TECHNOLOGY TO IMPROVE SIGNAL-TO-NOISE RATIO: A REVIEW

Reg. No. M2K01

An Independent project submitted in part fulfillment for the first year M.Sc (Speech & Hearing) to University of Mysore

ALL INDIA INSTITUTE OF SPEECH & HEARING

MYSORE - 570 006.

MAY - 2001.

Dedicated to....

.. Ma, Daddy & Akshay,

For all your faith in me.

.. Manjula ma'am

For turning my nightmare into a

dream-come-true!

Certificate

This is to certify that this independent project entitled **"TECHNOLOGY TO IMPROVE SIGNAL-TO-NOISE RATIO: A REVIEW"** is a bonafide work in part fulfillment for the degree of Master of Science (Speech and Hearing) of the student (Register No. M2K01).

DIRECTOR

All India Institute of
Speech and Hearing.
Mysore - 570 006.

Mysore,
May, 2001

Certificate

This is to certify that this independent project entitled **"TECHNOLOGY TO IMPROVE SIGNAL-TO-NOISE RATIO: 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.

Guide

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Declaration

This Independent project entitled **"TECHNOLOGY TO IMPROVE SIGNAL-TO-NOISE RATIO: A REVIEW"** is the result of my own study under the guidance of **Mrs. P. Manjula,** Lecturer in Audiology, Department of Audiology, All India Institute of Speech and Hearing, Mysore and not been submitted earlier in any other University for the award of any diploma or degree.

Mysore,

May, 2001 Reg. No. M2K01

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and see this project through.

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INTRODUCTION

Hearing & Speech Understanding

As is common knowledge, hearing is the phenomenon by which sound is perceived by an organism. This perception takes place through a combination of complex operations that serve to transmit acoustic energy from the sound source to the end organ of hearing, taking place throughout the length of the auditory system. The transmission involves attenuation, amplification, transduction of acoustic energy by turns, till it finally reaches the auditory cortex, is perceived, and its significance understood.

Speech perception is a specialized aspect of a general human ability to seek and recognize acoustic patterns. This ability has evolved by taking advantage of, and at the same time being constrained by, the human speech and auditory mechanisms.

A speech signal is first perceived in terms of its frequency and intensity characteristics, then the inherent temporal variations, and finally semantically and pragmatically decoded at the level of the auditory cortex.

Although there is a great deal of redundancy in the perception of speech (acoustic cues, contextual cues, etc.), a loss in hearing sensitivity most perceptibly affects speech understanding and it is with this symptom that individuals often identify their own problem.

Of the various factors affecting speech in intelligibility, the most widely documented factors are Noise and Reverberation. These are factors that show themselves overtly while providing amplification to compensate hearing loss, and result in added difficulties in the understanding of speech. There are various strategies that can be employed to cut down on these unwanted variables, which we will take a closer look at as we go on.

The normal hearing listener, even at a noisy cocktail party, is able to recognize and understand speech. By contrast, the listener with sensorineural learning loss performs quite poorly in such circumstances even when fitted with a hearing aid (Schum, 2000). There are a variety of factors that contribute to the understanding of speech in noise. These factors include Audibility, Squelch, Masking, Temporal resolution, 85 Frequency selectivity.

The understanding of speech in a masked situation depends on the individual's hearing ability. For example, if the noise is low frequency, an individual with normal hearing can search for cues in the high frequency region. However, in a person with sloping hearing loss, the noise masks the low frequencies, and the hearing loss reduces the audibility of high frequency sounds. Hence, speech understanding in noise becomes very poor in these individuals.

Killion and Fikret-Pasa (1993) described three types of hearing loss, the physiological problems involved, and their respective amplification requirements.

> ***Type I:*** 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 OHC function and normal IHC function (Berlin, 1996) since the loss is restricted to low levels, output limiting in hearing aids is superfluous for these patients. What is needed is gain for low level sound in order to make them audible and clear.

> ***Type II:*** A loss of 60 dB is too great to explain solely on the basis of loss of OHC function; it is therefore necessary to assume some loss of IHC function as well. Hence, we have not only a loss of sensitivity for soft sounds, but a loss of some speech cues as well. Thus not only is more gain required for low-level sounds but also to restore even loud sounds to normal loudness.

Type III: When the hearing loss has progressed into the 70-80 dB region, loudness ceases to be a primary concern; the IHC loss (and the resultant loss of normally redundant speech cues) is so great that one concern dominates: Intelligibility. The range of input SPLs, over which speech is intelligible, especially in the presence of noise, is very narrow. Hence this type of loss requires enhancement of signal in the presence of noise, apart from mere amplification.

It has been stated that hearing thresholds upto 50 - 60 dB HC reflect only outer hair cell damage with intact inner hair cells (Kates, 1993), and that even in noisy situations, for persons with OHC loss but no IHC loss, restoring audibility is the main key to hearing improvement. (Van Tassell, 1993). Plomp (1994) cited the extra deficit that persons with hearing loss have in listening to speech in noise that cannot be explained by audibility effects. He stated further that only a 3 - 6 dB loss for speech in noise represents a very serious handicap for hearing-impaired listeners. Killion and Villchur (1993) theorized that making both noise and speech audible should provide some benefit as long as the SNR is positive. Killion (1997) proposed that persons with OHC loss exhibit a loss of sensitivity, while those with IHC loss lose information. This data indicated that persons with even a mild hearing loss require a higher SNR for

50% word recognition than normal hearing persons. Peters, Moore And Baer (1998) found that ten elderly persons with moderate to severe hearing loss required on average about 8 dB and 11 dB higher SNR than ten young persons with normal hearing to obtain comparable performance using a single voice masker and a speech weighted noise modulated by speech, respectively.

Binaural Squelch is another factor that aids speech perception in noise. When a normal listener is given the opportunity to listen binaurally in background noise as compared to monomial listening, the squelch phenomenon is observed, which enhances the signal audibility by 3 dB or more (Carhart, 1965). This squelch effect is based on figure-ground principles, and can be said to be lost to a great extent in cases with bilateral hearing loss (Bronkhurst & Plomp, 1990; Ter-Horst, Byrne and Noble, 1993).

Upward Spread of Masking is the masking effect that takes place outside of the pure physical bandwidth of the masker. Since most maskers are low frequency, they may mask the mid and high frequency signals as well. This phenomenon is reported to be significantly greater in SN hearing losses than in normal ears. (Jerger, Tillman & Peterson, 1960; Rittmanic, 1962, Martin & Pickett, 1970; Gagne, 1988).

SN hearing loss patients, due to pathology of the auditory system, do not have fine temporal discrimination (Fitzgibbons 8B Wightman, 1982; Bacon & Viemeister, 1985) and there is greater likelihood of temporal overlap between the speech signal and the competition, which is called Temporal Smearing.

The ability of the normal hearing listener to respond to selected frequency components of a complex sound is remarkable. Needless to say, this ability is impaired in individuals with hearing impairment. Frequency selectivity is usually assessed using psychophysical turning curves, which are similar to neural turning curves. Inasmuch as patients with sensorineural hearing loss show poor speech discrimination in noise, they also show abnormally broad neural and psychophysical turning curves. Tuning curves are indicative of the maximum level of competing signal at which the target signal is still audible (at each frequency). Broad turning curves indicate that relatively low-frequency tones that are just above threshold can result in a rather widespread pattern of excitation in an ear with sensorineural hearing loss. This spread of excitation has important implication in hearing aid fitting. Amplification merely attempts to overcome the loss in sensitivity; it fails to restore the ear's frequency selectivity. What are needed in such cases are strategies to improve the speech to noise ratio.

In response to a sinusoid with a frequency below 5 kHz, nerve firings tend to be *"phase locked"* or synchronized to the stimulating waveform. A given nerve fibre does not necessarily fire on every cycle of the stimulus but, when firings do occur, they occur roughly at the same phase of the waveform each time. Thus, the time intervals between firings are (approximately) equal to or integral multiples of the time period of the stimulating waveform (Moore, 1998).

Although the effect of cochlear damage on phase locking is not entirely clear, it has been proved that for neurons with centre frequencies corresponding to frequencies where behavioural thresholds are elevated by 40 dB or more compared to normal, phase locking was significantly reduced (Wolf, Ryan & Bone, 1981). It is, however, an unquestioned fact that cochlear damage can certainly affect phase locking to complex sounds such as speech. The phase locking to formant frequencies observed in the normal auditory nerve may play an important role in the coding of formant frequencies in the auditory system. If so, the reduced phase locking associated with cochlear damage might contribute to problems in understanding speech.

The auditory system is capable of discriminating the timing of acoustic events, as well as determining their frequency. This timing behaviour, also called Temporal

the masking signal on & off.

Resolution, is a combination of the response of the auditory filters and the firing characteristics of the auditory nerve fibres (Kates, 1987). The nerve fibre has a strong reaction to the onset of a sound, while the end of acoustic stimulation results in a time period where the fibre cannot react at all. The psychophysical correlate to this neural firing behaviour is temporal masking, in which a burst of noise or a tone burst masks a signal presented just before the outset of the burst (backward masking) or just after the end of the burst (forward masking). Glasberg, Moore & Bacon (1987) indicated that the hearing impaired might suffer from abnormal temporal resolution in addition to abnormal frequency selectivity. When a sound stops, the impaired ear has an initial recovery from masking that is slower than the normal ear and will, therefore, have a more difficult time in reacting to the outset of the next sound if it occurs within a short time interval.

The results from experiments in gap detection such as those of Fitzgibbons & Wightman (1982) and Irwin & McAuley (1987) indicate that the hearing impaired have a more difficult time in detecting gaps in a signal. Florentine & Buus (1984) pointed out that some of this difficulty, however, may be due to the impaired threshold, since the hearing impaired also will have more difficulty in detecting high frequency transients associated with gating the masking signal on & off.

Extrapolating to speech, impaired temporal resolution means that there will be more difficulty in detecting pauses in the speech signal and in identifying the beginnings of the speech sounds that follow a pause. In addition, the temporal masking effects of reverberation and background noise may persist longer in the impaired ear than in normals, which would exacerbate the difficulties in understanding speech when such interference is present. Noise degrades speech by 1) Masking within critical band 2) Spread of masking between critical bands and 3) forward or backward masking in time.

Noise is often defined as being "composed of several frequencies that involve random changes in frequency or amplitude". A broader, more apt definition would be "Any sound that is undesired or interferes with one's hearing of something" (Schum, 1992). There are many classes of "noise" per se, but the most difficult noise to control is that of other people talking.

There are several general principles that govern the differences between the effects of various types of noises, as is much discussed in literature.

a) The greater the similarity between the spectrum of the masker and the speech signal, the more effective will be the masking noise in disrupting the understanding of the speech material. (Schum, 1992).

b) If speech understanding is compared for a constant versus amplitude modulated background noise, the performance will be better in the presence of modulated noise (Carhart, Tillman & Greetis, 1969). If, however, the modulations mimic the speech envelope, then it can be more disruptive than steady-state speech noise.

c) Linguistic similarity between masker and speech signal: Although literature advances equivocal findings, it is generally suggested that maskers with linguistically significant and identifiable material pose a greater challenge to speech understanding.

In a room, speech sound surrounds a talker. Part of the sound directly reaches the listener, and the rest strikes surrounding boundaries and forms reflections that reach the listener's ears some milliseconds after the direct sound. These reflections constitute **Reverberations.** As a measure of reverberation, **Reverberation time** (RT) was proposed by Sabine (1927), who defined RT, in seconds, as the time during which a sound level decreases by 60 dB from the level when the sound is stopped. According to

Nabelek & Pickett (1974), speech perception declined with an increase in RT.

The main concern of this project is the effect of noise on speech perception, and the various ways to overcome them with specific references to hearing impairment.

The levels of noises in various environmental settings were studied (Pearsons, Bennett & Fidell, 1977) in an attempt to assess acoustic characteristics of non-laboratory communication environments. The various settings were homes, schools, transportation vehicles and other public places. It was found that for a relatively low background noise, talkers produced about 55 dB (A). As the background noise levels increased from 48 - 70 dB (A), typical talkers raised their voices at the rate of 0.6 dB for each 1 dB increase in the background noise level. Therefore, at 70 dB (A) of background noise, the speaker produced around 67 dB (A). The above study also inferred that the overall noise levels vary with the type of environment. The noise levels as surveyed by Pearsons et al., (1977) in increasing order were homes, schools, patient rooms, nursing stations, Departmental stores, trains, aircrafts. A survey done by Schum (1991) comes to the overwhelming conclusion that persons with a sensorineural hearing loss are at a distinct disadvantage in situations with background noise.

A very relevant acoustic characteristic of noise is its overall level in dB SPL in a listening environment. The absolute noise level is important, but of more essence is the level of noise in comparison to the level of the speech / signal that the listener is attending to i.e., the signal - to - noise ratio (SNR, S/N Ratio). The higher the levels of background noise, the lower the S/N ratio, the greater the difficulty to perceive the signal.

Moore, Glasberg & Vickers (1995) and Eisenberg, Dirks & Bell, (1995) showed that SNR deficit can be much larger when the background noise is a single competing talker or amplitude modulated noise as compared to speech shaped continuous noise. Peters, Moore and Baer (1998) seemed to concur with the above findings when they pointed out that the SNR deficit for persons with hearing loss depends on the nature of background noise.

Most individuals with hearing impairment find it very difficult to listen in noisy environments, which may be attributed to a number of factors.

a) Most individuals with hearing loss have better hearing in low frequencies.

b) Noise is usually concentrated in low frequency regions.

c) In certain speech sounds like vowels, for instance, there is more energy in the low frequency regions.

d) It can also be said that low frequencies contribute maximally to the power of the speech signal and minimally to the clarity, whereas with high frequencies, it is vice versa.

Physiologically, the best approach would be normalization strategy, whereby it is ensured that the signal reaching the auditory nerve is adjusted such that it makes up for all the structural & sensory deprivation in its path. But this is not practically feasible.

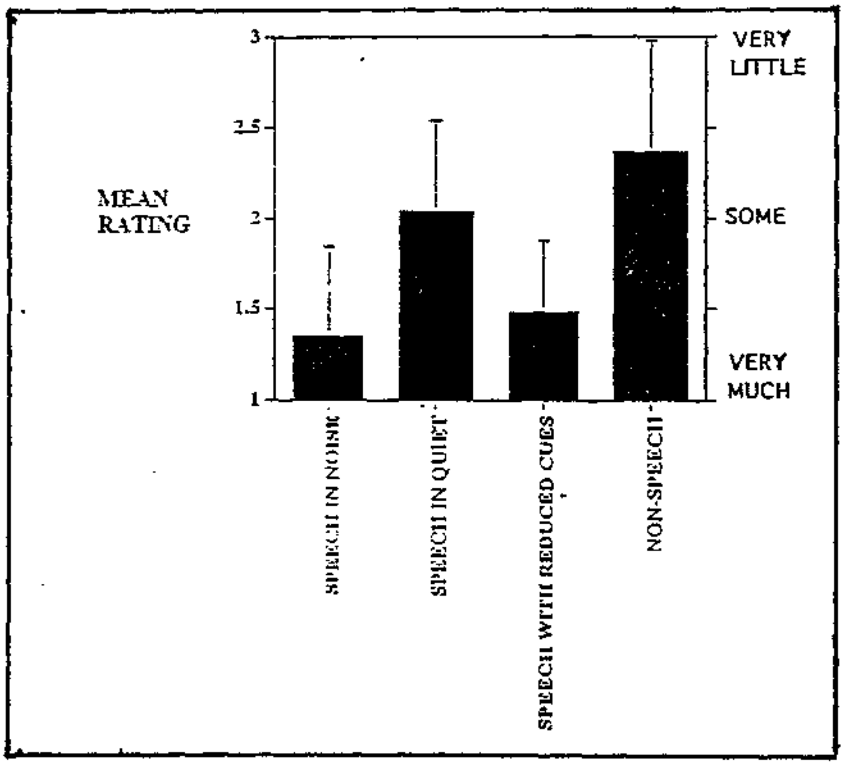
Fortunately, there are hearing aids available and under development, that can solve the problem of hearing in noise, even for those with profound hearing losses. The new developments directly attract the problem with what might be called a "head-through-the-wall" approach (Killion, 1997). There is nothing intrinsically new about these solutions which can improve the SNR by 5, 10 or even 20 dB. What is new is their practicality with today's electronic technology.

One of the severe limitations to hearing aid utilization is the large number of hearing aids that do not work well in noise. However, as the proportion of hearing aid wearers who do well in noise increases, we can expect hearing aids to move towards the status of glasses: a nuisance, but a welcome relief from not being able to see well!

Technology has seen a great deal of development towards improving the signal-in-noise problem. Since information in this context is diffuse and spread out over many years, this is an attempt to compile, as far as possible, a hands-on review of the available literature pertaining to the improvement of SNR.

REVIEW OF LITERATURE

It is almost a cliche, in this day and age of audiological nascence, to say that individuals with sensorineural hearing loss have difficulty in understanding speech in noise, even with the best-fit amplification.



<u>FIGURE 1</u>: Severity of listening difficulty. (Adapted from Preves, 2000)

The ability of a normal individual to extract meaningful speech information from a background of noise is dependent on a complex interaction of peripheral and central processes (Schum, 1992). Although many researchers have indicated that "there is hope" for the problem of speech in noise for sensorineural hearing loss, there remains a significant shortfall in actual versus desired performance with the use of hearing aids in such difficult listening environments, even if we correctly say that we have come farther in technology than we had ever dreamed possible.

Due to the incomplete understanding of this problem, and due to some technical limitations in what amplification can currently provide, this issue of speech understanding in noise will likely continue to be at the forefront of auditory research and clinical science for some time to come.

However, given the severity of listening difficulties that these patients exhibit, it is of essence to provide a detailed assessment of what we have discovered and learnt so far about the problems of understanding speech in noise, as well as the state of the art of various technological options that attempt to mitigate these difficulties.

PHYSIOLOGICAL BASIS FOR AMPLIFICATION

Something about the nature of hearing loss detrimentally affects a person's ability to understand speech in noise beyond simple loss of audibility. All of the techniques for improving intelligibility of speech so far, however, have been only indirectly related to the primary cause for the speech understanding deficit: the patient's hearing loss. The following statement bears stating simply to emphasize what is obvious but often overlooked: "Speech understanding would be restored to normal if auditory perception were restored to normal".

Since sensorineural hearing loss detrimentally affects speech intelligibility in noise, one strategy for improving intelligibility is to process sound such that the listener's cochlear damage is counteracted. While the feasibility of achieving this goal remains unclear, it is a goal that can drive many strategies for improving hearing and at the same time improving speech understanding.

The processing performed on acoustic stimuli by the auditory periphery, from the pinna to the auditory nerve, is a well-defined process. If any part of the auditory peripheral system does not function properly, then the signal transmitted to the brain from the auditory nerve will be distorted. The majority of sensorineural losses are caused by outer hair cell damage, so the acoustic signal received by the cochlea is not properly transmitted to the auditory nerve, resulting in the auditory nerve transmitting a distorted signal to the brain. The objectives of a hearing aid should be to process the signal such that the signal received by the auditory nerve, and thus the brain, is restored to normal as much as possible.

If inner hair cell damage exists, however, this normalization strategy must be revised, since the communication link to the auditory nerve fibres that innervate the affected hair cells is severed. To employ a normalization strategy, it must first be found out in what way is the auditory nerve signal distorted (Edwards, 2000).

TECHNOLOGIES THAT WORK TOWARDS IMPROVING S/N RATIO

Up until the recent past, most hearing aids had linear circuitries that were easily driven into saturation by even moderate levels of low frequency dominated environmental noise. Killion (1997) stated that low distortion hearing aids with good frequency response and compression characteristics could solve the loss of sensitivity problem and function well in noise.

Essentially, speech intelligibility can be improved by reducing the amplification of noise, by enhancing the speech signal, or a combination of both. Many methods for reducing detrimental effects of environmental noise have been attempted in hearing aids. Most of these efforts can be categorized in terms of spectral and temporal processing, using one or more microphones. Some of the methods employed reduce the desired signal as well as the noise, leaving substantially the same SNR after processing. Nevertheless, although the SNR is not improved, hearing aid wearers may benefit from the reduction of annoying noise, and possibly by reduced upward spread of masking (Preves, 2000).

As Killion (1997) has pointed out, hearing aids incorporating technology from over a decade ago were more a hindrance than help in noisy situations. Narrow bandwidths, peak clipping, distortion and peaky responses are many of the reasons that hearing aid users did better by removing their hearing aid in adverse listening environments. Additionally, hearing aid noise can be generated inherently, depending on the type of amplifier it uses.

Hearing aid technology has advanced considerably since then, however, and there are now several solutions to improving the speech-in-noise problem. Directional processing is one obvious technique for improving speech intelligibility in noisy environments. Also, the recent introduction of digital signal processing (DSP) into hearing aids allows more sophisticated techniques to be implemented than could be done with analog technology. With the improvement in electronic circuitry, the effects of hearing aid noise can also be reduced.

The following section reviews some techniques that can improve speech understanding in noise. These techniques are not necessarily exclusive; i.e., the techniques may be combined to provide more benefit than each on its own does. Some of these "techniques" have not been developed sufficiently to be used commercially, but they also form landmarks in the technological development towards speech enhancement.

Binaural amplification

A normally functioning binaural system affords the listener a significant advantage when listening in background noise, due to binaural squelch. Although the selection and fitting of binaural hearing aids does not guarantee a complete recapturing of normal binaural squelch, there is evidence to suggest that listeners fit binaurally perform somewhat better in noise than when fit monaurally. This evidence is drawn both from subjective patient reports. (Brooks and Bulmer, 1981; Erdman & Sedge, 1981) and from the results of clinical studies on the advantage of binaural amplification when listening to speech in the background of noise (Byrne, 1980).

It should be noted, however, that even with a well-fitted set of hearing aids, some hearing impaired listeners do not perform as well as normal hearing listeners in understanding speech in noise (Hawkins & Yacullo, 1984). As indicated earlier, there are issues other than audibility that are likely to reduce speech understanding in noise. Some investigators, though, (Festen & Plomp, 1986), have found limits on the benefits of binaural amplification for understanding speech.

CROS Hearing aids

Contra lateral Routing of Offside Signals refers to the amplification systems in which signals picked up on one side of the head are routed to the ear on the other side. Some variations of CROS, however, provide binaural amplification and hence help re-establish the binaural squelch phenomenon, which in turn enhances understanding of speech in noise. These varieties of CROS are described as follows:

CRIS - CROS:

This is a double CROS arrangement in which the user can take advantage of the head shadow to obtain maximum gain in each ear and slice retain the two - ear differences.

BIFROS:

Binaural frontal routing of signals utilizing two microphones, one on each side of the frame, each of which sends signals to a receiver in the temple on the same side. In essence, it is a true binaural system, but has the advantage of greater gain capability, directional characteristics and open molds if necessary.

BIFROS-270:

It is a FROS variation. It uses four microphones and two receivers. Two microphones are located in the frame, and one at each temple. The two microphones on each side deliver signals to the receiver on the same side. It provides 270° sound stimulation (a more full-rounded sound).

FM (frequency modulation) Systems

The simplistic approach of increasing the signal-to-noise ratio (SNR) by placing the microphone in close proximity to the signal source and thus decreasing the distance between speaker and listener, is highly effective. In the most basic implementation, the microphone is connected to the hearing aid with a cable. Such a system has been in use for years in the classroom using FM (frequency modulation) technology.

FM systems provide a wireless means of transmitting the sound from the source to the listener. The auditory signal is picked up by a microphone and is transmitted in the form of radio frequency-modulated carrier waves to a personal receiver that is worn by the hearing impaired listener. Each FM system consists of a transmitter with a specific radio carrier-frequency, an antenna, and a compatible receiver.

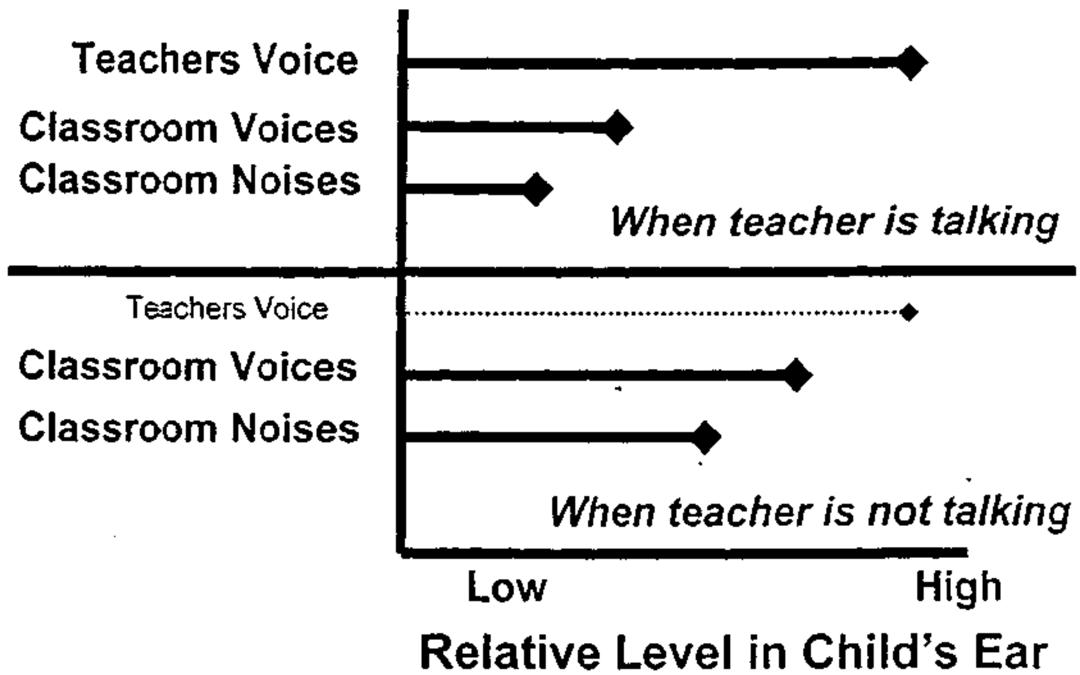
There are essentially two types of FM systems. The first of these is a complete system consisting of: (a) A FM microphone located on the transmitter with associated antenna, (b) An environmental microphone on the FM receiver, (c) An amplifier sufficiently powerful to allow the receiver to function as hearing aid. The second type of FM systems, often referred to as personal FM systems, involves the coupling of the FM system to the client's personal hearing aid. The FM system in this case functions as an assistive listening device (Katz, 1994).

FM systems provide a non-intrusive means of improving the SNR. According to Hawkins & Yacullo (1984), the use of a FM system can improve the SNR by as much as 15-20 dB. For use in noisy situations, the transmitter might include directional or multiple microphones. The applications of new FM systems are endless - the following guidelines maybe kept in mind when fitting a FM system:

a) FM communication systems can greatly improve the SNR, especially when the signal and listener are more than 10 ft away.

b) FM systems are easily accessible and simple to use with new wireless transmitted and receiver.

c) Directional systems may be integrated into a FM transmitter (Mims Voll, 2000).



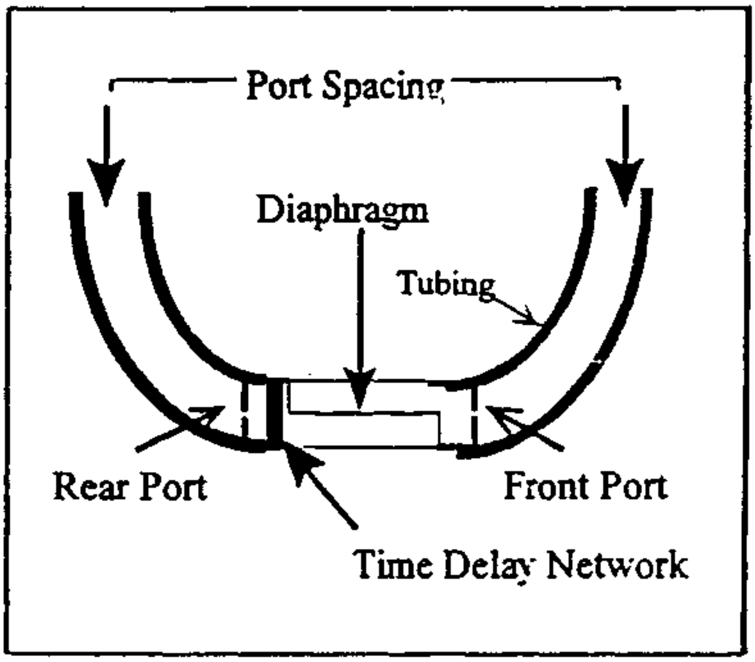
<u>FIGURE 2</u>: Relative levels of teacher's voice. (Adapted from Mims Voll, 2000)

Directional Microphones

One of the most obvious and beneficial techniques for improving the intelligibility of speech in noise is the use of directional processing which preserves sound arriving from the front of the hearing aid wearer, but attenuates sound from beside and behind the wearer (Edwards, 2000). Since the speech that a hearing aid user wants to hear is usually in front of him/her, and interfering noise is frequently all around or behind the user, directional-dependent gain can improve the overall SNR. This is true even if the noise has the same spectral characteristics as the signal and they occur at the same time. Maximum SNR improvement occurs with several microphones that span a distance of several centimeters (Soede, Berkhout & Bilsen, 1993; Hoffman, Trine, Buckley and Van Tassell, 1994). Cosmetic reasons and convenience, however, dictate that microphones be located on the body of the hearing aid, limiting the number of microphones to two, and the distance that they span to less than 15 mm. Given this practical constraint, directionality for hearing aids is implemented with either a single directional microphone or with two omnidirectional microphones placed on the hearing aid.

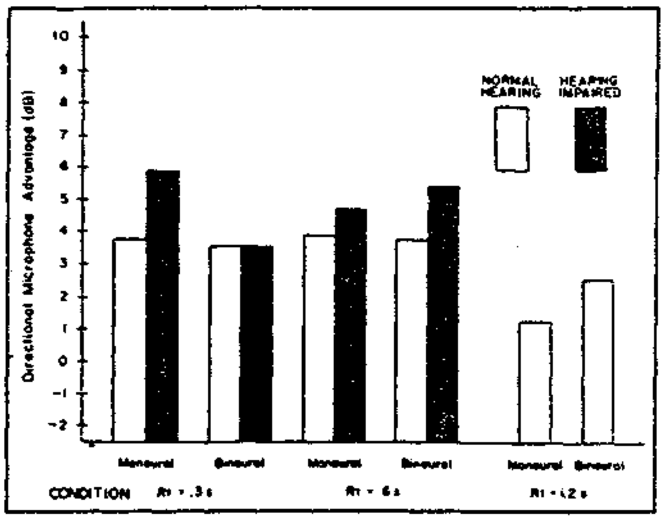
Designing a hearing aid with a single directional microphone:

A directional microphone has two sound ports. The back port has a time delay element that delays the signal slightly. The result is that sounds from the front enter the front port first, then are delayed while entering the back port. This results in an in-phase signal. When sounds come from the back, they first enter the back port, then the front, putting the sound out of phase. Out of phase sounds will cancel, thus reducing the intensity of sound from behind. (Mims Voll, 2000).



<u>FIGURE 3</u>: Schematic of a D-Mic. (Adapted from Edwards, 2000)

The effectiveness of directional microphones has been demonstrated to be on the order of 2.5-4 dB (Hawkins and Yacullo, 1984; Bachler and Vonlanthen, 1995). Although estimates vary, an increase in SNR of 3 dB could result in 40% improvement in intelligibility.



<u>FIGURE 4</u>: D-mic advantage for monaural & binaural listening at 3 RTs. (Adapted from Edwards, 2OOO)

However, it is to be noted that directional microphones are found to be useful only in non-reverberant conditions. In reverberant listening situations, the directionality is nullified, and there is little or no effective difference between directional and omni directional microphones.

Omnidirectional Microphone Array

Another approach to designing a directional system is to use omnidirectional microphones. In typical use, these micro phones are positioned on a hearing aid such that a line connecting the two microphones points to directly in front and directly behind the hearing aid wearer and is horizontal to the ground. The signal picked up by the rear microphone is delayed by an amount of time similar to the time it takes for sound to travel from one microphone to the other. For a microphone separation of 10 mm, that time is approximately 30 msec. The amount of delay determines the directional pattern (gain as a function of sound arrival direction). Varying the delay can vary the direction for which the gain is minimum (Edwards, 2000; Bachler and Vonlanthen, 1995). The use of two microphones basically allows the temporal and phase characteristics to be controlled in the amplifier for a more precise calculation. Valente, Fabry and Potts (1995) found a 7-8 dB improvement in SNR using a hearing aid with a two-microphone array. The effectiveness of one such multi-microphone system has been demonstrated to be 4-8 dB (Lurquin & Rafhay, 1996; Gravel, Fausel, Liskow and Chobot, 1999). In addition to improving SNR with a directional system, these hearing instruments also include a number of preset algorithms to optimize speech perception in a variety of listening situations (Mims Voll, 2000).

Wolf, Hohn, Martin & Powers (1999) demonstrated that hearing aid wearers prefer the directionality not to be 'on' all the time - users desire the ability to switch between directional and omnidirectional modes. One advantage that two omnidirectional microphones provide over a single microphone is that directionality can be switched 'on' and 'off by turning the back microphone 'on' and 'off. Since both patterns may be desirable for different situations, a further improvement to directional hearing aids is to allow the pattern to be selectable for the specific need of the user. This can be implemented by either allowing the pattern to be programmed when the hearing aid is fit, or by allowing the wearers to switch from one to the other.

Some instruments allow both possibilities by implementing the delay digitally. This allows different directional patterns to be programmed such that the wearer for the given noisy environment can select the optimal directional pattern, and thus optimal SNR improvement.

One difficulty with implementing two-microphone directionality is that the microphones must be very closely matched at all frequencies of interest in order to provide good SNR improvement at these frequencies. If the frequency responses of the two microphones are not identical, then directionality is degraded (Thompson, 1999).

Hence, Directionality can be said to be a strategy of choice, being either an analog or a digital processor depending on the use and requirements. It combines the most elementary concepts with state-of-the-art technology.

Higher-order Directional Microphone Systems

The addition of more microphones opens up possibilities for improved noise reduction (Preves, 2000). With a single microphone, attempts at filtering out noise are often based on hearing the characteristics of the noise during pauses in the speech signal. The effectiveness of their technique is limited to stationary or quasi-stationary noise. Multi microphone arrays produce higher order directionality or *beamforming,* in which the direction(s) of attenuation are moved adaptively (either by the wearer's head or automatically by an adaptive filter) in response to the movement in the direction of the desired signal and undesired noise(s) (Preves, 2000).

Beamformers have the ability to emphasize desired signals while filtering or nulling out one or more noise sources. Some beamforming arrays have nulls (much reduced output) at *fixed* locations on their polar directivity patterns that do not change with time. Other beamformers sense the directions of undesirable noises as they move and automatically place their nulls at the locations of these noises in their polar directivity patterns. These directional airays are called Adaptive Beamformers.

Adapting the location of the nulls is accomplished by using an adaptive filter to "steer" the array.

Higher order directional systems with fixed polar patterns can produce excellent performance when used in hearing aid application because the wearer's head normally turns in the direction of the desired signal, thus making even the fixed beamformer somewhat adaptive. These have the advantage of less hardware size and power consumption since minimal electronics are required. Combining the outputs of two or more directional microphones or the output of more than two omni-directional microphones can constitute higher order directional systems.

Results have been mixed with adaptive beamformers. The main problem has been that when the environment becomes too reverberant, performance with adaptive beamformers degrades, because the desired signal may be cancelled. Peterson (1987) showed that the 30-dB increase in SNR provided by a two microphone beamformer in a simulated anechoic chamber is degraded to no improvement by reverberation in a simulated conference room environment. Greenberg and Zurek (1992) achieved similar results. The discouraging results in reverberant environments with adaptive beamformers have led investigations to use fixed beamformers in high levels of reverberation.

Kates and Weiss (1996) evaluated the different systems in an office and a conference room with a male talker located at 0°, and multi-talker babble incident from five azimuths. They found that the number of microphones is not an important factor at low frequencies, but is important at high frequencies. They recommended using an adaptive array in low input SNR conditions, and converting to a super directive array for high input SNR and highly reverberant conditions, in acknowledgement of the deterioration in performance of adaptive beamformers in high levels of reverberation. Other investigators have also combined fixed and adaptive beamformers.

Multi-microphone arrays have been developed to incorporate up to seventeen microphones. However, research has shown that no additional benefit is derived from having more than five microphones. (Soede, Berkhout & Bilsen, 1993). These have been designed as broad side or end fire arrays.

Broad side configuration:

Five microphones are spaced 5.8 cm apart along the forehead of the user. The output from each microphone is weighed and summed.

End fire configuration:

Five microphones are spaced 2 cm apart along the side of the users' head. In this, the output from the second to fitter microphone is time-delayed relative to the first microphone and the output from the five microphones is summed. This is also called Delay and sum configuration.

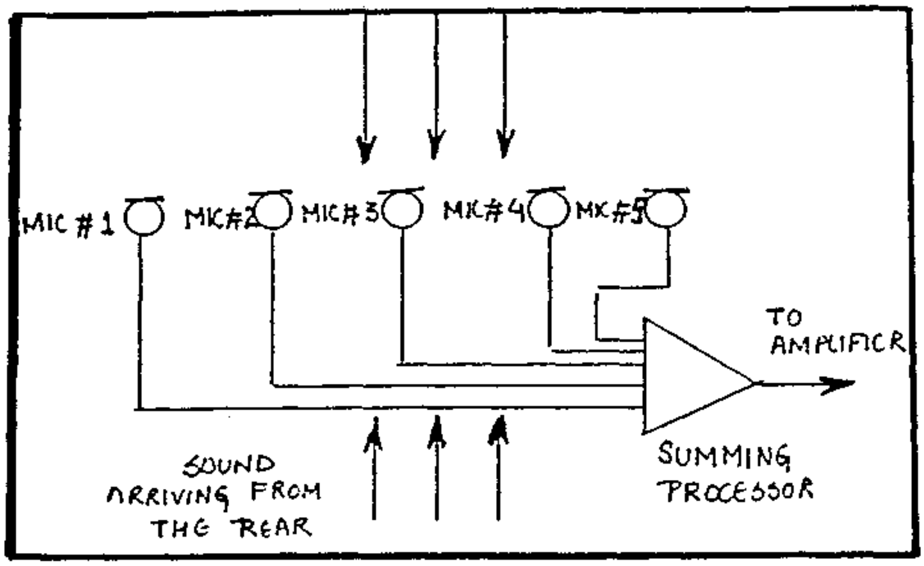


FIGURE 5: Block Diagram of Broadside Array. (Adapted from Agnew, 1997)

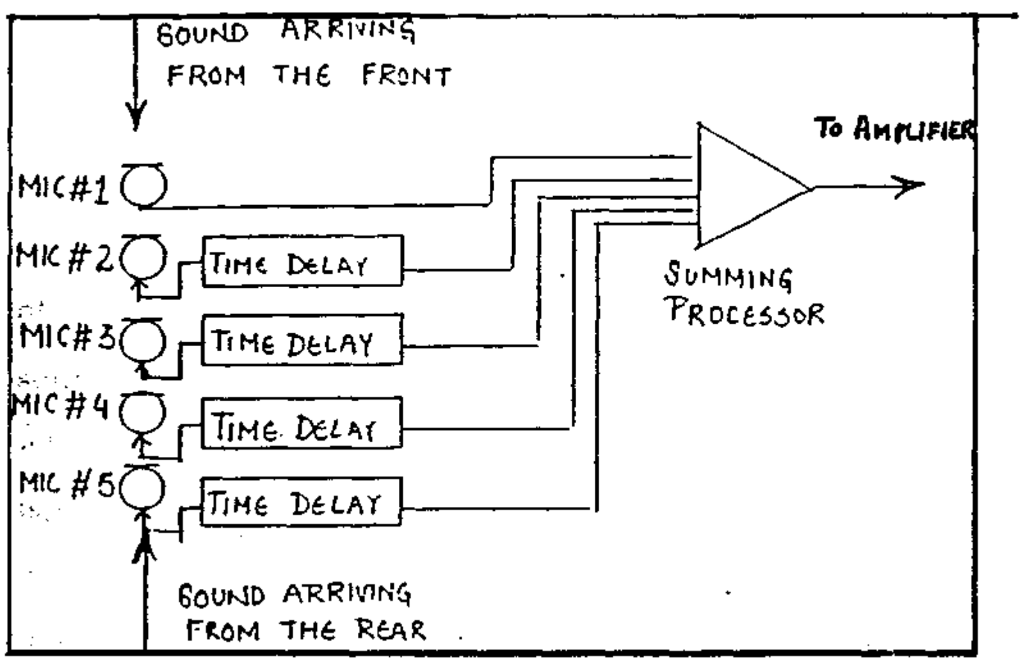


FIGURE 6: Block Diagram of Endfire Array.(Adapted from Agnew, 1997)

Low frequency tone controls

An active low frequency gain reduction via a multiple high pass filter has been in use for many years in hearing aids as a way of alleviating difficulties faced in high level environmental noise. This feature is available to the wearers in the form of a switch / knob that adjusts the amount of low frequency gain reduction. This also leads to a reduction of speech energy in the low frequency region and hence does not effectively improve S/N ratio (Fabry & Van Tasell, 1990; Cook, Bacon, Sammeth, 1997).

Narrow Band Noise and Band pass filtering

In the unusual case of the noise being in narrow band, reducing the band of frequencies occupied by the noise should be beneficial (Preves, 2000). Rankovic (1998) used selective attenuation of low, mid, and high frequency active bands containing high intensity noise and found a release from upward spread of masking that was associated with increases in consonant recognition scores by listeners with sensorineural hearing loss. These relatively favorable data differed from the mixed results of Van Dijkhuizen, Festen & Plomp (1991), who used a similar methodology but with a broad band noise added to active bands of noise. Both the above studies, and earlier studies by Rankovic, Fryman and Zurek (1992) and Horwitz, Turner & Fabry (1991) provide a rationale for the use of a multi-channel compressor having separate automatic gain control circuits for each channel.

Szozda (1987) pointed out that while revolutionary solutions might be some time away, evolutionary progress in extraction of the signal in the presence of noise has been steady, due mostly to filtering.

Active and Passive filters

Over the years, initially the move was made from passive to active filtering.

Passive filters are used for low and high cut filtering (resistor- capacitor networks) They have fixed characteristics (gain, slope, etc.), require a large number of components, and are thus bulky. These are ideal in the dynamic noise environment generally encountered by the wearer.

Active filters usually comprise transistor elements. An ideal active filter is an integrated circuit containing any number of transistors along with other elements. An active filter network can provide several times greater effectiveness than that of a passive filter network. For example, when passive control are used for low frequency modification, can achieve an attenuation of 3.5- 4.5 dB / octave. Active filter controls, on the other hand, can achieve an effective attenuation of 9-10 dB / octave.

As a word of caution, however, Preves (1978) noted that simply filtering out the energy in the low frequency range to eliminate noise also removes some of the low frequency components of speech.

Reduction in Low Frequency Amplification

The second strategy available in selecting is to set tone controls to emphasize a high frequency region. The assumptions behind providing a high-pass frequency response to provide better understanding of speech are that:

a) A significant amount of environmental noise is centered in the low-frequency region

b) Many patients with sensorineural hearing loss show an excess effect of upward spread of masking (as mentioned earlier). By reducing low-frequency amplification as compared to a broader frequency response, there is an increase in the likelihood that the more intense low frequency environmental sounds will not be amplified and the effects of the upward spread of masking will be reduced.

However, if low frequency amplification is reduced, it is often observed that the hearing aid users complain of reductions in the quality of speech. This technique, therefore, creates some problems in an attempt to solve some others.

Fitting high-frequency emphasis amplification to minimize the effects of noise has been criticized by pointing out that the SNR in any given frequency has not been modified using this approach. Further, it has been argued that significant speech information is present in the lower frequencies, which should not be attenuated or eliminated (Schum, 1992). Whereas these criticisms are accurate, there appears to be benefit from reducing low frequency amplification for at least some patients (Gordon-Salant, 1984).

Spectral subtraction

This technique is applicable particularly to speech mixed with noise that does not vary much in amplitude (quasi - stationary noise). An estimate of the interfering noise is made during the silent periods in the speech signal, and this noise estimate is subtracted from the noisy speech signal, leaving, hopefully, the speech signal itself (Boll, 1979). The method is particularly suited to environments with negative SNR, but has not been considered practical because it frequently produces musical tone artifacts.

Parametric estimation and Speech re-synthesis

A technique that has received some attention (O'Shaughnessy, 1989). A method proposed by Kates (1994) involved tracking the formant peaks and sinusoids (without noise) by reproducing only the formant as they changed in peak frequency and amplitude. Kates (1994) reported that reproducing the 16 most intense peaks would produce a 12 dB higher SNR if the suppressed part of the signal was uncorrelated noise. This method essentially increases the spectral contrast in the speech signal.

Increased Headroom:

This strategy to improve the understanding of speech in noise is to increase the "headroom". Some hearing aids use peak clipping to limit the output of the hearing aid. This clipping of the speech peaks results in the level of the speech to be reduced relative to the overall level of the noise. In other words, peak clipping will result in a poorer SNR as well as providing a more distorted signal. The methods of peak clipping have been used for two reasons

(Schum, 1992): First, a hearing aid can be placed into saturation if the output is adjusted too low when compared to the gain applied to typical input levels. The second reason is that the maximum output of the hearing aid maybe adjusted to a low setting to prevent effects of loudness.

Increasing the headroom by increasing the maximum output can certainly decrease the likelihood of clipping for speech peaks. Hearing aids with Class B or Class D amplifiers can provide the exact headroom needed to avoid peak clipping - induced distortions and decreased signal-to-noise ratios.

There is some evidence to suggest that patients will tolerate higher maximum output levels as long as signal distortion is kept to a minimum (Fortune and Preves, 1992a; 1992b). The use of an amplifier with increased headroom is potentially a useful strategy to improve the understanding of speech in noise if the audiologist is not concerned with loudness discomfort, in the presence of which the gain would be less than optimal to avoid excessively / uncomfortably loud output levels.

Automatic signal processing

This term was first used in the mid- '80s, but was used with variable implications. As Kates (1986) noted, signal processing is a tool, an engineering approach to solving problems. Signal processing is the manipulation of the signal to enhance or extract the information it contains. (Libby and Sweetow, 1987). Conceptually, the simplest form of processing is a linear system, which provides a constant change in gain or frequency response. For example: Normal-High (N-H), or Normal-Low (N-L) adjustment filters, or amplifiers giving constant gain. Signal processing can also be non-linear, in which case the processing transformation depends on the signal characteristics. For example: Automatic Gain Control (AGC) compression circuit. According to Kates (1986), automatic signal processing is a marketing term rich in implications but without a defined technical meaning. In contrast, he stated that Adaptive Signal Processing is an engineering term with constant technical meaning.

An ideal hearing instrument is one that would have flexibility to adapt its performance in the face of changing environments. Historically, hearing instruments have been non-adaptive, that is, their processing function did not change once they were set via the appropriate control. A typical AGC circuit may be non-linear in character, but may not necessarily be adaptive. The property of signal processing circuits to adapt and change their own parameters and operating characteristics in response to changes in the signal characteristics produces what is named adaptive signal processing. Kates (1986) has divided the above 'intelligent' hearing instruments into four categories

a) Adaptive (automatic) signal processors.
b) Adaptive compression signal processors
c) Adaptive signal processors with both suppression and expansion capabilities
d) Wearer operated active filter processors.

a) *Adaptive (automatic) signal processors*

These systems are designed to sample the environmental noise levels and automatically adjust the gain and output of the hearing instrument. In quiet, the circuitry provides for full quality reproduction by enhancing the entire frequency spectrum. Excessive low frequency energy triggers the compression in the low frequencies and reduces the gain automatically, relative to the high frequencies. Thus, the louder the low frequency noise, the more the spectrum is tilted to favor the critical high frequencies needed for word recognition.

b) *Adaptive compression:*

The need for some type of dynamic range reduction system remains unquestioned (Killion, 1979). While peak clipping can achieve the desired results, it does so at the cost of increased distortion and consequent loss of clarity and discomfort. Smriga (1985) stated that adaptive compression is an advanced compression circuit that helps maintain the S/N ratio present at the input by automatically adjusting its compression release time according to the duration of noise in the environment. For short-term noise in an otherwise quiet environment, limiting action occurs and the return to linear operations is rapid. If the noise persists, then the release time becomes progressively longer to a maximum of one second. In effect, the instrument reduces its gain for the duration of the noise, ignoring short-term fluctuations.

It may be beneficial here to include both ASP to improve S/N ratio and some form of advanced compression to help maintain the S/N ratio and eliminate distortion elements in the output.

c) Adaptive signal processors with both suppression and expansion capabilities

The multiple signal processor (MSP) is also used in noise reduction circuitry. As described by Staab

(1987), it is not an AGC or an ASP processor in he conventional sense. The MSP uses digital technology information and applies it to analog circuitry. In function, the MSP acts as a frequency suppressor and as a frequency expander. The circuit continuously samples the environmental signals and automatically adjusts the spectrum. Low frequencies may be suppressed as much as 30 dB at 500 Hz, and at the same time the circuit can provide for an expansion (or increase in gain) of the high frequencies as much as 20 dB at 2000 Hz. Staab (1987) adds that this can result in a maximum spectrum changes of up to 50 dB, with no loss of gain.

d) *Wearer operated environmental control circuitry:*
These circuits are now available with active filtering which allows the wearer to select the amount of low frequency amplification for comfort and clarity by the use of a manual control.

A differentiation of ASP circuits can also be made as given in Katz(1994): Those circuits that reduce gain at high levels and/or increase gain at low levels but 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).

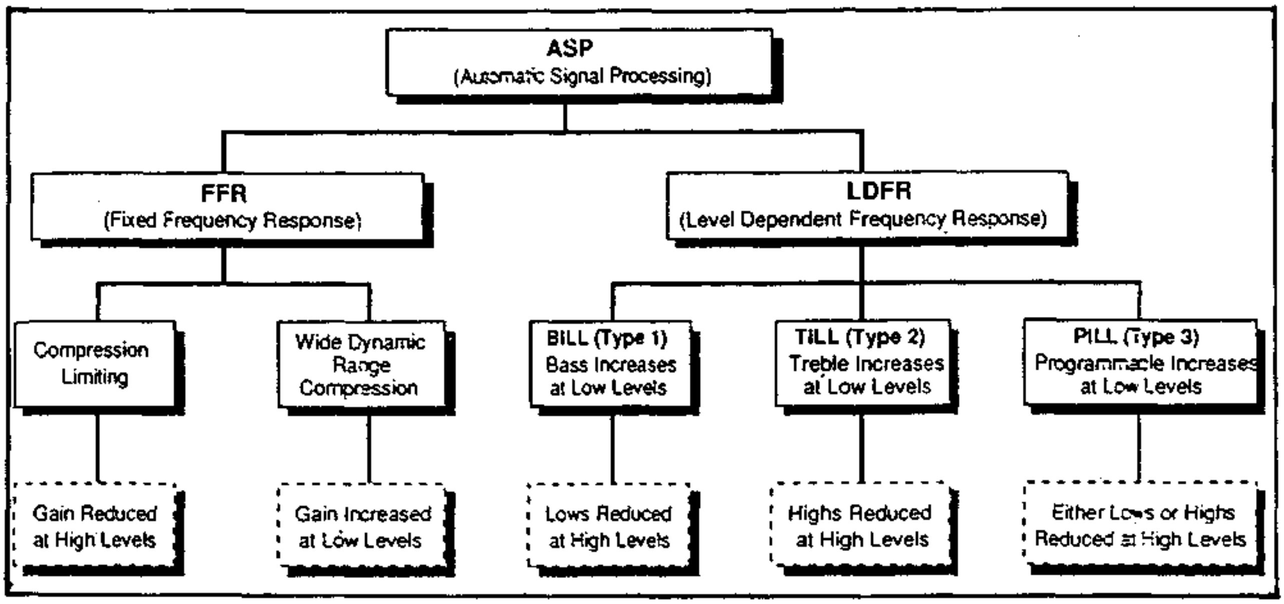


Figure 7: Outline of recommended classification system for ASP hearing aids. (Adapted from Killion, Staab, Preves, 1990)

Well-defined terms for these fixed-frequency response (FFR) automatic signal processing circuits already exist. This includes compression limiting (AGC) and Wide Dynamic Range Compression (WDRC). 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 ASP hearing aids (fig. 7). This classification distinguishes among these circuits in terms of their reaction to low-level rather than high-level inputs.

Wide Dynamic Range Compression (WDRC)

Hearing loss results in elevated thresholds and faster growth of loudness perception. To slow the growth of loudness perception, gain is reduced as input increases. WDRC can be employed in such cases as it provides more gain for inputs below 40-50dB SPL. The main reduction occurs above the point as determined by the comfort limit of the hearing impaired ear. In WDRC, volume control is placed after the feed back loop of the AGC circuit.

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 (Kate, 1994).

Limiting as a result of Instant Output Regulation

These limiting systems include hard and soft peak clipping. Both involve limiting the amplitude at a certain point but in a somewhat different manner.

Hard Peak Clipping is the simplest form of output limiting and can be defined as the removal, by electronic means, of one (asymmetrical) or both (symmetrical) extremes of alternating current amplitude peaks at a predetermined level. The advantages of hard peak clipping are that its construction is very simple and requires very little space to accomplish very effective instantaneous output limiting. Its primary disadvantage is that the SNR may be reduced if the noise levels do not reach the clipping threshold and only the speech peaks are clipped (Schum, 1992)

Peak Rounding/ Soft peak clipping is a form of nonlinear amplification that is evidenced by a gradual, ever-diminishing increase in output with each successive increase in input. The result resembles hard peak clipping in many respects, except that the onset of clipping is gradual.

Limiting by time-dependent gain Regulation: -

Compression/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 (Katz, 1994).

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 does not exceed the UCL of the listener, and to keep the distortion low.

2) To reduce the dynamic range of the output signal so that it is a better match to the dynamic range of an impaired ear.

The gain level is controlled automatically, hence the name automatic gain control (AGC). This action may be further described as the process of 'Compressing' a dynamic range into a lesser dynamic range. When saturation is approached, the feedback circuit causes the gain to drop automatically to a level such that saturation does not occur. If saturation is reached, there is more distortion and less speech intelligibility.

The location where the monitoring circuit "samples" the output to signals to the receiver provides or classification of the AGC as either input-stage or output stage compression regardless of where the volume control is set.

Schweitzer (1979) pointed out that an AGC circuit has an advantage over a linear hearing aid with respect to SNR at high noise levels because the gain is automatically reduced above the compression knee-point for both signal and noise.

Bill Processing:

A circuit that automatically reduces low frequency gain in proportion to the amount of overall or low frequency input energy, has been termed as Bass Increases at Low Levels (BILL) circuit (Killion, 1990). The rationale for this type of processing stems from the notion that a fixed frequency response is not always effective in dynamically changing listening environments (Kates, 1986).

One of the first BILL circuits available commercially was the Zeta Noise Blocker (ZNB). Interestingly, this circuit was also the earliest to utilize digital technology in hearing aids. About 50% of the chip was digital for control signals, and the other 50% was analog for the signal path. This circuit determines in each of the several bands whether the signal is steady state or fluctuating (Graupe, Grosspietsch and Taylor, 1986). If steady state, the signal is assumed to be noise and the gain in that band is attenuated. If fluctuating, it is assumed to be speech, and the gain is unmodified. This approach assumes temporal differences between speech and noise and is essentially a multi-channel adaptive gain adjustment. It is, as indicated, not very useful in eliminating competing speech.

A prototype of this product was evaluated by Stein and Dempesy-Hart (1984), who showed that speech recognition in a competing 6-talker babble improved significantly for five of fifteen hearing-impaired subjects. Anecdotal evidence suggests that BILL circuits may provide more comfortable listening in noise. However, studies examining whether upward spread of making may be reduced with BILL circuits have either suffered from too much distortion in the linear mode to conduct the experiment (Van Tassell and Crain, 1992), or have used fixed rather than adaptive low frequency reduction with laboratory equipment. The latter approach does not simulate the temporal processing of BILL circuits. Overall, significant improvement of speech recognition in noise has not been demonstrated with BILL processing for a majority of subjects.

TILL Processing:

A circuit that automatically reduces high frequency gain in proportion to the amount of overall energy or high frequency at the input energy has been termed a Treble Increases at Low Levels (TILL) circuit (Killion, Staab and Preves, 1990). The K-amp circuit is an example of a WDRC-TILL circuit. Although the K-amp has not been advocated by its inventors for reducing noise, evidence exists that it may alleviate some of the difficulties of listening in noise (Knight, 1992). Certainly the circuit is very low in distortion and does not saturate in most noisy environments, thereby avoiding making the noise problem worse.

Simultaneous BILL/TILL Processing:

A dual function compression scheme, and here the BILL processor is the first stage of amplification. It provides frequency range compression, reducing the low frequency gain as the input level increases and increasing the cutoff frequency. The aim of reducing background noise, upward spread of masking and amplifier overload is achieved by a voltage controlled high pass filter rather than a compressor (Preves, Sammeth and Wynne, 1999). The resulting signal is then fed to the TILL circuitry, which provides gain compression of the remaining high frequency energy, ensuring audibility of the weak speech components and keeping the signal within the individuals dynamic range.

PILL Processing:

Programmable Increases at Low Levels, describes the 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 decrease with increasing levels, or treble response decreases with increasing levels. (Waldhauer and Villchur, 1988). This form is the most versatile, since each of its processing bands are independent of each other.

Digital signal processing

The fundamental and overwhelming advantage of Digital over Analog Signal Processing lies within the arsenal of sound processing tools that Digital Signal Processing (DSP) exposes. By making designated frequencies of sounds louder, or by adjusting loudness in certain ways, hearing aids have attempted to compensate for hearing loss. This "One-dimensional toolbox" inherently limits solution possibilities.
With DSP, all domains of sound can be manipulated or altered. Amplitude, Spectral, and Temporal characteristics of sound can be reconfigured as necessary, offering a whole new range of solutions. (Smriga, 2000). As Hawkins and Yacullo (1994) suggested, and as has been agreed upon universally, speech understanding in the presence of background noise is the single, most tenacious barrier to successful hearing aid use.

In essence, the digital hearing aid is a wearable computer. It uses digital signal processing (DSP) as a sampling technique to eliminate the need for conventional analog components (i.e., transistors, resistors, capacitors, etc.) by implementing in software the hardware devices in which those components are normally used (Katz, 1994).

Applications of DSP

Digital dual Microphone Directionality:

The dual microphone concept described earlier can also be used and enhanced in digital devices. In such digital products, the electronic delay is used to improve the directional effect beyond analog system capabilities, and is managed via algorithms rather than hard-wired control. Thus, digitally created delays can be downloaded to enhance the directional effect, create varied directional patterns (Edwards and Zezhang, 1998), or compensate for microphone drift over time, thus preserving the desired directional effect.

Amplitude Modulation Noise Reduction:

Another approach to enhancing signal perception in the presence of background noise is modulation based noise reduction. This technique, executable only with DSP technology, is based on the principle that the amplitude modulation pattern of speech over time is distinctly different from the amplitude modulation pattern of noise (Smriga, 2000). In digital systems that incorporate multi-band amplification, it is possible through modulation analysis to determine which of those bands is likely to be processing speech energy, and which is processing speech energy. Once this difference has been determined, an algorithm can be designed to reduce the gain applied to those bands that appear to be processing predominantly noise energy.

If speech energy happens to be located in a band that has been diagnosed as containing dominant noise modulation patterns, then the gain applied to that speech energy will be reduced as the gain in that band is reduced. The risk of this happening can be decreased as the number of bands involved in the noise modulation detection scheme is increased.

Digital Feed back Suppression:

As has been clearly established by Goetzinger (1978), high frequency speech sounds contribute the most to speech intelligibility. When listening to speech in the presence of background noise, Nabelek (1982) has also concluded that the more high frequency speech energy available to the listener, the greater the success in understanding speech through the competition.

However, in many hearing-aid fitting, the amount of high frequency gain that can be effectively delivered is restricted by that system's ability to contain acoustic feedback. In contrast, a unique digital feedback suppression approach offers the ability to contain feedback oscillation with out reducing (and even accessing more) high frequency gain. (Smriga, 2000).

The principle behind this feedback suppression approach is phase canceling. By introducing an oscillation signal into the feedback pathway that is the same as the actual feedback oscillation present in the fitting, but 180 degrees out of phase with it, the actual feedback oscillation signal is rendered inaudible.

Combining Multiple Functions:

One of the most important advantages of digital hearing aids and digitally controlled analog hearing aids is that the instrument can be programmed not only to adjust the characteristics of the amplifying system, but also to perform logical operations and to store in memory different sets of hearing aid characteristics. The memory v capabilities of these modern hearing aids have been found to be particularly useful in helping the user accommodate to changes in the acoustic environment. The optimum hearing characteristics for listening in a quiet room are obviously quite different from those required in a noisy bus or train. Hearing aids that change their characteristics at the press of a button provide a practical way of addressing his problem. (Levitt, 1991)

Noise Reduction:

The term "Noise Reduction" will be used here to describe techniques that process a single signal that contains the noise and speech together (directional techniques process at least two signals and use differences in those signals to reduce the noise). Separating the desired speech from the competing noise after they have both been picked up by the same microphone is extremely difficult, particularly when the competing noise is speech from one or more other talkers (Edwards, 2000).

There exist many noise reduction algorithms (viz. COMPASS) that have different frequency-response programmability for various types of noises (party, music, traffic, etc.). Ideally, a noise reduction algorithm will take advantage of temporal and/or spectral mismatches between the target speech and the interfering noise in order to remove the interference while preserving the target. Unfortunately, it is difficult for a noise reduction algorithm to identify which part of the signal is the target speech and which is the interfering noise, since both usually have energy that occupies the same spectral and temporal coordinates.

Spectral subtraction techniques (Boll, 1979) attempt to estimate the amplitude spectrum of noise and subtract this from the amplitude spectrum of the speech and noise combined. This technique has had little success in improving intelligibility, partly because of the difficulty in estimating the noise spectrum separately from the speech spectrum. Levitt, Neuman, Mills & Schwander (1986) evaluated this algorithm (originally implemented by

Weiss, Aschkenasy and Parsons, 1974), with hearing impaired subjects, but had limited success. Even though the algorithm significantly improved the SNR, an examination of the 'noise-reduced' signal revealed that the vowel formats were enhanced but the noise-like consonants were removed.

Several algorithms have been developed that attempt to reduce the gain in frequency regions where noise occurs, and initial investigations into this gain reduction technique several years ago appeared to result in improved intelligibility (Stein and Dempesy-Hart, 1984).

It seemed that reducing the gain in the low frequency region when the level in this region was high would prevent high-frequency speech from being masked and thereby improve intelligibility. (This technique is in fact used in some digital hearing aids currently in the market). Despite the seemingly obvious reason why this technique should succeed, research has not found any benefit from this technique for speech intelligibility (Punch and Beck, 1986: Tyler and Kuk, 1989; Fabry and Van Tassell, 1990; Van Tasell and Crain, 1992). Reducing the gain in the low frequency region, in fact, risks making low frequency speech cues inaudible. Levitt (1991) noted that reducing the gain in regions of noise is unlikely to improve intelligibility in practical use of a hearing aid because upward spread of masking is typically negligible at realistic levels of noise experienced by hearing aid wearers, although he was careful to note that there were some individuals in previous studies who did show some improvement.

It is possible that listening of speech in a less noisy background may require less effort by the listener even though it doesn't increase the overall understanding of speech, providing for a less exhaustive experience. (Edwards, 2000). Thus reducing noise while preserving speech intelligibility is a reasonable and useful signal-processing goal for a noise reduction algorithm.

Feedback Cancellation:

Feedback is a significant problem for many hearing aid users, since it not only limits the gain that the hearing aid can provide, but is an annoyance to the hearing aid wearers and to those around them. Solutions to this problem usually degrade speech intelligibility since most solutions reduce the gain in frequency regions where feedback occurs, limiting audibility to speech at those frequencies (Edwards, 2000)

Feedback exists when sound from the output of a hearing aid leaks back to the microphone. This is not a problem unless the gain from the microphone to the receiver exceeds the attenuation from receiver to microphone. This suggests two solutions to reduce he feedback problems:

a) Increase the attenuation from the receiver to microphone.

b) Decrease the gain from the microphone to the receiver.

The amount of gain that the hearing aid provides in the first strategy can be achieved by either reducing the vent size, or closing the vent altogether. This, however, is typically not done except in severe cases because reducing the vent size increases the patient's feeling of occlusion which is objectionable. Therefore, the second strategy is usually followed, which is to limit frequency regions of feedback. Unfortunately, feedback typically occurs at high frequencies, which is exactly where patients usually need the most gain (since majorities of hearing aid wearers have high frequency loss).

Digital signal processing (DSP), however, can effectively increase the attenuation from receiver to microphone without actually changing the acoustics of the feedback path. Ideally, if the DSP in the hearing aid knew exactly what signal was feeding back to the microphone, then the DSP could generate the same signal and subtract it from he incoming microphone signal. In this case, the feedback signal is cancelled in DSP. Any other signals picked up by the microphone from the acoustic environment is unaffected, and amplified in a normal manner.

In practical implementation, however, the feedback signal cannot be perfectly predicted by the DSP and thus is not completely eliminated. The feedback signal is significantly reduced, though, resulting in an increase in the amount of gain (provided by the hearing aid) by the amount that the feedback signal is reduced. French St.George, Wood and Engebretson (1993) found that speech understanding improved in speech babble using feedback cancellation in a prototype digital device.

Multi-channel nonlinear technology

Over the past ten years, our field has gained a significant appreciation for the role of multi channel technology as the principal solution for the sensorineural hearing loss. Good speech understanding in noise, however remains a goal that has not been met fully. (Schum, 2000)

As a class of technology, advanced multi-channel hearing aids currently define the top end of the technology ladder at present. Both DSP based products and certain advanced analog products offer a series of distinct advantages for the patient with sensorineural hearing loss. Because of the audibility and dynamic range restrictions imposed by sensorineural hearing loss, the fitting parameters of multi-channel nonlinear technology have been optimized to make the most of the residual dynamic range (Dillon, 1999; Elberling and Schum, 1996).

The advantages of Multi-channel nonlinear technology can be said to be three-fold (Schum, 2000):

a) *Two-dimensional Audibility:* Compared, to linear technology, advanced hearing instruments provide broader bandwidth (due in part to the ability to actively control acoustic feedback) and greater gain for softer sounds. The effect is that the user has greater access to a broader range of input in both frequency and amplitude domains. (Schum, 2000).

b) *Intelligent use of compression:* Given the loudness dynamics of sensorineural hearing loss and the nature of speech, considerable effort has been expended over the past several years in finding optional strategies to use compression to place a broad range of inputs into the user's remaining dynamic range without unduly disrupting the important speech cues in signal (Dillon, 1996; Hickson and Byrne, 1997; Kuk, 1998; Souza and Turner, 1998). The ability to accurately predict or map the patient's loudness perception for a variety of signal types, input levels and frequencies is well developed. A variety of well-described schemes exist to ensure that soft speech is audible, moderate speech is comfortable, and loud speech does not become uncomfortable. Further, the specific effects of compression parameters such as optimal number of channels, compression threshold, the use of expansion, attack times, release times and crossover frequencies have been discussed in various forms.

c) *Excellent sound quality:* In most advanced technology solutions currently on the market, one or typically more compression and/or expansion stages are used to provide excellent sound quality for a broad range of inputs.

While providing for good audibility for soft inputs, these circuits have been designed to effectively control higher level inputs without the distortion or annoyance of more traditional linear technologies. The advanced technology products receive excellent subjective ratings by patients (Boymans, Dreschler, Schoneveld, Verschuure, 1999; Humes, Christiansen, Thomas, Bess, Headly-Williams, Bentler, 1999; Walden, Surr, Cord, Pavlovic, 1998).

Most multi-channel hearing aids implement a fully automatic approach. The careful selection and adjustment of gain and compression characteristics typically allow the uses to operate in multiple listening environments without the need for either a volume control wheel or remote control. The result is an amplification solution that more closely mimics the functionality of normal hearing: the ability to hear and understand comfortably for a wide range of signal input levels.

It is an accepted fact that no current noise reduction scheme actually improves SNR of either for the full bandwidth of the device or for the particular channel in which the noise control algorithm is active. (Fabry, 1991; Edwards, Hou, Struck, Dharan, 1998) However, with advanced nonlinear technology, a series of secondary benefits have been realized, *viz.*, improved comfort and less annoyance in noisy situations, better sound quality, and in certain situations, a reduction in the spread of masking from one frequency region to another (Kochkin, 1996).

Combining advanced technology noise-control solutions

As mentioned earlier, circuitry-based approaches to truly improve the SNR do not currently exist in modern hearing aids (Schum, 2000). However, two popular technologies that provide direct improvement of SNR are FM signal transmission and directional microphones. Both technologies are designed to reduce the level of the competing noise in a signal before it reaches the main signal processing circuitry of a hearing aid. In realistic environments, directional microphones on hearing aids have been demonstrated to improve the effective SNR by up to 5 dB (Roberts and Schulein, 1997). FM systems have been documented to provide improvements on the order of 10-20 dB (Hawkins, 1984). It is crucial to recognize that the benefits of these noise control technologies are not redundant with the benefits provided by multi-channel nonlinear technology.

Although speech understanding in noise is a primary consideration for patients, using advanced noise control without consideration, for other needs constitutes incomplete clinical treatment. To get the most out of amplification, combinations of effective noise control technologies and multi-channel nonlinear technology can be used.

Combining FM Technology and advanced nonlinear circuitry:

The most common application of FM technology is in the classroom. The primary goal of FM, as we know it, is to provide the child with a clear audible representation of the teacher's voice. The secondary, yet important goal is to also allow the child to learn other voices in the class along with his own, through the environmental microphone on his personal hearing aid. These goals seem contradictory to each other, since the noises in the classroom would also enter via the environmental microphone, potentially negating the excellent SNR for the teachers' voice.

Schum (1999) has recently published guidelines on combining FM with both linear and nonlinear hearing aids, one of the conclusions of which is that multi channel nonlinear technology is the most effective way to achieve both goals, provided the FM signal level is set in proper relation to the signal entering the environmental microphone.

In short, since the teacher's voice is picked up by the FM microphone at a over 10 - 20 dB higher than the moderate level used to set the FM signal gain, a good SNR for the teacher's voice is maintained, and is placed within the comfortable range by the nonlinear circuit. When the teacher stops talking, the nonlinear circuit increases the gain (and hence the audibility) for signals entering the environmental microphones. The FM transmission path should be essentially linear; the hearing aid takes care of all nonlinear processing needs.

Combining Directional Technology and Advanced Nonlinear Circuitry:

To provide the best total solution to the client, certain considerations must be made when combining directional and nonlinear technologies. Not all directional microphone systems work in the same manner (Ricketts and Dhar, 1999), hence the technical characteristics of the hearing aid and the fitting algorithm should be matched to the directional characteristics of the microphone.

SUMMARY & CONCLUSION

Today, there exists a wide array of amplification for those with hearing impairment due to the advances that have occurred over the years in hearing aid design. It can, in fact, be said that there is a hearing instrument befitting for most hearing losses. Literature over the years has indicated that hearing in noise has been the greatest challenge for amplification devices, and hence S/N ratio improvement and maintenance can be said to be the single, most important determinant of perceived benefit from amplification. Lim (1983) noted that there exist many techniques for improving the SNR of speech in noise, some by as much as 12 dB, but that very few actually improve the speech intelligibility. The reason is that improving the *acoustic* SNR does not necessarily improve the *perceptual* SNR.

Hearing instrument wearers, dispensers and manufacturers are all searching for the ideal hearing instrument which will compensate for all hearing problems in a near-invisible package that would optimize auditory perception under all possible conditions. Although this seems difficult (almost Herculean) to achieve, certain aspects of hearing impairment are common enough for appropriate amplification strategies to have widespread application. These problems are *abnormal amplitude response* and *recruitment, abnormal frequency selectivity* and *spread of masking*, and *abnormal temporal resolution* and *gap detection.* Even individuals with mild hearing losses, who have little trouble understanding speech in quiet situations, exhibit significant performance degradation in noise. Since these problems are common to a large number of impaired ears, they must be considered in the design of the ideal hearing aid.

An important step towards the ideal instrument was the introduction of ***Signal Processing,*** which addressed the problems of understanding speech in noise and have resulted in more effective instruments. The use of wideband, smooth, transparent hearing instrument responses, together with binaural amplification and directional microphones, can demonstrate a S/N improvement of ***12-14 dB*** compared to a monaural, directional, narrow band response.

Today's first order, gradient ***directional microphone systems,*** in either analog or DSP instruments, can provide *4-5 dB* improvement in SNR in noisy situations. Multi-microphone directional arrays offer promise for improving SNR by an additional 4-6 dB in high levels of environmental noise.

FM transmission systems, based on the simple principle of reducing the distance between speaker and listener, provides maximal benefit to the hearing impaired individual. However, the necessity of additional equipment worn by the listener and the speaker precludes their wide acceptance.

Adaptive / active filtering and ***automatic signal processing*** technology promises significant benefits for many hearing instrument wearers, and are currently very popular. The success of advanced signal processing technology, such as programmable multiple compression, in a variety of listening situations have had the effect of encouraging manufacturers to expand upon and enhance such technology.

By taking ever-increasing advantage of the signal processing tools that ***Digital Signal Processing*** affords, hearing instruments are now showing evidence of significantly increased ability to improve speech understanding and signal quality in noisy environments. This is accomplished in a variety of modes such as enhancing directional sensitivity, input signal analysis based on signal modulation over time, and acoustic feedback suppression. And yet, this represents only the beginning. There are even more robust algorithms focussed on overcoming the negative effects of noise on speech perception that are under research and development.

At this juncture, however, note must be taken of the fact that although the current trend looks encouragingly at Advanced Signal Processing and Digital/ Programmable amplification devices, factors such as accessibility and cost-efficacy have prevented the complete popularization of these systems. The majority of our clientele still opt for conventional, analog hearing aids, even if they have to compromise on the quality of amplification: a situation that needs to be remedied (hopefully) in the near future.

In conclusion, however, it is gratifying to note that until recently, most hearing instrument wearers were counseled that hearing aids were of limited use in noisy situations. But today, there are a variety of technologies that a client can choose from to improve the signal-to-noise ratio, thus allowing him/her to participate in conversations and activities that take place in the presence of background noise.

This review has focussed solely on the various technologies that help improve the user's ability to hear and understand speech in background noise. These technologies should be applied appropriately. When suitably coupled to each other, along with basic strategies like reduction in room noise and reverberation, and the use of visual cues, they can come closer to the goal of making speech understanding possible in any situation for listeners with any degree of hearing loss.

BIBLIOGRAPHY

1. Agnew, J. (1997) .How multi microphone arrays can improve directionality. *Hear* J.48 (8), 18-20.

2. Bachler, H., 8B Vonlanthen, A. (1995). Audio Zoom-Signal processing for improved communications in noise. *Phonal focus*, 18, 3-18.

3. Bacon, S.P., & Viemeister, N.F. (1985). Temporal modulation transfer functions in normal - hearing and hearing impaired listeners. *Audiology*, 24, 117-134.

4. Bactschi, A. (1999). Wireless solutions for hearing instruments: high performance hearing solutions. *Hearing Review* (Suppl 3), 48-52.

5. Berlin, C.I. (Ed.). (1996). Hair cells and Hearing Aids. San Diego, CA: Singular Publishing.

6. Boll, S.F. (1979). Suppression of acoustic noise in speech using spectral subtraction. In Preves D. (2000). Hearing Aids and listening in noise. *Seminars in hearing*, 21 (2), 103-122.

7. Boymans, M., Dreschler, W., Schoneveld, P., & Verschuure, H. (1999). Clinical evaluation of a full-digital in-the-ear hearing instrument. *Audiol,* 38,99-108.

8. Bronkhurst, A.Q., & Plomp, R. (1990). A clinical test for the assessment of binaural speech perception in noise. *Audiology,* 29, 275-285.

9. Brooks, D.N., & Bulmer, D. (1981). Survey of binaural hearing aid users. *Ear Hear, 2,* 220-224.

10. Byrne, D. (1980). Binaural hearing aid fitting: Research findings and clinical application. In E.R. Libby (Ed.). Binaural Hearing and Amplification (pp 125 -127). Chicago: Benetron, Inc.

11. Carhart, R. (1965). Monaural and binaural discrimination against competing sentences. *Int. Audiol,* 4(3), 5-10.

12. Carhart, R., Tillman, T.W., & Greetis, E.S. (1969). Perceptual masking in multiple sound backgrounds. J Acoust Soc Am, 45, 694-703.

13. Cook, J., Bacon, S., & Sammeth, C. (1997). Effect of low-frequency gain reduction on speech recognition and its relation to upward spread of masking. *J Speech Lang. Hear Res,* 40, 410-422.

14. Dillon, H. (1996). Compresion? Yes, but for low or high frequencies, for low or high intensities, and with what response times? *Ear Hear,* 17, 287-307.

15. Dillon, H. (1999). NAL-NLI: A new procedure for fitting non-linear hearing aids. *Hear J,* 52,10-16.

16. Edwards, B., & Zezhang, H. (1998). Signal processing algorithms for new, software based digital hearing device. *Hearing Journal,* 51(9), 44-52.

17. Edwards, B., Hou, Z., Struck, C, & Dharan, P. (1998). Signal processing algorithms for a new software-based digital hearing device. *HearJ,* 51,44-52.

18. Edwards B.W. (2000). Beyond Amplification: Signal processing Techniques for improving speech intelligibility *in* noise with hearing aids. *Seminars in Hearing,* 21(2), 137-154.

19. Eisenberg L.S, Dirks D.D, & Bell T.S. (1995). Speech recgonition in amplitude modulated noise of listeners with normal and impaired. *J Speech Hear Res,*

38, 222-233b. (1996). A new digital hearing instrument. *Hear Review,* (5), 38-39.

20. Erdman, S.A., & Sedge, R.K. (1981). Subjective comparisons of binaural versus monaural amplification. *Ear Hear,* 2, 225-229.

21. Fabry, D.A., & VanTasell, D.J. (1990). Evaluation of an articulation index based model for predicting the effects of adaptive frequency response hearing aids. *J Speech Hear Res,* 33, 676-689.

22. Festen, J.M., & Plomp, R. (1986). Speech Reception threshold in noise with one and two hearing aids. *J Acoust Soc Am,* 79, 465-471.

23. Festen, J.M., & Plomp, R. (1990). Effects of fluctuating noise and interfering speech on speech reception threshold for impaired and normal hearing. *J Acoust Soc. Am,* 88 (4), 1725-1736.

24. Fitzgibbons, P.J., & Wightman, F.L. (1982). Gap detection in normal and hearing impaired listeners. *J. Acoust SocAm,* 72(3), 761-65.

25. Florentine, M., & Buus S. (1984). Temporal gap detection in sensorineural and simulated hearing impairments. *J Speech Hear Res,* 27, 449-455.

26. Fortune, T.W., 8B Preves, D. (1992a). Hearing aid saturation and aided loudness discomfort. *J speech Hear Res*, 35, 175-185.

27. Fortune, T.W., & Preves, D. (1992b). Hearing aid saturation, coherence, and aided loudness discomfort. *J Am Acad Audiol*, 3(2), 81-93.

28. Gagne, J.P. (1988). Excess masking among listeners with a sensorineural hearing loss. *J. Acoust. Soc. Am*, 83, 2311-2321.

29. Goetzinger, C. (1978). Word discrimination testing. In: Katz, J. (Ed). Handbook of clinical Audiology.(3rd Ed). Baltimore: Williams and Wilkins.

30. Gordon-Salant, S. (1984). Effects of reducing low-frequemry amplification on consonant perception in quiet and noise. *J Speech Hear Res*, 27, 483-493.

31. Graupe, D., Gosspietsch, J., & Taylor R. (1986). A self adaptive noise filtering system, part 1: Overview and description. *Hear Instrum*, 37, 29-34.

32. Gravel J., Fausel, N., Liskow, C. & Chobot, J. (1999). Children's speech recognition in noise using omnidirectional and dual microphone hearing aid technology. *Ear Hear*, 20, 1-11.

33. Greenberg, J., & Zurek, P. (1992). Evaluation of an adaptive beamforming method for hearing aids. *J Acoust Soc Am,* 91,1662-76.

34. Hawkins, D. (1984). Comparisons of speech recognition in noise by mildly to moderately hearing impaired children using hearing aids and FM systems. *J Speech Hear. Dis.* 49:409-418.

35. Hawkins D, & Yacullo W. (1984). Signal to noise ratio advantage of binaural hearing aids and directional microphones under different levels of reverberation. *J Speech than Disorder,.* 49, 278-286.

36. Hawkins, D., & Naidoo, S. (1992). A comparison of sound quality and clarity with peak clipping and compression output limiting. In Preves D. (2000). Hearing aids and listening in noise. *Seminars in hearing,* 21(2), 103-122.

37. Hickson, L., & Byrne. D. (1997). Consonant perception in quit: Effect of increasing the consonant-vowel ratio with compression amplification. *J Am Acad Audiol,.* 8, 322-332.

38. Hoffman, M.W., Trine, T.D., Buckley, K.N., & Van Tassel. D.J. (1994). Robust adaptive microphone array

processing for hearing aids: Realistic speech enhancement. J Acoust Soc Am, 96, 759-770.

39. Horwitz, A., Turner, C, & Fabry, D. (1991). Effects of different frequency response strategies upon recognition and preference for audible speech stimuli. *J speech Hear Res,* 37, 1185- 96.

40. Humes, L., Christiansen, L., Thomas, T., Bess, F., Headly-Williams, A., & Bentler, R. (1999). A comparison of the aided performance and benefit provided by a linear and a two channel W.D.R.C hearing aid. *J Speech Lang Hear Res.* 42:65-79.

41. Irwin, R.J., & McAuley. (1987). Relations among the temporal acuity, Hearing loss and perception of speech distorted by noise and reverberation. *J Acoust Soc Am,* 81(5), 1557-65.

42. Jerger, J.W., Tillman, T.W., & Peterson, J.L. (1960). Masking by octave bands on noise in normal and impaired ear .*J Acoust, Soc. Am,* 32, 385-390.

43. Kates, J.M. (1986). Siemens technical update, signal processing circuits. In: E .R.Libby, & Sweetow.R. (1987). Fitting the environment some evolutionary approaches. *Hear instrum.* 38(8): 8-16.

44. Kates, J.M. (1986). Signal processing for hearing aids. *Hear Insturm 37(2),* 19-22.

45. Kates, J.M. (1987). Theoretical and practical considerations in signal processing. *Hear Instrum,* 38(8), 23-28.

46. Kates, J.M. (1993). Hearing aid design criteria. J Speech Lang Pathology and Audiology, *Monograph* (Suppl 1), 18-19.

47. Kates, J.M. (1994). Speech enhancement based on a sinusoidal model. *J speech Hear Res,* 37,449-464.

48. Kates, J.M., & Weiss, M. (1996). A comparison of hearing aid array - processing techniques. *J Acoust Soc* Am, 96(1), 3138-48.

49. Katz, J. (1994). Handbook of clinical audiology.4th Ed. Baltimore: Williams & Wilkins.642-643.

50. Killion M.C. (1979). Evaluation of high fidelity hearing aids. In Libby, E.R., & Sweetow .R. (1987). Fitting the environment some evolutionary approaches. *Hear Instrum.* 38(8): 8-16.

51. Killion, M.C. (1985). The noise problem: There's hope. *Hear Instrum,* 36(11), 26-32.

52. Killion, M.C. (1990). A high fidelity hearing aid. *Hear lustrum,* 41, 38-39.

53. Killion, M.C, Staab, W. & Preves, D. (1990). Classifying automatic signal processers. Katz, J., (1994). Handbook of clinical audiology.(4th Ed.)(pp 683) Baltimore: Williams & Wilkns.

54. Killion, M.C, & Fikret-Pasa, S. (1993). The three types of sensorineural hearing loss. Loudness and intelligibility considerations. *Hear J,* 46(11), 31-36.

55. Killion. M.C, & Villchur, E. (1993). Kessler was right - partly: But SIN test shows some aids improve hearing in noise. *Hear J,* 46(9), 31-35.

56. Killion, M.C. (1996). Talking hair cells: What they have to say about hearing aids. In C.I. Berlin. (Ed.). Hair cells and Hearing Aids (pp 125-172). San Diego, CA: Singular Publishing.

57. Killion M.C (1997). Hearing Aids: Past, Present, future: Moving towards normal conversations in noise. *British Journal of Audiology,* 31, 141-148.

58. Knight, J. (1992). A subjective evaluation of K-amp Vs linear hearing aids. *Hear Instrun,* 43, 8-10.

59. Kochkin, S. (1996). Customer satisfaction and subjective benefit with high performance hearing aids. *Hear Rev,* 3,16-26.

60. Kuk, F. (1998). Rationale and requirements for a slow acting compression hearing aid. *Hear J,* 51,45-53.

61. Levitt, H., Neuman, A., Mills, R., & Schwander, T. (1986). A digital master hearing aid. In Edwards, B.W. (2000). Beyond amplification: Signal processing techniques for improving speech intelligibility in noise using hearing aids. *Seminars in hearing,* 21(2), 137-156.

62. Levitt, H., Neuman, A., & Sullivan, J. (1990). Studies with digital hearing aids. *Acta otolaryngology.* Suppl. 469: 57-69.

63. Levitt, H. (1991). Future directions in signal processing hearing aids. *Ear Hear.* 12:125-130.

64. Levitt, H. Digital Hearing Aids. In C.A. Studebaker, & I, Hockberg, (1993). Acoustical factors affecting hearing aid performance. New York. Allyn and Bacon.

65. Libby, E.R., & Sweetow, R. (1987). Fitting the environment-some evolutionary approaches. *Hear I*snstrum,38(8), 8-16

66. Lim, J.S. (1983). Speech Enhancement. In Edwards, B.W. (2000) Beyond amplification: signal processing techniques for improving speech intelligibility in noise with hearing aids. *Seminars in hearing,* 21(2), 137-156.

67. Lurquin, P., & Rafhay, S. (1996). Intelligibility in noise using multi-microphone systems. *Acta. Oto-Rhino-Larynga-logica Belgica,* 50, 103-109.

68. Martin, E.S., & Pickett, J.M. (1970). Sensorineural hearing loss and upward spread of masking. *J Speech Hear Res,* 13, 426-37.

69. Mims Voll, L. (2000). Application of technology to improve signal - to -noise ratio. *Seminars in hearing,* 21(2), 157-167.

70. Moore, B.C.J., Glasberg. B., & Vickers, D. (1995). Simulation of the effects of loudness recruitment on the intelligibility of speech in noise. *B.J. Audiol,* 29, 131-143.

71. Moore, B.C.J. (1998). Cochlear Hearing Loss. London: Whurr Publishers.

72. Nabelek, A.K. (1982). Temporal distortions and noise considerations. In Smriga, D.J. (2000). Three

digital processing schemes that can influence signal quality in noise. *Seminars in hearing,* 21(2), 123-136.

73. Nabelek, A.K. (1993) Communication in noisy and reverberant environments. In Studebakers and I. Hochberg. (Eds.). Acoustical factors affecting hearing aid performance. (2nd Ed.) .15-28. Needham Heights, MA: Allyn and Bacon.

74. O' Shaughnessy, D. (1989). Enhancing speech degraded by additive noise or interfering speakers. In Preves, D. (2000). Hearing aids and listening in noise. *Seminars in hearing,*1 (2), 103-122

75. Papso, C.F., & Blood, I.M. (1989). Word recognition skills of children and adults in background noise. *Ear Hear,* 10,235-236.

76. Pearsons, K.S., Bennett, R. D., & Fidell, S. (1977). Speech levels in various levels noise environments. In: Schum (1996). Speech understanding in background noise. In: Valente M (eds.). Hearing Aid: Standards, Options and Limitations, (pp.368-406). New York Thieme Medical Publishers. Inc.

77. Peters, R., Moore B.C.J., & Baer, T. (1998). Speech Reception thresholds in noise with and without spectral

and temporal dip for normally hearing and hearing impaired people. *J Acoust Soc Am*, 103, 577-87.

78. Peterson, P. (1987). Using linearly constrained adaptive beamforming to reduce interference in hearing aids from competing talkers in reverberant rooms. In; Preves, D. (2000) hearing aids and listening in noise. *Seminars in hearing,* 21(2), 103-122.

79. Plomp, L. (1994). Noise, amplification and compression: considerations of three main issued in hearing aid design. *Ear Hear,* 15, 2-12.

80. Preves D. (1978). Directivity of in-the-ear aids with non directions and directional microphones. *Hear Aid J,* 29(6), 52 - 56.

81. Preves, D., Sammeth C, & Wynne, M. (1999). Field trial evaluations of a switched directional / omnidirectional in the ear hearing instrument. *J Am Acad Audiol* 10, 273-284.

82. Preves D. (2000). Hearing Aids and listening in noise. *Seminars in Hearing,* 21 (2), 103-122.

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

84. Rankovic, C, Fryman, R., & Zurek P. (1992). Potential benefits of adaptive frequency gain characteristics for speech reception in noise. J Acoust Soc Am, 91, 354-362.

85. Rankovic, C. (1998). Factors governing speech reception benefits of adaptive filtering for listeners with sensorineural hearing loss *J Acoust Soc Am.* 103: 143-157.

86. Ricketts, T., & Dhar, S. (1999). Comparison of perfomance across three directional hearing aids. *J Am Acad Audiol,* 10,180-189.

87. Rittmanic, P.A. (1962). Pure tone masking by narrow-noise bands in normal and impaired loss. *Aud Res,* 2, 287-304.

88. Roberts, M., & Schulein, R. (1997). Measurement and intelligibility optimization of directional microphones for use in hearing aid device. In Schum, D.J. (2000). Combing advanced technology noise control solutions. Seminars in hearing, 21(2), 169-177.

89. Schum, D. (1992). Speech understanding in background noise. In Valente (Ed.). Hearing Aids: Standards, Options and limitations. (pp. 368-406). New York: Thieme medical publishers. Inc.

90. Schum, D.J. (1999). Clinical procedures for mating advanced technology hearing aids with FM systems. In Schum, D.J. (2000). Combing advanced technology noise control solutions: seminars in Hearing, 21(2), 169-177.

91. Schum, D.J. (2000). Combing advanced technology noise control solutions: seminars in Hearing, 21(2), 169-177.

92. Schum, D.J. (2000). Combining advanced technology noise control solutions. *Seminars in hearing.* 21(2): 169-177.

93. Schweitzer, H.C. (1979). Tutorial paper: principles and characteristics of automatic gain control hearing aids. *J Am Aud Soc,* 5, 84-94.

94. Smriga, D.J. (2000). Three digital signal processing schemes that can influence signal quality in noise. *Seminars in Hearing,* 21(2), 123-136.

95. Soede, W., Berkhout, A., & Bilsen, F. (1993). Assessment of a directional microphone array for hearing impaired listeners. *J Acoust Soc Am,* 94, 799-808.

96. Souza P.E., & Turner, C.W. (1994). Masking of speech in young and elderly listeners with hearing loss. *J Speech Hear Res,.* 37,661-665.

97. Souza, P., & Turner, C.W. (1998). Multichannel compression, temporal cues and audibility. *J. Speech Lang Hear Res,* 41,315-326.

98. Staab, W. (1987) Personal communication. In E.R. Libby, & R.Sweetow. Fitting the environment some evolutionary approaches. *Hear Instrum,* 38(8), 8-16.

99. Stein, L. K., & Dempesy-Hart, D. (1984). Listener-assessed intelligibility of a hearing aid self-adaptive noise filter. *Ear Hear,* 40, 15-19.

100. Szozda M.E. (1987). Active filtering for hearing aids. Hear .J. 40(4).

101. Ter-Host. K, Byrne D, Nobel W. (1993). Ability of hearing impaired listeners to benefit from separation of speech and noise. *Aust Audiol* 15(2): 71-84.

102. Thompson, S.C. (1999). Dual microphones or directional - plus - omni: Which is the Best? In S. Kochkin, & K.E. Strom. High performance hearing solutions. *Hear Review,* Suppl 3,31-35.

103. Trammell, J.L; & Speaks, C. (1970). On the distacting properties of competing speech *J Speech Hear Res.* 13: 438-448.

104. Tyler, R.S., & Kuk, F.K. (1989). The effect of" noise suppression" hearing aids on consonant recognition in speech babble and low frequency noise. *Ear Hear,* 10,243 - 249.

105. Valente, M., Fabry, D., & Potts, L. (1995). Recognition of speech in noise with hearing aids using dual microphones. *J Am Acad Audiol,* 6, 440-449.B (1991). The effect of frequency selective attenuation on the speech reception threshold of sentences in conditions of low frequency noise. *J Acoust Soc Am,* 103,1043-1057.

106. Van Tassell, D.J., Festen J., & Plomp R. (1991). The effect of frequency selective attenuation on the speech reception thresholds of sentences in conditions of low frequency noise. *J Acoust Soc Am.* 90, 885-894.

107. Van Tassell, D.J. (1992). Hearing loss, speech and hearing aids. *J Speech Hear Res,* 36,228-224.

108. VanTassell, D.J., & Crain, T.R. (1992). Noise reduction hearing aids: Release from masking and release from distortion. *Ear Hear,* 13, 114-121.

109. Villchur, E. (1993). A different approach to the noise problem of the hearing impaired. In Killion M.C. (1996). Talking hair cells: What they have to say about hearing aids. In C.I. Berlin. Hair cells and hearing aids. (pp .125-172). San Diego, CA: Singular Publishing.

110. Walden, B., Surr, R., Cord, H., & Pavlovic, C. (1998). A clinical trial of the Resound BT2 personal hearing system. Am J Audiol, 7, 85-100.

111. Waldhauer, R., & Villchur, E. (1998). Full dynamic range multiband compression in a hearing aid. *Hear J, 9,* 19-32.

112. Weiss, M.R., Aschkenasy, E., & Parsons, T.W. (1974). Study and development of the INTEL technique for improving speech intelligibility. In Edwards, B.W. (2000). Beyond amplification: Signal processing techniques for improving speech intelligibility in noise using hearing aids. Seminars in hearing, 21(2), 137-156.

113. Wolf, R.P., Hohn, W., Martin, R., & Powers, T.A. (1999). Directional microphone hearing instruments: How and why they work. In Kochkin, S., & Stom, K.E. (Eds.). High performance hearing solutions. *Hear Review,* Suppl 3, 14-25.

114. Woolf, N.K., Ryan, A.F., & Bone, R.C. (1987). Neural Phase locking properties in the absence of outer hair cells. *Hear Res,* 4, 385-46.