearing aid as a function of the magnitude of the signal being amplified. Gain is reduced by means other than peak clipping. Two major purposes of these systems are : (1) to reduce the gain of an aid as the input SPL increases so that the output capability of the aid is not exceeded and distortion is thus kept low, and (2) to reduce the dynamic range of the output so that it is a better match to the dynamic range of an impaired ear. Gain level is controlled automatically, thus the name automatic gain control (AGC). This action may be further described as the process of "compressing" a given dynamic range into a lesser dynamic range, but the circuit actions claimed by different manufacturers have led to confusion in use of the term compression.

A wide variety of AGC characteristics exists in hearing aids, providing for individual differences. The generalized input / output curve of an AGC aid has three main components : a linear section at lower input SPLs where increments of input SPL cause smaller increments in the output SPL; and a limiting section, where increments in input SPL do not significantly increase the output SPL. (Refer Fig. 4)

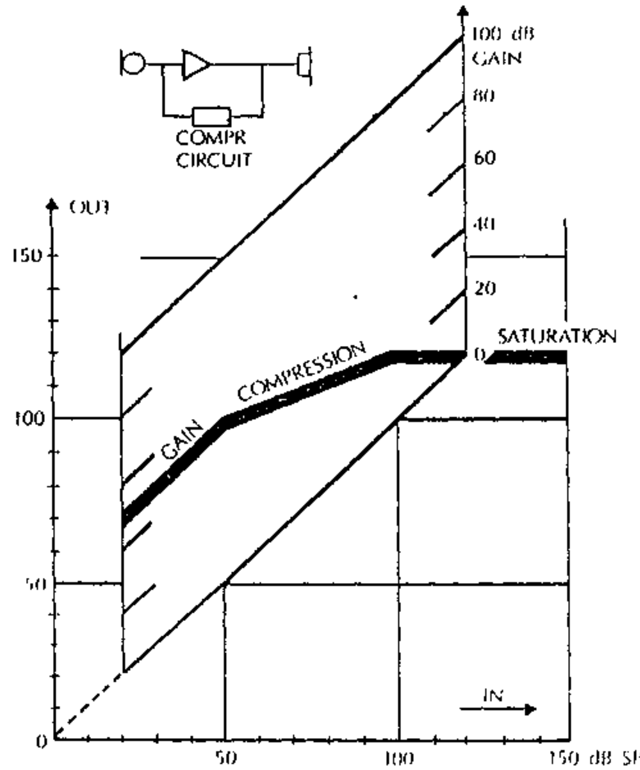

Figure 4 : Compression as that portion of the input/output characteristic between gain and saturation (Nielsen 1973).

Source : Preves, D.A. (1991)

Terms used to describe AGC system characteristics include :

1. **Limiting level**: The level to which the saturation output of the hearing aid is limited. This is also known as AGC knee-point the breakaway point, compressor threshold or AGC threshold. It is the point at which the curve first departs by 2 dB on the output SPL scale from the extension of the linear portion of the curve that exists at input levels below the compression

or limiting portion. The level at which the AGC knee occurs differentiates high level AGC from low level AGC systems.

2. **Compression range** : The range encompassed between the threshold of compression and the SSPL90 of the hearing aid.
3. **Compression ratio** : Identified also as the stiffness ratio or degree of compression, it is the quotient of a change in level of the input divided by the corresponding change in level at the output in the compression portion of the curve.
4. **Slewing rate** : This is the rate of gain change in msec. / dB the circuit can handle.
5. **Time constant** : These are the time lags caused by the feedback circuit in stabilizing to a new gain value. The attack time refers to the length of time required for the feedback circuit to set the new gain value following a strong input signal. The release time refers to the length of time required for the reduced gain to return to normal amplification after the strong input signal is no longer present. Release times must be slower than attack times to avoid what is known as AGC flutter. If too fast, the compression action would follow the instantaneous amplitude of individual cycles, thus introducing severe waveform distortion. IS : 10776-1984 defines attack time and release time as :

Attack time : The time interval between the moment when the input signal level is increased abruptly by a stated number of decibels and the moment when the output sound pressure from the hearing aid with the AGC circuit stabilises at the elevated steady state level within + 2 **dB.**

Release time : Refers to the length of time required for the reduced gain to return to normal amplification after the loud signal is no longer present.

The compression characteristics make use of relative terms such as high / low, short / long. The release time is considered to be long if it is longer than a typical syllable duration for eg : 200 msec.

Compression threshold is considered to be low if speech at normal input level (eg : 65 to 70dB SPL) causes the compressor to activate for a significant preparation of time. Compression ratio is high if it has a value greater than 5, so quite large variation in input level are needed to produce significant variation in output level.

Eight different compression systems can be constructed by taking various combinations of the compression parameters :

Five of them are more useful

	Low Compression threshold		High Compression threshold	
	Low compression ratio	High compression ratio	Low compression ratio	High compression ratio
Short time constants	Whole range syllabic compression	Whole range syllabic compression	-	Compression limiting
Long time constants	Slow acting automatic volume control	Slow acting automatic volume control	-	-

Table 1 : Showing different compression types.

Whole -Range Syllabic Compressors

Syllabic compression circuits are designed to follow the temporal pattern of speech and to reduce the dynamic range of listeners hearing. Hearing aids with syllabic compressor are, in general, characterized by low compression thresholds, low compression ratios and fast attack and release time. In addition to compressing speech sounds syllabic compression circuits may also provide better protection against transient sounds than do AGC circuits. Niederjohn and Grotelueschen (1976) studied the effects on speech intelligibility, in high levels of competing noise, of high-pass filtering followed by automatic amplitude normalization (syllabic compression). They hypothesized that, unlike infinite amplitude clipping, syllabic compression does not introduce harmonic distortion and so may result in better processed sound quality. Additionally, they pointed out that because of its quick response time, the compressor would produce an attenuation of high-level vowel energy which would tend to increase the CVR. They found significant intelligibility advantages for high-pass filtering followed by amplitude normalization compared to unprocessed speech and to high-pass filtering followed by infinite amplitude clipping.

Bustamenta and Braida (1987) also found that, depending on the amount of compression high-frequency emphasis, intelligibility of vowels or consonants would be enhanced. They concluded that high-pass filtering preceding single channel, wideband syllabic compression provides an advantage over linear amplification for lower input level signals but not for higher input level signals. Better speech intelligibility for high-pass filtering and compression has been found by Vargo (1977) compared to linear processing, only at 10 dB SL presentation level, but not at 20 or 30 dB SLs. Dreschler (1988) indicated that syllabic compression equalizes levels between successive sounds, thus bringing up consonant levels. This is especially valuable for persons with severely reduced dynamic ranges. He concluded that for better speech perception, the compression threshold should be set below the mean presentation level.

Walker and Dillon (1982) and Dreschler (1988) speculate that the benefits of recruitment compensation may be nullified in effect by temporal distortions from the compressor attack and recovery times and its alterations of the normal intensity cues of speech. The effectiveness of syllabic compressors as output limiters is somewhat questionable because of their low compression ratios. For this reason, King and Martin (1984) recommended that syllabic compression employ lower compression thresholds so that louder sounds do not exceed the listener's loudness discomfort level.

Compression Limiters

A compression limiter generally employs a much higher compression ratio than the syllabic compression and may thus be a more effective limiter in terms of controlling the peak output SPL from a hearing aid. Compression limiters frequently are used in place of peak clippers in hearing aids to limit SSPL90 without producing harmonic distortion. Because compression limiting is only active at high signal levels, it may provide output limiting with some CVR enhancement without significantly altering the dynamics of conversational speech signals compared to the effect of a syllabic compressor (Caraway and Carhart 1967; Walker and Dillion 1982).

In one of the first uses of compression Edgardh (1952) recommended a compression limiter for hearing-impaired listeners to equalize the dynamic differences between consonant and vowel levels. Peterson, Feeney and Yantis (1990) concluded from recordings of processed speech through a single channel AGC hearing aid that AGC improves speech perception of low intensity high-frequency fricatives and stops, in quiet, in comparison to linear amplification for those hearing-impaired listeners with significantly reduced dynamic ranges. However, they found that this advantage was not shown for listeners with larger dynamic ranges. They also concluded that the single channel AGC hearing aid produced proper speech perception, as compared to linear amplification, for low-intensity, high-frequency fricatives and stops in cafeteria babble at a single SNR. The compression threshold for the hearing aid employed in their study was set at 60 dB SPL so that it was always compressing for the 75 dB SPL speech signal utilized. While this situation might be categorized as an example of syllabic compression, the compression ratio was 10:1, more typical for output limiters, and the recovery time of the compression circuit was 180 ms. This recovery time is much slower than the suggested in most studies for syllabic compressors so as to prevent the AGC time constants from affecting the dynamics of the speech signal.

Fabry and Olsen (1991) tested 15 subjects with mild to moderate hearing impairment objectively in lab and subjectively following trial periods. Each wore a FDRC for a month and then a linear compression limiter for one month. No preference could be shown either objectively or subjectively.

Slow acting AVC

Slow acting automatic volume control (AVC) systems are useful in limiting maximum output while preventing harmonic distortion from peak clipping saturation. Although this type of compression commonly is used in cassette tape recorders for keeping the signal level fairly constant, it has not been widely used in hearing aids. In a hearing aid application, this feature may reduce the need for hearing aid wearers to adjust the gain control over a wide range of environmental input levels. There is scant research in the compression literature as to the viability of employing slow acting AVC in hearing aids. King and Martin (1984) showed that slow acting AGC with a 1-second recovery time with a high compression threshold was preferred by listeners over AGC circuits with a faster recovery time. They also point out that the exact compression ratio is not critical, except for persons with very restricted dynamic ranges.

With low compression ratios, it is not atypical for input AGC to reduce input signal dynamic range from 30 dB to 70 dB (Dreschler 1988). Dreschler, Eberhardt, and Melk (1984) were not able to find a difference in speech reception threshold in quiet or in noise between a linear hearing aid and two AGCo and two AGCi hearing aids. In this study, to cover all biases, evaluations were performed both with the gain controls fixed and with the subjects adjusting the gain controls for the AGCi hearing aids. Likewise, King and

Martin (1984) found no preference between AGCo and AGCi as long as the AGCo compression threshold was suitable for each individual. Dillon and Walker (1983) recommend either AGCo or AGCi for whole-range syllabic compressors, AGCo for compression limiters, and AGCi and slow acting AVC systems.

Adaptive Compression

Adaptive compression system would be one that has a variable release-time capability, applying the appropriate release duration for each of the variety of listening situation that a user encounters. This dual time constant technology has been applied to hearing aid industry in the form of adaptive compression.

The adaptive compression circuit is distinguished by an automatic variable release time duration. The length of this release time is controlled as a function of the compression activating input. The shorter the compression activating input the shorter the release time and the longer the compression activating input, longer the release time. Only adaptive compression has the variable release action necessary to maximize speech intelligibility and sound quality in all listening situations. Variable time release systems are said to be effective for suppressing annoying impulsive sounds.

Fabry, Stypulkowski (1993) studied 2- band processors utilizing compression or linear processing in either band. For noisy backgrounds the linear processing in the high band was superior to compression.

Automatic gain control (AGC) action is fundamental to all circuits within the category of limiting by time-dependent gain regulation, even though additional functions may be incorporated. Because the circuitry is designed to act "automatically" to changes in the input stimulus and to process the incoming signals in some predetermined manner, it has more recently been termed automatic signal processing (ASP).

The advantage of AGC would likely be apparent to a listener who encounters listening environments that frequently vary in level, such as individual might find gain control adjustments unnecessary most of the time. One disadvantage of AGC is that weak elements of speech can be lost if the gain reduced by compression in not restored quickly enough toc amplify these sounds to audibility. Another possible problem is that sound quality may deteriorate if the AGC compression ratio is too high (Neaman et al, 1994).

Some studies have indicated improvement in speech recognition using AGC relative to linear amplification (Moore, Lawrence, and Wright, 1985) while other have indicated either no difference or poorer performance with

AGC **(hippmann,** Braida, and Darlach, 1981; Boothroyd, Spunyer, Smith, and Schulman, 1988). Peterson, Feeney, and Yantis (1990) reported that AGC significantly improved speech perception in quiet for listeners with narrow dynamic range. King and Martin (1984) reported hearing -impaired listener preferred the sound quality of AGC over linear amplification at high listening levels, in quiet and in noise. AGC can benefit hearing and was by allowing them to listen to a wider range of sound levels without either stain or discomfort and if time constant cue well chosen, without adverse effects on speech intelligibility in quiet or in noise.

Automatic Signal Processing (ASP): A differentiation of ASP circuits can be made as follows : Those circuits that reduce gain at high levels and / or increase gain at low levels but do not change the frequency response of the hearing aid in the process, include the "traditional" automatic signal processing circuits (i.e, the AGC or compression circuits). Well-defined terms for these fixed frequency response (FFR) automatic signal processing circuits already exist. More recently developed circuits that automatically change not only the gain but also the frequency response of the hearing aid as a function of the input signal are more accurately identified as level dependent frequency response (LDFR) circuits. Because of the variety of ways in which LDFR is performed, and because no simple, rigorous terms were used to describe their action, Killion, Staab, and Preves (1990) proposed a classification to distinguish them. This classification distinguishes among these circuits in terms of their reaction to low-level rather than high-level inputs because the former yielded easy to remember acronyms (Refer Fig.5).

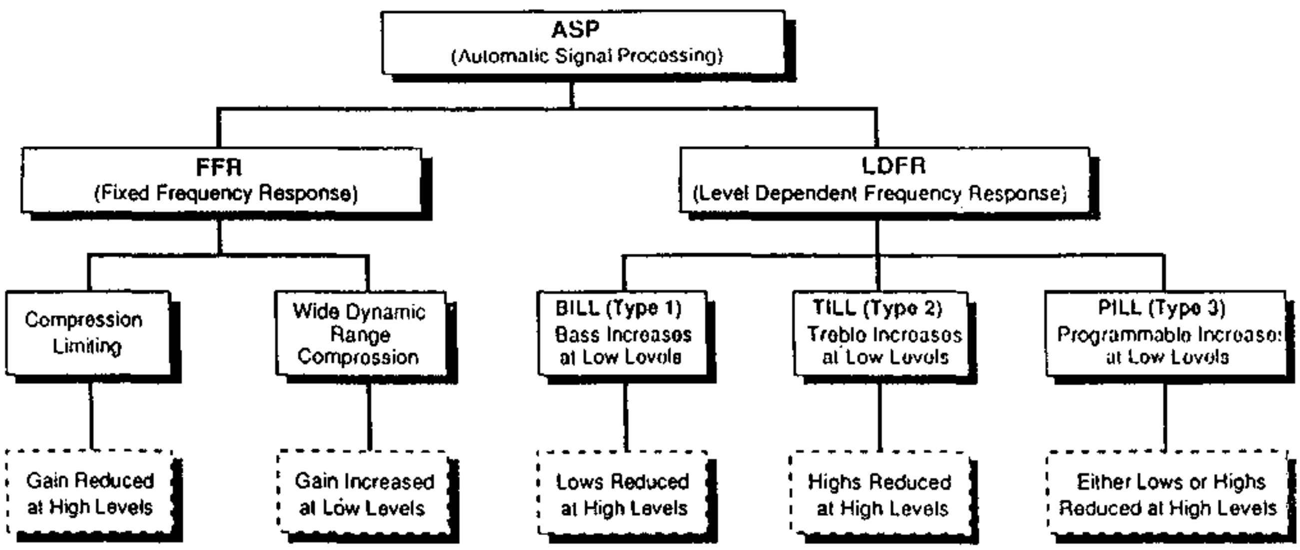

Figure 5 : Outline of recommended classification system for ASP type hearing aids.

Adapted from Killion M, Staab W, and Preves D. Classifying automatic signal processors. Hearing instruments, 1990 ; 41:24-26.

Types of LDFR

BILL (Type 1). Bass increases at low levels (bass decreases at high levels). This type of ASP describes circuits that provide relatively more bass response for low level inputs than for high-level inputs. These circuits are intended for wearers who frequently find themselves in noisy environments, especially environments where low-frequency noise predominates. Under these circumstances there is evidence to indicate that reducing the low-frequency response may be helpful (Crain, 1988). If the overall gain of the instrument is also automatically reduced for high level inputs so as to prevent overload distortion, a further improvement is obtained (Preves and Newton, 1989).

Zeta Noise Blocker (ZNB), which is the noise reduction chip is an implementation of an adaptive Weiner filter that was small enough to be marketed in ITE hearing aids. It consist of a custom -designed special-purpose digital computer implemented with a combination analog digital CMOS integrated circuit. The ZNB use analog-switched capacitor filter and attempted to separate speech from noise in several frequency bands using speech temporal, differences between speech and quasi -steady state noise. The noise was scored during the silent intervals between vocal cord pulses of speech.

Till, Treble increases at Low Levels (Treble decreases at high levels) : This type of ASP describes the operation of circuits which provide relatively more treble response for low-level inputs than for high-level inputs. The K-AMP (Killion, 1990) is this type of circuit and is intended for wearers having high- frequency hearing loss but who need more high-frequency gain for quiet sounds than they do for loud sounds. The amount of high-frequency gain that might produce a harsh or shrill sound in a linear hearing aid may become quite acceptable if the treble boost is automatically reduced for high level inputs. If the overall gain is also automatically reduced for high input levels, it is possible to prevent audible distortion under nearly all listening conditions.

Kruger and Kruger (1993) in their study, reported that experienced hearing aid wearers were generally successful K. Amp hearing aid wearers. They preferred the K-Amp device to their previous hearing aids with linear/or wide band 2:1 compression circuitry.

Some of the experienced wearers preferred to have an operational K-Amp "linear" switch. They report using the linear mode which preserves the low-level K-Amp features but provides greater gain for more intense sounds in most situations and the K-Amp mode in noisy situations.

PILL, Programmable increases at Low levels. This type of ASP describes for operation of circuits which provide for programmable level dependent frequency response modification in more than one amplification band and can be adjusted to provide either bass response decreases with increasing level or treble response decreases with increasing level (Waldhauer and Vilchur, 1988). This form is the most versatile of the three because each of its independent processing bands can ignore strong influences in another band.

Mutlichannel compressors

To eliminate the problem of single channel compressors in which gain across the entire frequency range is reduced by a low-frequency noise, a multiband compressor having independent AGC circuits for each frequency band has been utilized. Such a system is thought to be superior to a single band compressor, especially for severely hearing - impaired persons with extreme loudness recruitment (Villchur 1973; Waldhauer and Villchur 1988). With multichannel compression, low-frequency noise, theoretically, would cause gain reduction only in the low frequency band(s) and the weaker high-frequency components of speech, critical for good speech intelligibility, would continue to be maximally amplified (Kates, 1986). In addition, severe loudness recruitment in the high frequencies may be compensated for by a separate high-frequency band compressor (Goldberg 1982; Laurence, Moore, and Glasberg, 1983).

Several reports have documented the benefits of multichannel compression hearing aids over single channel and linear hearing aids (Yanick 1976; Mangold and Leijon 1981; Goldberg 1982; Laurence, Moore and Glasberg 1983; Moore and Glasberg 1986; Moore 1987; Moore and Glasberg 1988). There also are studies that have generally failed to demonstrate significant improvements in speech intelligibility in noise using multichannel AGC systems over single channel AGC and linear amplification hearing aids (Barrford 1976; O'Loughlin 1980; Abramovitz 1980; Lippman, Braida, and Durlach 1981; Byrne and Walker 1982; Nabelek 1983).

Bustamante and Braida (1987) compared three approaches to syllabic compression : single channel wideband, multichannel with 16-bands, and principal components. The authors controlled two short term principal components : overall level and spectral tilt. It was thought that control of these components would greatly reduce the range of speech level variations without reducing the peak / valley structure of the short-term speech spectrum. They found that over a 10 to 15 dB range of input levels, multiband compression did not provide higher intelligibility than either carefully shaped linear amplification or wideband compression (from single channel compression and first principal component compression); manipulating the second principal component, spectral tilt, degraded speech intelligibility relative to linear amplification, possibly due to reduced spectral differences between weak fricatives.

High compression ratios with multichannel AGC may degrade the relative intensity cues required to identify stops or fricatives (De Gennaro, Braid, and Durlach 1986; Plomp 1988). The problems with interference of the speech signal by syllabic compressors may apply even more strongly to fast-acting multichannel AGC (King and Martin 1984). The optimal time constants for multiband compression in hearing aids are still debatable in terms of whether the circuit should react to and correct for the rapid level changes of speech - a kind of deliberate distortion of the speech signal in itself or whether the natural temporal and spectral cues of speech should be preserved (Braida, Durlach, Gennero, Peterson, Bustamante 1982; Plomp 1989). Fast-acting multichannel compressors with many independent compression bands reduce the natural amplitude contrasts in the speech signal (Plomp 1988). Therefore, Plomp contends that longer compressor time constants be used with multichannel AGC. However, Villchur (1989) reminded us that Licklider and Pollack (1948) showed that speech was perfectly intelligible after infinite amplitude clipping, working with a signal with no amplitude contrasts. Villchur stated that although multichannel AGC decreases the peak-to-valley level differences within speech, the audibility of weaker components of speech such as consonants may be preserved after compression. He concluded that only field experience with 2-channel compression will prove the viability of multichannel AGC.

Frequency - Dependent Compression (FDC)

Although not a true multichannel compression system, a single channel AGC system with frequency-dependent compression provides a restricted amount of control over compression threshold by frequency. In frequency - dependent compression the compression threshold changes as a function of frequency. In most implementations of single channel compressors, a greater low-frequency SPL than high-frequency SPL is required to cause a gain reduction in the high frequencies. This pseudomultichannel AGC effect is achieved by making the compression threshold higher in the lower frequencies, than in the higher frequencies. This feature is easily provided with a high-pass filter near the front end of the amplifier. This filter may take the form of a fixed low-frequency slope in the preamplifier, a variable tone control, or a high-pass microphone response. Virtually every hearing aid that incorporates a tone control prior to the AGC sensor in the forward path of the hearing aid has frequency-dependent compression (Walker and Dillon, 1982).

Recently, frequency - dependent compressors have been developed with the high-pass filter in the AGC feedback loop (Killion, 1988). These systems have higher compression threshold for low frequency energy than for high-frequency energy (Walker and Dillon, 1982). However, they also exhibit a pseudomultichannel AGC effect by lowering high-frequency gain more than low-frequency gain when the input signal level is increased (Killion, Staab, and Preves, 1990; Dillion, 1988).

Multiple Dynamic Compression Control (MDCC)

An optimal communication should offer a variety of signal processing and output limiting. The options can then be used selectively by the user depending on the listening situation. This is done using MDCC, which is used in programmable hearing aids. MDCC offers five signal processing ouput limiting option.

1. Compression limiting with adaptive release times.
2. Compression limiting with fixed release times.
3. Peak clipping
4. Soft peak clipping
5. WDRC

Each option can be programmed independently in each memory. The main advantage is its ability to provide the appropriate form of amplification for patient in most if not all listening situations (Kuk, 1999).

Multi focus hearing aid

A form of adaptive non-linear amplification monitors user's sound environment for fluctuation in sound intensity, it responds to a range of subtleties allowing the user to perceive a whole new dimension of depth and space. Sounds will be audible (distant and soft sounds) and loud sounds are never too uncomfortable.

The technology of compression is applied not only in devices for the hearing impaired, but also for normal hearing individuals with hyperacusis (a device exists that alternates high level sounds without making low level-sounds inaudible) and those working in noise area (active hearing protector amplitude sensitive devices).

Rintlemann (1972) reported a substantial benefit to intelligibility with compression to no compression in a few whereas in others minimal or no improvement in speech perception was found with amplitude compression. Blegvad (1974) found that one third of forty-two selected patients chose the compression amplifier hearing aid, while the remainder preferred the conventional amplifier. Compression is 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. The results of the study by Lawrence, Moore and Glasberg

(1983) indicated that compressor aids were superior for discrimination over wide range of sound levels. If compression limiting is used output control is preferable. On the other hand if AVC is used input control is preferable (Dillion and Walker, 1983). Extending the compression to much lower input levels appears to carry more disadvantages than advantages at least for clients with mild and moderate hearing losses, when fitted with single-channel compression aids with a 2:1 compression ratio (Baker and Dillon, 1999). Speech recognition advantages of WDRC hearing aids are believed to be a direct result of improved audibility. Compression does not introduce determental changes to the speech signal that offset the benefits of improved audibility. (Esouza and Turner, 1999).

The speech in quiet at a comfortable level, no compression scheme yet tested, offer better intelligibility than individually selected linear amplification. In broadband noise, only one system, containing wide band compression followed by fast acting high frequency. Compression, has so far been shown to provide significant intelligibility advantages. (Dillion, 1996). Negative effects of multichannel compression (MCC) were found only for very extreme MCC condition. MCC processing with compression ratios adjusted in each channel for the individual subject, and having as many as thirty-one channels revealed no negative effects on vowel or voiced stop-consonant discrimination (Crain and Yund, 1995).

The adaptive procedure for fitting multi-band compression hearing aids, based on use of speech stimuli the results are encouraging. The procedure can produce fitting of a 2-channel compression system that is more satisfactory than those derived using manufacturers algorithm (Moore, Alcantara and Glasberg 1998).

WDRC processing, fitted using the DSL (i/o) method has potential application in hearing aid fittings for listeners with moderate to severe loss because it provides an audible, comfortable and tolerable amplified signal across a wider range of inputs than linear gain processing, without the need for volume control adjustments (Jenstad, Pamford, Seewald and Cornelisse 2000).

Infinite Amplitude Clipping (IAC)

IAC is not simply asymmetrically or symmetrically chopping off the tops of high level signals with 5 or 10 dB of peak clipping. This amplifies even the very lowest levels of a signal between its zero crossings being amplified to the limits of the circuitry. This results in signal expansion at low input levels and signal compression at high input dynamic range of a signal. Hence this may be suitable for recruiting ears with narrow dynamic range.

Licklider and Pollack (1948) showed that IAC preceded by high pass filtering improved speech intelligibility in white noise for normal hearing subjects. They used a single pole high pass filter having a 16 kHz cut off frequency prior to clipping. The white noise was added after clipping.

Niederjohn (1979), suggested based on data obtained from several studies, that

1. The high pass filter / clipper process results in greater speech intelligibility than high pass filtering alone at higher S/N ratios.
2. High pass filtering alone results in greater speech intelligiblity at lower S/N ratios.

Other ways of controlling the output of the hearing aids is to include a hearing aid gain control. This technique of output limiting is not expected to provide speech enhancement. To the contrary, it may detract from speech intelligibility, discomfort from high input levels, especially in background noise.

Lowering the gain of the hearing aid may be accomplished automatically with compression (AGC). This accomplishes output limiting without introducing harmonic distortion. Compression systems have a finite dynamic range of input levels they can function over, especially with limited headroom available in hearing aid circuitry; when the input level becomes so high as to permit the signal to exceed the range of compression, saturation occurs; i.e., compression is a part of input / output function between gain and saturation (Fig. 4).

There have been conflicting reports on comparison of compression to linear processing and while attempting to optimize compression parameters. Possible reasons for the lack of consensus are

1. Comparing results for subjects with different types of auditory system, pathologies, different hearing loss configuration and different dynamic ranges.
2. Studying the effect of varying compression parameters with different types of recorded speech stimuli, some being precompressed during recording.
3. Comparing data from hearing aids having different attack and recovery times, different compression ratios, different kneepoints.
4. Comparing results from studies with actual hearing aids and lab equipment.

ELECTROACOUSTIC MEASUREMENT OF AGC HEARING AIDS

Procedures different, from that used for linear hearing aids, have to be used for measurement of electroacoustic performance of a non linear of AGC hearing aids. The difference is in terms of stimulus parameters (level of input signal type of signal duration of signal etc) and measurement procedure. According to BIS-1984 standard document, the input level of signal for measuring the gain of AGC is 50dBSPL (rather than 60 dBSPL for linear hearing aids).

Terms as per BIS : 10776 : 1984 (Part 3):

Automatic Gain Control (AGC) - A means in a hearing aid by which the gain is automatically controlled as a function of the magnitude of the envelope of the input signal or other signal parameter.

Study State Input Output Graph - The graph illustrating the output sound pressure level as a function of the input sound pressure level for a specified frequency, both expressed in decibels on identical linear scales.

Lower AGC Limit or AGC Threshold - The input sound pressure level which, when applied to the hearing aids, gives a reduction in the gain of 2 ± 0.5 dB with respect to the gain in the linear mode.

Compression Ratio (Between Specified Input Sound Pressure Level Values) - Under steady state conditions, the ratio of an input sound pressure level difference to the corresponding output sound pressure level difference, both expressed in decibels. (Fig.6)

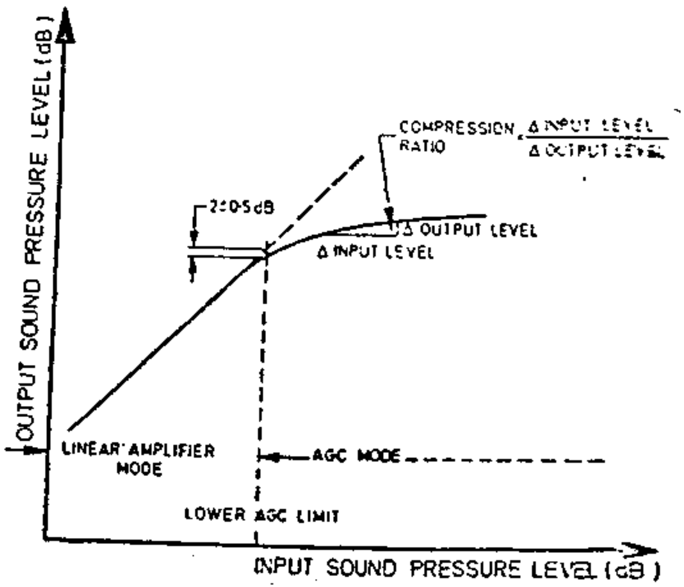

Figure 6 : Example of steady state input/output graph

Dynamic output characteristics - The output sound pressure envelope shown as a function of time when an input sound signal of a predetermined frequency and level is modulated by a square envelope pulse with a predetermined pulse amplitude.

Attack time - The time interval between the moment when the input signal level is increased abruptly by a stated number of decibels and the moment when the output sound pressure from the hearing aid with the AGC circuit stabilizes at the elevated steady state level within ±2 dB.

Attack time for the Normal Dynamic Range of Speech - The attack time, when the initial input sound pressure level is 55dB and the increase in input sound pressure level is 25 dB.

High Level Attack Time - The attack time, when the initial input sound pressure level is 60 dB and the increase input sound pressure level is 40 dB.

Recovery Time - The time interval between the moment when the steady state input signal levels is reduced abruptly to a level a stated number of decibels lower after the AGC amplifier has reached the steady state output under elevated input signal conditions, and the moment when the output sound pressure level from the hearing aid stabilizes again at lower steady state level within ±2dB. (Fig.7)

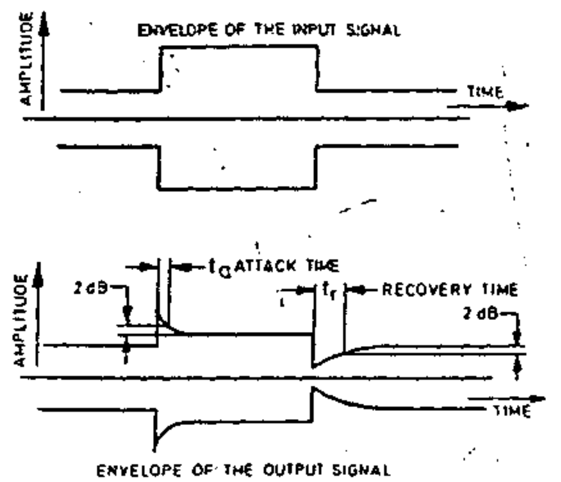

Figure 7 : Dynamic output characteristic of an AGC circuit

Recovery Time for the Normal Dynamic Range of Speech - The recovery time, when the initial input sound pressure level is 80 dB and the decrease in input sound pressure level is 25 dB.

High level Recovery Time - The recovery time, when the initial sound pressure level is 100 dB and the decrease in input sound pressure level is 40 dB.

Conditions of measurements

Although a pure-tone input signal of 1600 Hz or 2500 Hz when appropriate, is specified for various measurements throughout this standard, it is intended that pure-tone signals of other spectral compositions may be used in addition where they would provide important information.

Measurements as in IS 10776 - 1984 (Part 3):

Steady State Input / Output Graph

Graph showing the relation between input sound pressure level and output sound pressure level.

The graph shall have the input sound pressure level as abscissa and the output sound pressure level as ordinate, both expressed in decibels on linear scales having divisions of identical size.

Methods of Measurement - The gain control is adjusted to its maximum setting. Any adjustable gain control after the AGC loop shall be adjusted in such a manner that overload of the hearing aid is avoided.

An input sound signal of frequency 1600 Hz is applied at the lower possible level consistent with an adequate signal-to-noise ratio of preferably more than 10 dB. The input sound pressure level is increased up to 100 dB in sufficiently small steps, and the corresponding output sound pressure level is measured after steady state conditions have been reached. The graph is plotted with the input sound pressure level as abscissa and the output level as ordinate, as described earlier.

Where separate adjustable controls exist, such as AGC, gain or output controls, which will influence the shape and other characteristics of the steady state input / output graph, it is recommended that input / output graphs be plotted, when useful for various additional state setting of such controls.

Dynamic Output Characteristics

The purpose of this test is to determine the dynamic characteristics of the AGC circuit, particularly attack and recovery times. It should be emphasized that all these characteristics will depend on test frequency as well as on such factors as signal level, control settings and battery voltage.

Methods of Measurement

Dynamic output characteristics of speech levels - At the maximum setting of the gain control, an input signal of 1600 or 2500 Hz when appropriate with a sound pressure level of 55 dB is applied. Any adjustable gain control after the AGC loop shall be adjusted in such a manner that overload of the hearing aid is avoided.

This signal is modulated by a square envelope pulse raising the input level by 25 dB. The pulse length shall be at least five times longer than the attack time being measured. If more than a single pulse is applied, the interval between the two pulses shall be at least five times the longest recovery time being measured.

Dynamic Output Characteristics for high level input - At the maximum setting of the gain control an input signal of 600 or 2500 Hz when appropriate with a sound pressure level of 60 dB is applied. Any adjustable gain control after the AGC loop shall be adjusted in such a manner that overload of the hearing aid is avoided. This signal is modulated by a square envelope pulse raising the input level by 40 dB. The pulse length shall be at least five times the attack time observed.

If more than a single pulse is applied, the interval between two pulses should be at least five times the longest recovery time being measured.

Measurement of Amplitude Non-linearities in Hearing Aids - The purpose of this test is to determine the degree of the amplitude nonlinearity in the sound output under specified conditions.

The amplitude non-linearity can be described by the degree of :

a. **Harmonic Distortion -** Distortion products generated by the action of a non-linear transfer function at integer multiples of the test signal frequency. The harmonic distortion products appear at frequencies above the input signal frequency. This means that for hearing aids at higher frequencies they may fall outside the frequency range of the earphone as measured in a coupler.

   For the lower frequency range, the harmonic distortion products give a suitable description of the non-linearity.

b. **Intermodulation Distortion -** Distortion product generated by the action of a non-linear frequency distortion method, are more sensitive to non-linearity in the higher frequency range than are harmonic distortion products.

Harmonic Distortion

Harmonic distortion is measured using an input signal of one sinusoidal tone having the frequency f. The distortion products have the frequency nf, n being an integer. Total harmonic distortion, or harmonic distortion of the nth order, is defined as the ratio of the output sound pressure of the total harmonic distortion products, or at the frequency nf respectively, to the total output sound pressure and can be expressed as a percentage or in decibels.

The total harmonic distortion is given by the formula

$$\sqrt{\frac{P_2^2 + P_3^2 + P_4^2 + \dots\dots\dots\dots}{P_1^2 + P_2^2 + P_3^2 + P_4^2 + \dots\dots\dots\dots}}$$

A

$$\sqrt{\frac{P_n^2}{P_1^2 + P_2^2 + P_3^2 + P_4^2 + \dots\dots\dots\dots}}$$

the formula

Where, P1 is the sound pressure of the fundamental frequency of the signal in the coupler and P_2, P3, P4............P_n are the sound pressure of the harmonic components of the second, third, fourth.........nth order.

Test procedure

a. Adjust the gain control of the hearing aid to the reference test gain position. The position of other controls shall be stated in the report, preferably these should be set to a position that gives the widest bandwidth.

b. Vary the frequency of the sound source over the frequency range 200 Hz to 8000 Hz with an input sound pressure level of 70 dB and analyze the output signal for levels at the harmonic frequencies nf or record the total harmonic distortion content. The bandwidth of the filter should be stated. For continuous recording, the sweep rate shall be such that the response does not differ more than 1 dB from the steady-state value at any frequency.

In the event that the response curve rises 12 dB more between any test frequency and its second harmonic, distortion tests at that frequency may be omitted.

c. If required, repeat the procedure described in item (b) with other input sound pressure levels.

d. Plot the harmonic distortion against the frequency of the sound source versus the input sound pressure level.

Intermodulation distortion - Difference - Frequency Distortion - Difference-frequency distortion is measured using an input signal composed of two sinusoidal signals f1 and f_2, having amplitudes within 1.5 dB, f_2 being higher in frequency than f *i* . The levels of the second order f_2 - f1 and the third order 2 f1 — f_2 distortion products shall be measured and expressed as a percentage or in decibels referred to the output level of f_2 . Higher order components may also be measured.

Test procedure

a. Adjust the gain control of the hearing aid to the reference test gain position. The position of other controls shall be stated in the report.

b. Adjust the frequencies of the test signals f1 and f_2 such that f_2 - f1=250 Hz.

c. Select a suitable number of frequencies f1and f_2 of the sound source or sources within the frequency range 350 to 5000 Hz, maintaining the selected difference frequency and keeping the sound pressure level of each of the two test tones at 64 dB. Measure the sound pressure levels at the frequencies f_2 - f1 and 2f1 - f_2 with a suitable filter. The output level at the filter terminals should decrease by at least 10 dB when the test signal f_2 is switched off. The bandwidth of the filter should be stated.

d. If additional information with respect to input level is deemed to be significant, repeat the procedure described in (c)at other appropriate input levels.

e. Plot the difference - frequency distortion products as two curves or tabulate them for each input level, one for the second order, and one for the third order products as a function of the higher frequency (f_2)

Effect on Amplitude Non-linearities of Variation of Battery or Supply Voltage and Internal Impedance

The purpose of this test is to determine the effect on the amplitude non-linearity of variation of battery or supply voltage variation of internal impedance of the battery or the power supply.

ANSI STANDARDS

AGC measurement:

The 1996 revision of ANSI S 3.22 standard has recommended that the AGC hearing aids be tested using the RTG position as testing the AGC hearing aid at FOG will reveal an artificially high distortion and current drain. However, to ensure that the compression circuits are activated at the reduced gain control position, the input SPL for the attack and release time measurement is changed to 55-90-55 dB SPL (from 55-80-55 dB SPL).

The earlier method for measuring attack time and release time requires that the hearing aid output settle to within 2 dB of the steady state value after a sudden increase or decrease in input level. For AGC hearing aids with variable release times, use of 2 dB settling level can produce considerable variation in results. Hence this has been changed from 3 dB and 4 dB for attack time and release time measurements, respectively.

3 SELECTION OF NONLINEAR HEARING AIDS

Although the objectives of a non-linear fitting are similar to those of a linear fitting, the method used to accomplish these objectives can be much more complicated. Before fitting a compression hearing aid, the reasons for doing so should be clear, so the appropriate type of circuit can be selected and appropriate adjustments made. If an individual has an adequately large dynamic range (> 30 dB) and compression is needed primarily to prevent loudness discomfort, compression limiting is an appropriate choice. The compression circuit should have a relatively high compression threshold, a high compression ratio, and fast time contants, either by design or by external controls.

Linear amplifiers have as their feature the characteristic of adding the same amount of amplification to all levels of input levels until saturation is reached. The low level signals are amplified with the same amount of gain as high level input signals, (Smriga, 1993).

In Fig.8 the effects of placing linear amplification on a recruiting ear are illustrated. A linear aid can only provide adequate gain for a very limited range of input levels. For most loudness range within which the patient must operate, this linear system would either provide too little or too much amplification. The non-linear hearing aid (Fig.8) can provide the desired performance.

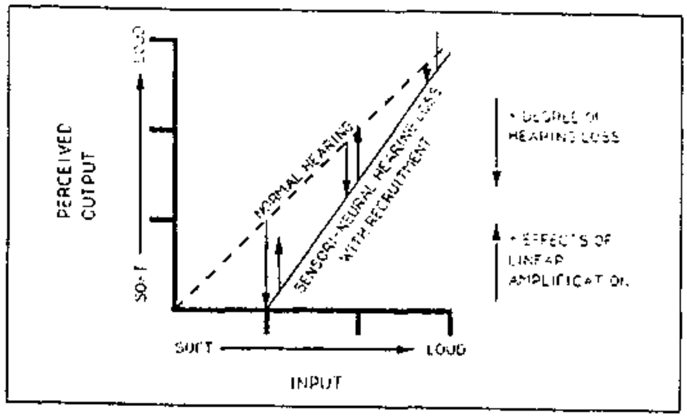

Figure 8 : The effects of linear amplification when applied to a sensorineural hearing loss with recruitment. Note that when middle intensity inputs are amplified to an adequate level, soft input levels are under-amplified and loud inputs are over-amplified.
Source : Smirga, D.J. 1993, Hearing Instruments, Vol. 44, No.ll.

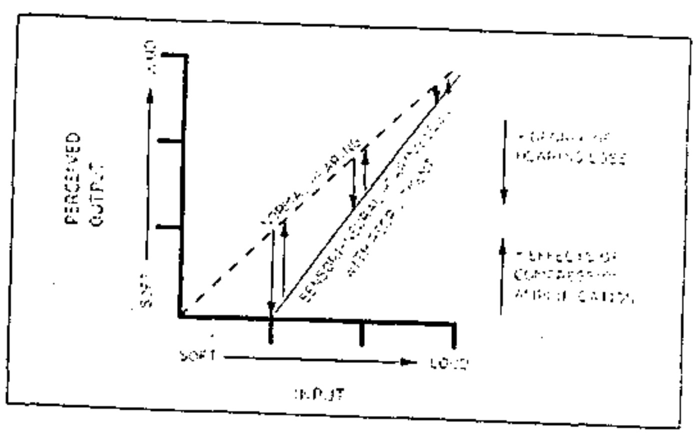

Figure 9 : The effects of compression amplification when applied to a sensorineural hearing loss with recruitment. Note that soft and loud intensity input levels are automatically amplified with a different amount of gain than mid-intensity input levels, thus creating a more normal loudness perception.
Source : Smirga, D.J. 1993, Hearing Instruments, Vol. 44, No.ll.

There are a variety of technologies available to accommodate the requirements of an sensorineural hearing loss. Non-progammable (class-D, AGC-I, AGC-0, Kamp) and programmble (multichannel compression, adjustable (R., interactive time constants) devices, along with a variety of prescriptive procedures for selection of non-linear hearing aids.

If the person to be fit is frequently exposed to changing impulse-type sounds, an AGC instrument occasional might be the circuit of choice. The compression circuit should have a low compression threshold, a fairly high compression ratio, and a long release time. An AGC instrument will not cause appreciable spectral or temporal distortion of speech as long as signal processing is limited to only a few processing bands, and should have acceptable sound quality as long as the compression ratio is not too high. Finally, an individual with a narrow dynamic range (< 30 dB) might benefit most from an instrument with syllabic compression.

When syllabic compression is necessary, however, it should only be used in a hearing aid with no more than three processing bands. The instrument should have a low compression threshold and relatively fast time constants. The compression ratio should be adjusted to provide only as much compression as is needed for the individual, a compression ratio of 3:1 should probably be avoided if a ratio of 2:1 provides satisfactory amplification.

The earlier fitting rules, such as the half-gain rule, POGO, NAL-R offer a single gain target for a hearing loss, because linear hearing aids provide same gain for all input levels. Fitting methods for non-linear hearing aids offer more than one target, because they provide different gain for different input levels. The philosophy, of nonlinear hearing aid fitting lies in equalizing Vs. normalizing the loudness of adjacent speech frequencies.

IHAFF Approach

A method based on loudness of speech, was developed as a known Independent Hearing Aid Fitting Form (IHAFF Mueller, 1994). In this procedure, the gain targets are calculated for all three input levels, (soft, moderate and loud levels) based on measured loudness level. It is necessary to conduct the contour test, the resulting patient specific loudness values are used to calculate three targets, for soft, average and loud speech expressed in 2cc coupler gain. When these targets are displayed, one has to pickup the hearing aid characteristic. This can be made easier by the use of VIOLA. For each set of parameters, a frequency specific input output curve is plotted. Adjustments can then be made to give a best fit.

FIG -6 : In the FIG. 6 software, one has to enter the patient's audiogram and then FIG. 6 will provide with frequency specific insertion gain targets for 3 input levels i.e., 40 dB SPL, 65dBSPL and 95dBSPL. FIG 6. is based on average loudness data. The individual patients loudness judgements are not used in the calculation. FIG 6 is designed for WDRC instruments, has a low compression knee point. Desired compression ratios, based on the predicted dynamic range, are displayed for two input ranges (40dB SPL to 60dB SPL to 95dB SPL) for both a low frequency range (500Hz - lOOOHz) and a high frequency range (2000 Hz-4000Hz). These compression ratios are helpful in instruments with two or three channels. It restores normal loudness for low level, comfort level, and high level sounds. Target gain for 3 input levels (40, 65, 90dB SPL) are computed.

DSL *[Vo]* approach

Most prescriptive methods for adjusting gain and frequency response are not well suited to the fitting of compression hearing aids, because they do not provide a way to evaluate level-dependent signal processing. Cornelisse, Seewald and Jamieson (1994) however, described on alternative prescriptive approach that may be quite appropriate for fitting compression hearing aids. The Desired Sensation Level (DSL) approach that they have developed prescribes the desired output of the hearing aid of each of several input levels. The method generates a desired I/o function that is based on the dynamic range of the listener at a particular frequency. Currently this method is designed to fit full dynamic range compression instruments. This has now evolved into DSL

4.0 version, which provides targets for both linear and non-linear hearing aids (Cornelisse, Seewald and Jamieson, 1994)

DSL (4.0) software - This includes DSL (I/O) algorithm in addition to the old DSL procedure. This DSL I/O algorithm adjusts for either linear or WDRC hearing aids, particularly applicable for instruments with very low compression knee-points. This DSL (4.0) also includes a speech spectrum for use with adults. The DSL (I/O) algorithm provides 2CC coupler gain and output targets and real gain and output targets for puretone input ranging from 45dB SPL to 100 dBSPL. Also provided are targets for aided sound field thresholds and UCL's (Upper limits of comfort) that are predicted. Targets are then adjusted to the compression knee point that has been selected.

In addition to this, the DSL (I/O) algorithm calculates desired compression ratios for nine different frequencies ranging form 250 Hz - 6000 Hz. This calculation of the compression ratio is based on the relationship between the patient's dynamic range and has UCL value. These nine compression ratios are useful when fitting a multiband instrument one can select the ratio for the frequency that falls in the centre of a given band. Verification of the DSL (I/O) targets can be accomplished through ear canal SPL measurements REAR and RESR or insertion gain (REIG's).

The Desired sensation level (DSL) Procedure was specially designed for use with children (Seewald, 1992). The programme produces age related transforms to account for the smaller size of children's ear canals. It also provides both gain and maximum output targets. In the verification phase, a display of the amplified long time average speech spectrum and the real ear saturation response (RESR) is shown. This display allows to evaluate both audibility and head room. The DSL method is fully PC based software system and facilitates the fitting success in several ways. Once the child's threshold and real ear to coupler difference (RECD) values are entered, the program programmes several points of target values for real ear hearing aid performance including.

It was developed for children especially prelingual and difficult - to test. Rationale is to select frequency or gain that makes the long term average spectrum (LTAS) accessible across the widest possible frequency range. The computer software calculates the required gain to place the LTAS at desired sensation level for amplified speech and desired real ear SSPL values.

Goal of DSL is to fit, the normal dynamic range of speech into the residual dynamic range of a hearing impaired. DSL formula is frequencies specific and provides target values for ideal amplified output over a range of input levels. It provides :

1. The desired sensation levels (DSL's) for amplified speech.
2. Target real ear aided response (REAR) and real ear saturation response (RESR) values, and
3. Target aided sound field thresholds.

In addition, once the particular hearing aid type is selected, the programme calculates the desired 2CC coupler gain and SSPL characteristics for the child under consideration and aided results in a format, known an SPL-O-gram. By plotting the variables of aided and unaided SPL-O-grams, one can compare the child's threshold levels to the predicted levels of conversational speech within his unconcluded ear canal.

NAL - NL1 : Stands for National Acoustic Laboratories procedure for non-linear hearing aids (first version). The aim of NAL-NL1 is to provide the gain frequency response that maximizes speech intelligibility while keeping overall loudness at a level no greater than that perceived by a normal hearing person listening to the same sound. The gain frequency response that achieves this varies with input level, and thus the procedure is for nonlinear hearing aids.

The NAL-NL1 prescribes more low cut below l000Hz thus providing less low frequency emphasis of speech. It usually prescribes less gain than other procedures in the region of greatest hearing loss. For steeply sloping looses, NAL-NL1 will therefore prescribe less variation of gain across frequency than the other procedures. It also tends to prescribe lower compression ratios than the other procedures.

The NAL-NL1 prescribes cross over frequencies compression ratios, compression thresholds, low and high level gains but not compressor attack and release times. Thus NAL-NL1 procedure offers an important potential improvement over existing prescriptive procedures by explicitly maximizing intelligibility within the constraints of loudness comfort.

NAL procedures for selecting SSPL of hearing aids

Use either measured LDL values or hearing thresholds SSPL should be low enough to present loudness discomfort but high enough to prevent hearing aid from being excessively saturated by speech. SSPL can be predicted by measuring LDL or by estimating LDL from hearing thresholds. Acceptable range of SSPL for all degrees of hearing loss can be predicted using this procedure.

The Guided Selection Method of Hearing Aid Fitting

This method, introduced Magilen (1991), allows the hearing system specialist (HSS) to assist the client in selecting the best amplification to optimally fulfill the clients needs. The general goal of the guided selection method (GSM) is to optimize the individual's quality of hearing.

Prior to selecting a particular haring aid system, the HSS must determine the possible degree of success of various types of amplification. This can be done using master hearing aid, stock hearing aids and molds. The HSS must determine a variety of gains, frequency responses, output levels, compression characteristics, multi-program features, and aid types, to demonstrate to the client the effectiveness of each at addressing the stated hearing needs of the client. While selection, the client should be introduced to the range of hearing instrumentation available. The benefit and limitations of each should be presented clearly and thoroughly. While fitting, the client is adjusted for comfort, quality of sound, maximising their hearing threshold level, noisy environment, distance hearing tests. Verification of audiometric benefit is detemined by sound field audiometry and by speech discrimination in noise tests. A trial period is given and then the statement of confirmation of benefit is obtained from the client.

Hearing-aid Preselection through Artificial Neural Network

Artificial neural network is a powerful tool in comparing to recognize patterns for hearing aid selection (Arnsten, Koren and Storm, 1996). The neural network has to be trained to recognize specific patterns. This programming techniques thus assist the hearing aid fitter in fitting of hearing aids. These networks are capable of recognizing audiograms and relating them to suitable hearing aids. But in order to train the network we will require at least two audiograms for each aid are required.

Loudness Growth in Octave Band (LGOB) Strategy

This was the first commercially available clinical loudness growth (LG) procedure developed for the fitting of hearing aids (Ricketts, 1998). It is an easy to use, computer controlled, categorical scaling of loudness procedure. This scheme presents three 1/2 second bursts (separated by *V2* second silences) at intensities randomly distributed across listener's dynamic range. The individual's dynamic range is determined in the initial trial phase of testing using systematically varying stimulus levels. These presentations of upto fifteen equally spaced intensity levels are presented at 500Hz, lOOOHz, 2500Hz and 4000Hz, during the test session. Listeners are asked to indicate the perceived loudness of the stimulus by pushing a button corresponding to one of the seven categories.

LGOB has an advantage of being a portable, self contained unit, which distributes presentation intensities equally across the dynamic range of hearing. However, the distribution of the scaling categories has been questioned. The number of categories is said to be insufficient and adding categories may prove beneficial. LGOB has been modified to incorporate insitu target gain measures.

The Cox Contour Test

The procedure involves presenting the stimuli (preferably warble tones) through a insert earphone. Thresholds must be converted to 2CC SPL values. Using an ascending approach, UCL is obtained. The objective is to define an approach for measuring loudness perception that is reliable, clinically feasible and widely used.

Visual input output locator algorithm (VIOLA)

It is a software assisted method for prescribing amplification targets and selecting a hearing aid to match the targets. The approach is suitable for either linear or non-linear instruments. The goal of the procedure is to restore the loudness of typical speech input levels to those that are experienced by normal hearing persons. Thus selection and fitting of hearing aids are based on their puretone input/output functions in a coupler. It is assumed that a hearing aid that matches the coupler prescription targets will produce specific amplified speech levels in the patients ear canal.

Categorical loudness scaling (Scal Adapt)

This provides a considerable number of fitting parameters to be digitally programmed. Two cross over frequencies, three channel gains, three AGC onset levels. This method can be used for 3-channel AGC and dynamic range compression hearing aids. Here the hearing aid user is placed in front of a loudspeaker connected to an audiometer. The opposite ear is blocked to ensure monaural stimulation. The listener scales the loudness perceive in eleven loudness categories (Not audible /O, very soft /5, intermediate /10, soft /15 intermediate /20, comfortable /25, intermediate/ 30, loud /35, intermediate /4, very loud / 45, too loud /50 are used).

Fitting WDRC through loudness mapping (restores normal), loudness relationships, among environmental sounds. There the hearing instrument is adjusted such that the normal dynamic range of sounds is transformed into the narrower dynamic range of the hearing impaired listener. This is accomplished without altering loudness relationships between sounds. However, since individual loudness perception for hearing impaired cannot be well predicted from corresponding individual audiometric threshold data, additional measures - categorical loudness estimation, in fitting was developed.

Fitting multichannel fast acting compression

Rationale is to place as much of speech spectrum as possible above absolute threshold for a given overall loudness. The Cambridge formula (Moore, Alcantara and Glasberg, 1998). The frequency selective insertion gain (IG) results in a near normal overall loudness (about 23 sones for binaural listening) for speech with an overall level of 65dB SPL. It should also result in a specific loudness pattern that is flat (constant specific loudness or loudness per critical band) over the frequency range that is most important for speech, i.e., 500-4000 Hz.

IG = HL X 0.48 + INT, where HL is absolute threshold in dBHL, INT is a frequency-dependent intercept. The INT for each audiometric frequency is given in a table below. Above 5 KHz gains are limited to the value at 5 kHz. Below 500 Hz, gains are reduced to achieve specific pattern to reduce masking of speech by environmental sounds intercept (INT) in Cambridge formula.

Table : INT values for different frequencies

Frequency	0.125	0.25	0.500	0.75	1.0	1.5	2.0	3.0	4.0	5.0
INT.	-11	-10	-8	-6	0	-1	1	-1	0	1

The VIOLA component of the IHAFF protocol allows to select adjustments that best match the patients loudness -growth function. The DSL I/O provides with different ear canal SPL targets for different compression settings. The Fig. 6 method provides with three different gain targets based on average loudness -growth functions, predicted from the patient's own thresholds. NAL -NLI is based on equating the loudness of adjacent speech frequencies. Scal Adapt allows adaptive and interactive hearing aid fitting monitored by category loudness scaling in sound field condition. Sound field audiometry should be used to see if soft speech is made audible.

Audiologist must consider the potential for adaptation to loudness and/or sound quality Initial adjustments may thus have to be modified, especially in high frequencies to increase speech intelligibility (Lindlay 1999). Softwares are available that guide the selection of crossover frequencies, CR, time constants, knee-points, etc.

Fitting of output limiting hearing aids involve two stages

1. Initial fitting based on audiometric and/ or psychoacoustic data,
2. Fine tuning to suit the individual. These two aspects of fitting procedure define the gains, compression ratios and compression thresholds for non linear device of for each channel of a multi-challal compression system .

To exploit the assessment of auditory and hearing aid characteristics using measured in the individuals ear canal, audiologists need, suitable audiological equipment and complex measurement systems.

Loudness growth measures are frequently used in fitting where fairly steep loudness growth dictates less gain as a function of increasing output levels than a fairly shallow function.

The audiogram based method is inappropriate for programmable non linear hearing aids (Kiessling, Scherbert, Archut 1995). Adaptive fitting strategies are required for advanced analog and digital hearing aids. The idea is to normalize the aided loudness perception for 2 input levels, one around the most comfortable level (MCL) and the other just below the uncomfortable level (UCL).

Fitting of multichannel compression devices is more complex than single channel fittings.

SUMMARY

The purpose of this project was to review selected journal articles and books on output limiting /non-linearity in hearing aids. From the vast literature that is available it is interesting to know how the normal hearing takes place, what happens when hearing is impaired in terms of dynamic range of hearing and what technologies are incorporated in hearing instrument to overcome this problem.

The psychoacoustic and neurophysiologic information reviewed suggests that in sensorineural hearing loss the auditory processing changes from normal processing. SNHL (Cochlear hearing loss) is usually accompanied by abnormal rapid growth of loudness, i.e., there is hearing loss for soft sounds whereas hearing loss is not evident for loud sounds. Also, the reduction in useable cochlear hair cell density predisposes the impaired cochlea to perform poorly, especially in the presence of noise, performance is degraded more as the efferent neural mechanism, that assist cochlea in perceiving speech in noise, is impaired, (Smriga 1993).

Use of amplification is the choice of treatment in SNHL patients. Many amplification devices operate essentially as linear amplifiers over much of the their operating range, i.e; gain is independent of the level of the input signal. Very soon it was apparent that linear amplification was not practical for loss of audibility caused by cochlear damage, due to loudness recruitment Most hearing aids incorporate a way of limiting the maximum output so as to avoid discomfort to the user. This is achieved by peak clipping in the output stage of the hearing aid, this introduces unpleasant distortion (Crain and Van Tasell, 1994 ; Cited in Moore, 1996). Even with this it is impractical to compensate fully for the loss of audibility. It was suggested many years ago that problems associated with reduced dynamic range could be alleviated by the use of automatic gain control (AGC) (Steinberg and Gardener, 1937 ; Cited in Moore, 1996).

With AGC or compression, 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 a narrower dynamic range of the hearing impaired individual. There are many ways of implementing AGC, and there is till no consensus as to the "best" method, if there is such a thing, The different types are developed on the basis of different rationales or design goals. Some of them are :

1. Compression limiters (C.R. is large, compression threshold is high, small AT; of < 5 msec, and fairly small recovery time; 20-100 m sec.).

2. Whole range syllabic compressors (CR is lower, compression threshold is lower, short time constants ; 20-100 m sec).
3. Wide dynamic range compressors.
4. Slow acting compression, etc.

These compression can be either in input stage or output stage of the amplifier and can be single channel or multichanneled. The technology used in such hearing aids can be either analog, programmable or digital.

AGC has more recently been termed Automatic Signal Boressing (ASP). These reduces the gain at high levels and or increase gain at low levels but do not change the frequency response of the hearing aid in the process. This includes fixed frequency response (FFR) circuits, compression limiters and WDRC. The other type that changes the frequency response inaddition to gain is called level dependent frequency response (LDFR) circuits (Treble increases at low levels or TILL, Bass increases at low levels or BILL, programmable increases low levels or PILL).

There have been conflicting reports on comparison of compression to linear processing and while attempting to optimize compression parameter, because of lack of consensus, on comparing results for subjects with different pathologies and compression parameters.

Several studies, referred to earlier, have shown that compression at least doesn't cause a decrease in intelligibility over a linear system, whereas peak clipping does.

Various models / procedures have been used for fitting non-linear hearing aid. Loudness equalization which places as much of speech spectrum as possible above absolute threshold for a given overall loudness. An alternative approach is to restore loudness perception to 'normal'. Here, the frequency - and level dependent gains for any given sound should be such that the loudness perceived by the hearing impaired listener is the same as would be perceived by a normally hearing listener without amplification.

If multiband systems are considered, the audiologist can presently choose aids that allow a loudness contour normalization fitting strategy, or ones that allow a low frequency noise reduction strategy. The former leads to a greater degree of compression in the high frequencies, the latter to a greater degree in the low frequencies and there is no evidence that either is superior to a single band compression system.

Much evaluation work, preferably using wearable hearing aids with a realistic range of signal and known input levels, is required. Extensive research by audiologists, electronic engineers and auditory hair cell phyoiologists is required to recoup the lost natural hearing ability through artificial means, i.e., through a hearing and we can still hope for better technologies in future and look forward to bridge the gap of one of natures most astounding, gift -the sensation of hearing, catering the needs of those people who have definitely lost it, in varying degrees.

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