EARPLUG ATTENUATION – A COMPARISON OF REAL EAR ATTENUATION AT THRESHOLD (REAT) & INSERTION LOSS (IL) PROCEDURES.

REGISTER NO: **AO390003.**

A dissertation submitted in part fulfillment of the Final Year

M.Sc. (Audiology),

University of Mysore, Mysore.

All India Institute of Speech and Hearing,

Naimisham campus, Manasagangothri, Mysore-570006.

May, 2005

DEDICATED TO LORD JESUS

&

MUMMY, PAPPA

CERTIFICATE

This is to certify that this dissertation entitled **"EARPLUG ATTENUATION - A COMPARISON OF REAL EAR ATTENUATION AT THRESHOLD (REAT) AND INSERTION LOSS (IL) PROCEDURES."** is the bonafied work in part fulfillment for the degree of Master of Science (Audiology) of the student with **(Reg. No. A0390003.)** This has been carried out under the guidance of a faculty of this institute and has not been submitted earlier to any other university for the award of any other diploma or degree.

Place: Mysore
May 2005.

Prof. M. Jayaram,
Director,
All India Institute of Speech
& Hearing, Manasagangothri,
Mysore - 570006.

CERTIFICATE

This is to certify that this dissertation entitled **"EARPLUG ATTENUATION- A COMPARISON OF REAL EAR ATTENUATION AT THRESHOLD (REAT) AND INSERTION LOSS (IL) PROCEDURES."** has been prepared under my supervision and guidance.

Manjula. P

Mysore.
May 2005.

Mrs. Manjula, P.
Guide,
Lecturer in Audiology,
Department of Audiology,
All India Institute of Speech & Hearing,
Manasagangothri, Mysore - 570006.

DECLARATION

This dissertation entitled **"EARPLUG ATTENUATION- A COMPARISON OF REAL EAR ATTENUATION AT THRESOLD (REAT) AND INSERTION LOSS (IL) PROCEDURES",** is the result of my own study under the guidance of Mrs. Manjula, P., Lecturer, Department of Audiology, All India Institute of Speech and Hearing, Mysore, and has not been submitted earlier at any university for any other Diploma or Degree.

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INTRODUCTION

Hearing protective devices (HPDs) were initially used in the "high tech" fields such as the aircraft industry and later in a more general field such as "low tech" industrial environment. The industrial usage initially resulted in a "one size fits all" philosophy regardless of input noise spectrum. However, as more became understood about hearing protection and its relationship to hearing in noise, the specifications became more appropriate.

It is not always possible to reduce noise levels by treatment of the source, and then the problem may be solved by covering surrounding surfaces with acoustically absorbent materials, by the use of noise barriers, or by moving either the offending noise source or the person being exposed to another location. When it is impractical to attain enough noise reduction by these means, personal protective devices must be used. Hearing protectors provide immediate, effective protection against occupational noise. However, use of HPDs should be a last and temporary resort.

Hearing protective devices can be either earplugs, semi inserts, earmuffs or helmets with earmuffs, based on their position relative to the ear. Earplugs are inserted into the ear canal and usually remain there without any additional support. Semi inserts are located close to the entrance of the ear canal without being inserted into it and supported by a headband. Earmuffs cover the entire outer ear in much the same way as an earphone mounted in an earphone socket. The helmet type may be made to cover most of the bony portion of the head to try to prevent bone conduction. The protection afforded by a hearing protector depends upon its design features and physical characteristics of the user. Hearing protection devices (HPDs) should meet certain other requirements in design and features such as they shall be constructed in terms of material that are imperforate, material that minimizes vibration, that offers effective mass for attenuation of acoustic energies, and that will conform to the ear or ear canal to effect the seal. Other factors to be considered include comfort, bio-compatibility, durability, cleanability and its cost.

The noise reduction provided by hearing protective device is popularly referred to as ‘attenuation,’ a dictionary definition of which reads, “To lessen the amount” (Webster, Thompson and Schroter, 1956). However, attenuation is not a precisely defined acoustical term. Rather, it is more accurate to speak in terms of transmission loss, insertion loss and noise reduction.

Transmission loss is normally defined as the difference between the incident and transmitted sound power for a particular partition or barrier (Beranek, 1971, Irwin & Grat, 1979). Insertion loss, (IL) is the difference between the sound pressures levels (SPLs) or sound intensity levels, measured at a reference point before and after a particular noise reducing treatment is applied. Thus, the IL paradigm requires sequential measurements, which necessitates the ability to exactly replicate the test stimuli for open and occluded conditions. The IL of an HPD will provide the most accurate assessment of the quantity we wish to measure, namely the reduction in SPL at the listener’s ear drum/ear canal when the HPD is in use.

The hearing protective devices are capable of reducing the noise levels at the ear by 10 to 45 dB and occasionally by 50 dB, depending on their make, fit and frequency of sound. A personal hearing protective device or a combination of personal ear protectors often permit the reduction of noise at the ears, if not to a pleasant level, at least to a harmless one.

Hearing protective devices are older than the present century. But only since World War II, they have been investigated systematically in the laboratory. This has resulted in a better understanding of the way they work, how best to use them, and what can be expected out of them; this has speeded up their improvement.

There are different methods to measure attenuation characteristics of hearing protective devices. One of the popular methods of attenuation measurement is the real ear attenuation at threshold. This technique simply involves the measurement of hearing threshold for narrow bands or pure tone acoustic signals both with and without the hearing protector placed on the listener. The difference in these two-threshold estimates represents the attenuation provided by the hearing protector. The attenuation measured in this manner is used in the calculation of the effective reduction of industrial noise levels by the protector, while linearity of attenuation from low to high noise levels must be assumed. That is the real ear attenuation at threshold procedure estimates the attenuation provided by the protector utilizing threshold levels of noise intensities that seldom exceed 50-60 dB SPL.

However, the HPDs are required to be used in high noise environments and not at threshold. Measurement of attenuation at threshold may not give the same attenuation at high noise levels as investigators (Jay, 1977, Webster, Thompson and Schroter, 1956, Damongeot & Lataye, 1973) have reported a decrease in attenuation at high noise levels. However, a number of studies reveal linearity of attenuation as the levels of noise increase (Martin, 1979, Rudmose, 1982).

From the above observations it is not conclusive as to whether different procedures to measure the sound reduction bring about the same results and if the attenuation values are linear or not, with increase in noise levels. Hence the objectives of the present study are:

1) To compare the sound reduction of ear plugs using Real Ear Attenuation at Threshold and Insertion Loss procedures.

2) To compare the Insertion Loss of earplugs at 50 dB, 70 dB and 90 dB SPL stimuli presentation levels, to evaluate linearity of insertion loss.

3) To study the speech identification performance in the presence of noise with and without earplugs.

REVIEW OF LITERATURE

The most effective way to conserve the hearing of persons working in an industry is to reduce the source of noise to a level that will not cause adverse effects on the auditory mechanism as well as on the other physiological and psychological aspects. Unfortunately, in many cases this will take years to accomplish, and in others it may be nearly impossible to reach the desirable limits, because of the nature of the noise producing source. In such instances as a temporary measure or as a last resort, effective personal hearing protection devices must be used to protect the ear from the noxious levels of noise. These devices are of four major types, earplugs, semi inserts, earmuffs, and helmet with earmuffs. Earplugs are inserted into the ear canal and usually remain there without any additional support. Semi inserts close the entrance to the ear canal without being inserted into it and supported by a headband. The earmuffs are a set of cups designed to cover both the ears and to be held snugly to the head by a headband. The helmet with ear muffs type may be made to cover most of the bony portion of the head to try to prevent bone conduction and earmuffs to attenuate sounds through air conduction.

Earplugs are inserted into the ear canal and primarily intended to weaken the air-borne sound just before it impinges upon the eardrum. These fall into three categories premolded, formable and custom or individually molded types. Premolded type protectors are available in different sizes or one shape and size for a universal fit. Formable hearing protective devices (HPDs) can be expected to fit all types of ear canals and provide excellent attenuation within the insert category. Material from which these plugs are made include cotton, fine glass wool and expandable plastic. The individually made custom molded protectors are made similar to an ear mold of a hearing aid, but without a sound bore.

The basic requirement in design and characteristics of hearing protection devices include the amount of attenuation provided, comfort, cost, biocompatibility, material that will conform to the ear or ear canal to affect seal and materials that offer effective mass for attenuation of acoustic energy.

Virtually all hearing protective devices can be evaluated on the basis of their attenuation efficiency. Under ideal conditions, the intensity may be weakened (attenuated) for air-borne sounds by as much as 50 dB. In actual practice, however, the attenuation realized by inserting a protector into the ear is seldom more than 40 dB at best, and it is important for safety engineers to be cognizant of this limitation.

Noise can be transmitted through a protected outer ear directly through the protecting devices, through the device altered by the wearers, or by the device itself, setting it into vibration, by the sound pressure waves impinging on it. The effect then is that transmission of sound to the middle and inner ear is only partially attenuated at the low frequencies, or even at all frequencies. In addition, vibrations of the skull caused by impinging sound waves are transmitted to the inner ear by way of outer and middle ear or directly to the inner ear. This limits the amount of attenuation attainable with hearing protectors.

Acoustic Attenuation

The acoustic attenuation of hearing protectors is usually expressed in decibels at various test frequencies. According to a study by Waugh, 1974, the dBA attenuation of an ear protector is a function of the C-A value of the noise spectrum in which it is used and may vary more than 20 dB in noises of different C-A value. However, in noises of similar C-A value, a given EPD provides similar amounts of dBA attenuation. Pure tone and 1/3-octave band measurements of ear protector attenuation are identical. The influence of noise spectrum shape on the octave band attenuation resulting from a given set of 1/3 octave attenuation value is practically negligible, in typical broad-band industrial noise spectra. In the octaves centered at 500 Hz and above, hearing protector attenuation should be measured at 1/3-octave intervals to avoid the substantial errors, which can occur when measurements are restricted at octave intervals.

Now considering the factors determining the sound attenuation provided by the hearing protectors, the most important one is the Insertion Loss introduced by the hearing protector between the sound source and the eardrum of the listener. This is accomplished by a change in the sound field, which is usually considered negligible, and the transmission and insertion loss between the outer and inner surfaces of the hearing protector, which can be defined as the ratio of the sound pressure at the inner surface of the hearing protector to the sound pressure at its outer surface (Zwislocki, 1957).

Speech recognition with and without hearing protective devices

Adequate verbal communication with workmates and the perception of acoustic warning signals are of prime importance for persons working in a noisy occupation. The problem in communication is the usual reason behind the resistance to wearing ear protectors shown by workers with normal hearing or hearing loss.

It seems reasonable to assume that hearing protectors attenuate as much speech as noise, which means that the signal-to-noise ratio (S/N) remains the same. Therefore, the use of hearing protectors in a noisy environment should not affect speech recognition significantly (Pekkarinen, Vilajanen, Salmivalli, and Suonpa, 1990) in practice; the situation seems to be far more complicated. In addition to the signal-to-noise ratio, speech perception is also affected by speech and noise levels. Speech recognition increases as the speech level rises, until it reaches the maximum and remains good also for higher intensities. Due to the spread of masking at high speech level if noise is present, performance starts to deteriorate. This phenomenon may be significant in practice, since along with the rise in ambient noise level, the speaker increases his vocal effort. The extensive aural overload may in addition be reinforced by reverberation.

In the early work of Kryter (1946), the test subjects were college students with normal hearing. He found that the use of earplugs had no effect against noise of less than 80 dBA, and actually contributed to a gain of about 10% in the discrimination of words for higher noise levels. The benefit of earplugs was more marked when the test was performed in a reverberant chamber (time = 1.6 s). Pollack and Picekett(1958) concluded that there is little difference in speech perception with or without ear protection up to noise levels of 100-110 dB SPL. Also Wilkins and Martin (1978) as cited in Berger (1986) measured the change in masked thresholds for a siren or a bell signal and found a slight advantage of 3-4 dB with the protectors at 75 and 95 dB SPL noise. All their subjects had normal hearing.

Berger (1980) cited in Berger (1986) studied effect of hearing protectors on communication. He reported that HPDs have little or no effect on the ability of normal hearing to understand speech in moderate back ground noise (75 dBA) but HPDs begin to decrease speech discrimination as back ground noise is reduced even further. HPDs will decrease speech discrimination for hearing impaired listeners in low to moderate noise situations. At high noise levels greater than 85 dBA HPDs actually improve speech discrimination for normal hearing. For hearing impaired the effect of HPDs on speech discrimination at these high levels is not unequivocal, but the results seem to indicate no significant effect.

The results of the extensive studies of Abel, Alberti, Haythornthwaite and Riko (1982) showed that at a noise level of 85 dBA, the intelligibility of monosyllables deteriorated, with a decrease in the speech-to-noise ratio, and were poorer in crowded noise than in white noise. The ear protectors had no effect on the speech perception of normal listeners, but caused a substantial decrement in those with hearing impairment. On the contrary the results of Chung and Gannon (1989) indicated that at a high signal-to-noise ratio of +10, subjects with normal hearing obtained higher word discrimination scores with ear protectors. Conversely, at a low signaltonoise ratio of -5, or when the subjects suffered from hearing loss, speech recognition scores deteriorated with the use of hearing protectors. Bauman and Matson (1986) also found that at 85 dB SPL (signal-to- noise ratio of 0), subjects, both with normal hearing and with high-tone loss, achieved significantly poorer scores in the California Consonant Test when using Bilsom Universal Muff protectors.

Pekkarinen, Salmivalli, Vilajanen and Suonpa (1990) studied recognition of low-pass filtered words in a noisy and reverberate environment with and without earmuffs. They presented the unfiltered and filtered words at two speech levels. 60 dBA and 85 dBA to a group of students with normal hearing. The background noise level (quiet, S/N+10, + 5.0) and reverberation time (2.1 s and 1.6 s) were altered in a hall. Unfiltered words were recognized better without earmuffs in quiet conditions, at S/N + 10 and +5. At S/N 0 dB, however, better scores were obtained when earmuffs were used. Filtered words were recognized significantly better without earmuffs at both reverberation times at all S/N conditions tested. The decrease in the reverberation time improved all the recognition scores essentially.

Pekkarinen, Vilajanen, Samivalli and Sunopa (1990) studied speech recognition in a noisy and reverberant environment with and without earmuff. The benefit of earmuffs was especially demonstrated at high speech and noise levels (85 dBA, S/N 0), and under high reverberation (2.1 s). At normal speech level (60 dBA) and in quiet conditions, the recognition of the test items was better when earmuffs were not used. The variations in test conditions in the earlier studies referred may be the possible reason for the discrepancy of the results. In quiet conditions, the perception was worse with the earmuffs than without them, although at a speech level of 60 dBA.

In noisy conditions, speech perception decreases, because S/N remains poor. The recognition of non-words decreased much more than that of sentences and words. Non-native workers may have overwhelming problems in speech communication in noisy environments. The discrimination depends not only on S/N, but also on the spectral and time characteristics of the noise. That is why speech perception in listening situation with other competitors may vary, and may deviate from these results.

Methods for measuring attenuation

There are many standardized methods available for measuring attenuation characteristic of a hearing protection device. Most manufacturers publish the attenuation curves of their products; but the published test data must be carefully examined to determine if a device will be suitable for a given user.

There are basically two main ways of measuring attenuation of a hearing protection device, they are

I. Subjective methods

   A. REAT,

   B. Above threshold procedures (Masking, Loudness balance, Midline lateralization, Temporary threshold shift and Speech intelligibility)

II. Objective methods

I. SUBJECTIVE METHODS

Let us first discuss the subjective methods in brief. There are two main procedures:

A. Real Ear Attenuation at Threshold (REAT)

B. Above threshold procedures

A. Real Ear Attenuation at Threshold (REAT) or Sound field REAT

Probably the oldest, and certainly the most common method of measuring HPD attenuation is the absolute threshold shift technique, often labeled as REAT (Watson & Knudson, 1944). Virtually all available manufacturers' reported data are derived via this method. Conceptually the idea is very simple; determine a subject's threshold of hearing without a HPD (open threshold), and then remeasure the subject's hearing threshold level while wearing the HPD (occluded threshold). The difference between the two thresholds, the threshold shift, is a measure of the attenuation afforded by the device.

REAT method has been incorporated in a number of standards in India and abroad. Three ANSI standards have been promulgated for obtaining REAT measures, they include, the original 1957 standard (ANSI Z.24.22-1957) that involves the use of pure tone stimuli presented through a speaker in a directional sound field, and the 1973 (ANSI S3.19) and 1987 (ANSI S12.6) standards employing 1/3 octave bands of noise in a diffuse sound field.

There are a number of factors that affect REAT measurements. They are ambient noise, stimulus bandwidth, directivity, frequency range, physiological noise, level dependence, number of subjects, inter and intralaboratory variability, subject fitting, selection, procedural difficulty and experimental accuracy.

Ambient Noise

A fundamental requirement of accurate REAT measurements is that the test room be “sufficiently” quiet, since high ambient noise levels will tend to mask, and hence elevate, the open ear thresholds while leaving the occluded ones unaffected. This reduces the threshold shift, resulting in lower measured attenuation values. Waugh (1973) clearly demonstrated this effect for two different earmuffs. He suggested a background noise criterion based upon a comparison of the open ear hearing threshold levels of the test subjects to normative data.

The basis for this calculation is the 1/3-octave band (OB) diffuse-field threshold data from Berger (1985), combined with Hawkins and Stevens’ (1950) data on the ratio in dB between the masked threshold of a pure tone and the pressure spectrum level of a uniform masking noise at the same frequency. This ratio has come to be known as the critical ratio. Assuming that open ear unmasked thresholds are themselves masked by an internal or intrinsic noise which combines on a power basis with other noises that may be present (French & Steinberg, 1947; Lochner & Burger, 1961 and Zwislocki, 1957), tolerable noise levels can be computed. The calculation procedure is treated in greater detail by Berger and Kerivan (1983) who experimentally verified it by comparing the measured and predicted threshold elevations of 1/3 octave band test stimuli that were produced by a low-level, low-frequency masking noise.

Waugh (1973) experimentally determined the levels of broadband background noise that would mask low frequency open ear threshold levels by 0, 3, 5, 7, 5 and 11.5 dB; averaged across 11 test frequencies from 100 to 1000 Hz. His 0 dB masking level was the quietest sound level he could attain in his test chamber, and actually lower than necessary for determining subjects' unmasked thresholds. The procedure was checked against Waugh's data by predicting the masking that should have been created by the sound levels he reported. For the frequencies of 125 Hz, 250 Hz, 500 Hz and 1 kHz, the predictions for the 11.5 dB masker the agreement was within 1.4 dB, 1.1 dB, 0.6 dB, and 0.8 dB respectively. Thus, assuming that the basic threshold data used for the computation represent typical values, the procedure should yield a suitable estimate of permissible ambient noise.

Stimulus Bandwidth and Directivity

The earliest REAT standard, (ANSI Z24, 22-1957) specified the use of pure-tone stimuli, preferably presented in an anechoic chamber and generally at a frontal incidence. Measurements in this type of acoustical environment can be easily perturbed by small amounts of subject movement. In terms of directionality, the same cannot be considered to be characteristic of typical industrial or military noise environments. Since HPD attenuation can be a function of the angle of sound incidence, especially for earmuffs, where effects as large as 15 dB have been noted for angular changes of 90 degree the similarity between the directionality in the test and use conditions must be ascertained (Bolka, 1972).

Another potential problem of pure-tone test data at selected discrete frequencies is that protector resonances can cause attenuation to vary rapidly with small changes in frequency in the region above 500 Hz. Therefore, pure-tone attenuation measured at octave band (OB) or even 1/3-OB center frequencies may not accurately reflect the average noise reduction afforded by the test device in those particular octave bands.

Webster, Thompson and Beitscher (1956) compared REAT for pure-tone, 1/2-OB, and 1/1-OB stimuli. In computing the acoustic attenuation for the noise-band data, they applied critical-band corrections to both the open and occluded thresholds, since they argued that the threshold-determining critical band could differ for the two thresholds, thus leading to inaccurate attenuation estimates if one simply compares the two thresholds without making the requisite correction. However, they also indicated that the error would never exceed 2 dB, value of small practical significance within the context of the other errors inherent in the overall methodology. They concluded that the use of bands of noise or pure tones yielded results that were “roughly equivalent”.

Bolka (1972) and Waugh (1974) have compared REAT results using pure-tone and 1/3 OB stimuli. Their conclusions were similar to those of Webster, Thompson and Beitscher (1956) showing even closer agreement between the two types of stimuli, as would be expected due to the narrower band-widths of the noise bands they employed. The small differences that existed tended to indicate generally lower attenuation for the noise-band data, especially at and above 1 kHz.

Differences of this type would be expected since the occluded threshold for a band of noise will be determined by the minimum attenuation (maximum sound transmission) within the band. Statistically, this attenuation is likely to be less than the attenuation found at any one pure-tone test frequency.

Not withstanding the similarity between pure-tone and 1/3-OB, REAT results by Waugh (1974) demonstrated that small but significant errors arise when the assumption is made that a protector's attenuation for a particular OB noise is identical to the attenuation for pure-tone or 1/3 OB stimuli centered within that octave band. He measured the pure-tone attenuation at 1/3-octave center frequencies from 100 Hz to 10 kHz, assumed that it was identical to the 1/3-OB attenuation at those same frequencies, and then computed the OB attenuation from the contiguous 1/3 – OB values. He found that the OB-attenuation values that were derived by using only the central 1/3- OB within each octave over estimated the computed OB values by 2 dB or more (on the average) for the OBs from 1 to 4 kHz, with errors for individual protectors of as much as 9 dB being observed at 4 kHz.

ASA STD 1-1975; ANSIS12.6-984; BS 5108: 1974, 1983 and ISO 4869-1981 cited in Berger(1986) have attempted to rectify the acoustical deficiencies of pure-tone anechoic testing by specifying stimuli that are 1/3 OBs of noise presented at random incidence, in a non-directional (quasi diffuse) sound field.

Bolka (1972) has compared ANSIZ24.22-1957 standard with ASA STD 1-1975 data, and Whittle and Robinson (1977) and Martin (1977) have compared ANSI Z24.22 with BS 5108:1974 data. In reviewing their results, it was assumed that ASA

STD 1 and BS 5108 should yield equivalent data due to their very similar electro acoustical requirements. All authors found the diffuse-field tests to yield lower attenuation values for earmuffs in the 1 kHz to 4 kHz region, especially at 1 kHz, where the differences were approximately 5 dB. For earplugs, the attenuation values were not significantly different.

Whittle and Robinson (1977) attributed the differences at 1 kHz for earmuffs to diffraction effects in the pure-tone tests. Bolka (1972) pointed out that, since earmuff attenuation is a function of the angle of incidence, “a minimum of attenuation at a specific angle of incidence will appear as the controlling value and will tend to decrease the attenuation compared with that measured in free-field unless the minimum occurs at the incidence at which the free- field measurement was made” (Bolka, 1972). These studies have also found a tendency for lower standard deviations for diffuse-field noise-band measurements than for anechoic pure-tone data, although a more recent report (Martin, Mozo & Patterson, 1982) that compared ANSI Z24. 22-1957 with ASA STD 1-1975 results for 12 different HPDs found no significant differences. The possibility that noise-band data yield lower standard deviations may be due to the greater ease with which subjects can determine noise-band thresholds (Whitham & Martin, 1976) cited in Berger (1986) as well as the more stable sound field characteristics in the diffuse environments.

Frequency Range

Although the range of required and / or optional test frequencies specified in the standardized REAT methods is either 63 Hz or 125 Hz to 8 kHz, a few authors have reported data beyond this frequency range. In the low-frequency region, Nixon, (1979) reported pure tone REAT data for three unspecified earmuffs at frequencies down to 35 Hz. Their results showed attenuation to be relatively constant below 125

Hz, but they noted that, at those low frequencies, masking due to physiological noise would be a significant artifact contaminating the data.

Since hearing sensitivity decreases at the rate of approximately 100 dB/octave at the upper end of the audio range and since most HPDs provide relatively good inherent attenuation at higher frequencies, it becomes very difficult to generate ultrasonic acoustical stimuli at levels sufficient to be detected by hearing-protected test subjects. Accordingly, ultrasonic-REAT data appear to be absent from the literature and only a few studies even discuss the range above 8 kHz (Berger, 1985; Schroeter and Poesselet, 1984 and Townsend & Bess, 1973).

Berger's study was conducted in conformance with the electro acoustic requirements of ASA STD 1-1975, and extended through the 1/3-OB test frequencies of 12.5 and 16 kHz. He found that attenuation was either constant or decreased slightly above 8 kHz, but since all the HPDs tested provided attenuation of at least 32 dB at those frequencies, exact estimation of the attenuation was somewhat academic.

Kumar, Venkatesh and Ragini (1982) used REAT measure to study attenuation of E.A.R. ear protectors. An attenuation characteristic of ear protectors was measured at 250 Hz, 500 Hz, 1000 Hz, 2000 Hz, 4000 Hz, 6000 Hz and 8000 Hz. Under TDH-49 ear phones using Beltone-200 C Diagnostic audiometer. Subjects were normal hearing individuals of age 18-21 years. A minimum real ear attenuation of 27 to75 dB HL was observed at 250 Hz, as maximum of 46 dB HL was observed at 8000 Hz. Attenuation was observed to increase with increasing frequency.

Physiological Noise

An artifact that contaminates REAT data is masking of occluded-ear thresholds as a result of physiological noise. This noise, which is vascular and or muscular in origin, is primarily a low-frequency phenomenon (below 1 kHz) (Anderson & Whittle, 1971, Shaw, 1979). When the ear is occluded or covered, physiological noise is amplified due to the occlusion effect (Watson & Knudsen, 1944; Zwislocki, 1957; Huizing, 1960 and Tonndorf, 1972).

The occlusion effect describes the phenomenon of the enhancement of the outer ear bone conduction path when the ear canal is occluded by an insert, or covered by a canal cap or earmuff. Low-frequency acoustical energy generated by canal wall vibrations will more effectively radiate and couple to the eardrum in this condition due to the modified termination impedance presented by the blockage. According to Schroeter and Poesselt (1984), another, and perhaps more important, low-frequency mechanism (below 500 Hz) in the case of circum-aural HPDs is the generation of sound pressure under the earmuff cup as a result of its vibration against the side of the head. This phenomenon, which they called the HPD inertia effect, was first reported Shaw (1979).

Once the eardrum is excited, regardless of the mechanism, the energy is transmitted along the ossicular chain to the cochlea as though it were a normal external acoustical signal. The occlusion effect is maximum at low frequencies, diminishes with increasing frequency, and becomes negligible by approximately 2 kHz. Since physiological noise causes both skull and ear canal vibrations, it is amplified in the occluded ear, as are all other sources, which excite canal wall vibrations.

Amplified physiological noise will mask occluded-ear thresholds at low frequencies (Rudmose, 1982 and Villchur, 1972), elevating them by approximately 4.5 – 11 dB at 125 Hz and 1-10 dB at 250 Hz. The lower estimates appear more reliable (Schroeter & Poesselt, 1984). Although Shaw (1979) estimated the influence of physiological noise on REAT evaluation of earmuffs, and Rudmose (1982) has recently fully explained its effect upon threshold testing under audiometric headphones (“the missing 6 dB”), its exact relationship to threshold testing of hearing protector attenuation was first experimentally verified by Berger and Kerivan (1983). Using an ear canal mounted subminiature microphone, they measured physiological noise in occluded and unoccluded test conditions and also measured the IL of insert, semi aural, supra aural, and circum-aural HPDs when exposed to broadband noise with an overall SPL of 93 dB. Measurements spanned 1/3 OBs from 125 Hz to 2 kHz. Insertion loss was also measured using an REAT procedure, and the magnitude of the occlusion effect was examined via bone conduction audiometry.

The REAT and IL data reported by Berger and Kerivan (1983) agreed well at 500 Hz and 1000 Hz, but not at 125 Hz, 250 Hz, and 2000 Hz. The degree by which the REAT values exceeded the IL values at 125 Hz and 250 Hz could be predicted by measuring the physiological noise in the occluded threshold. This was expected since the amount by which the occluded threshold was elevated due to masking would add directly to the amount that threshold was elevated by the IL of the HPD. The REAT measurement, which is based upon the difference between the open and occluded thresholds, would therefore be inflated by the amount of masking.

Berger and Kerivan (1983) also found that the occlusion effect, the degree of amplification of physiological noise, and hence, the amount by which the REAT values were spuriously amplified, were all dependent upon the occluded volume. They concluded that an REAT evaluation of an HPD would yield data representative of the attenuation that the device would actually provide. In the lower frequencies (125 Hz and 250 Hz), they found that the REAT paradigm yielded attenuation values that were spuriously high by 2 to 5 dB for semi aural, supra-aural, and low-volume circum-aural devices. For insert devices and medium-to-large-volume earmuffs, the over estimation tended towards zero.

Level Dependence

When HPDs are evaluated by an REAT test, the incident sound is typically at low levels, less than 60 dB SPL. The question then arises: “Does this accurately represent HPD attenuation at higher sound levels?” One of the primary reasons, for the development of the above-threshold procedures was to answer this question. Briefly, it can be stated that for intentionally linear protectors, i.e., devices not containing valves, orifices, thin diaphragms, or active electronic circuitry, attenuation appears to be independent of SPL up to very high sound levels. In fact, the more recent U.S. and Australian REAT standards (ANSI S12.6-1984 and AS 1270-1983) specifically addressed this issue with statements such as, “Although the methodology of this document yields data that are collected at low sound levels (typically 10-60 dB, re: 20 μPa), they are representative of the attenuation values of hearing protectors as elevated sound levels” (ANSI S12.6, Sec. 1.3). Therefore, REAT test results can be assumed to be indicative of performance in high-level-noise environments. This will be discussed at greater length in subsequent sections of this review.

Number of Subjects

A critical issue affecting the accuracy and reliability of all subject-based tests is the size of the subject population, as well as the related concerns of subject selection and training. In one of the earliest studies that addressed the problems of intra-and inter subject variability, Dickson, Hinchcliffe and Wheeler (1954) considered 20 subjects to be the “very minimum” needed to indicate the behavior of a particular HPD. Although they measured the performance of two ear-plugs and four supra-aural and circum-aural types of protectors, using both 20 subjects tested once and one subject tested 20 times, their data were not analyzed or presented in such a manner as to lead to firm conclusions regarding the best experimental paradigm.

Later, Hansen and Blackstock (1958) reported an experiment in which the attenuation of a V-51R earplug was measured using 20 subjects with five replications per subject. Of their 20 original subjects, two had to be replaced because they could not get a "good" fit with the four earplug sizes (small, medium, large, and extra large) that were in use at that time. The fitting criteria that were used included evaluation of attenuation – if a test value "indicated a poor seal by showing little or no attenuation at the low frequencies" (Hansen & Blackstock, 1958), subjects were instructed to remove and then reinsert the earplugs. The analysis indicated that, with three replications, a 6-dB-wide 95% confidence limit would be achieved using only eight subjects at the frequencies from 125Hz to 4000 Hz, and 17 subjects if the two highest test frequencies (6 kHz and 8 kHz) were also considered. They concluded that ten subjects with three replications was a reasonable compromise.

A more comprehensive analysis of experimental designs was conducted by Howell and Martin (1973). They examined data from nine studies, including that of Hansen and Blackstock, and concluded that intersubject variance was significantly greater than intrasubject variance thus warranting testing of additional subjects. The authors recommended an experimental design of 15 subjects, with two replications per subject, and suggested that this would typically yield a 95% lower confidence limit of 3 to 4 dB below the mean. Whittle and Robinson (1977) also recommended using additional subjects for testing earplugs, but concluded, for earmuffs, that inter subject variance was not significant and therefore stressed the importance of additional replications rather than additional subjects.

Examination of the REAT-variability problem was reported by Brinkman and Richter, (1985) cited in Berger (1986) who measured the attenuation of two different earmuff type protectors three times on different days with a group of 30 subjects. They used the procedure specified in ISO 4869-1981. The subjects varied in experience from novices who were participating in tests for their first time to well-trained wearers who had taken part in more than ten protector evaluations. Brinkmann and Richter concluded, "The random uncertainty of sound attenuation measurements using groups of ten test subjects is very high even in the case of easily fitted hearing protectors. Since the intersubject variability clearly dominated the repeatability, the most efficient way to decrease the uncertainty is to increase the number of subjects tested rather than to repeat the measurement several times". They indicated that further work would be required before the optimum number of subjects could be determined.

Since laboratory protocol (Berger, Kerivan & Mintz, 1982 and Webster & Rubin, 1962), i.e., how subjects are selected and how they are fitted with the HPD, is perhaps the single most important testing variable, this could well explain the somewhat divergent opinions expressed above concerning the relative importance of intra and intersubject variability. Although the standard deviation should indicate the ability of a device to fit a randomly sampled group of subjects (who are hopefully representative of the intended population of users), it has in many instances been found to be an indication of "measurement accuracy which perhaps depends more on experimental skill than on any other factor" (Shaw, 1979).

Current standardized practice, with a few exceptions, that a minimum of ten subjects shall be utilized for real-ear testing, although the British, in the foreword to the current version of BS 5108:1983 recommend that more subjects and / or replications than the minimum be used whenever possible. The available data suggest that this recommendation should be carefully considered, since it is likely that the ISO practice (ISO 4869-1981) of requiring a minimum of ten subjects to be tested is based more upon an appreciation of the difficulties in obtaining larger test samples than in the potential inaccuracies in extrapolating small-sample laboratory data to in-field populations.

Inter- and Intra-laboratory Variability

Regardless of the sample size, variability of test data both within and between laboratories must be examined. As early as 1962, researchers observed that "In comparing results from different laboratories then, the crucial factor is the experience and tolerance of the listeners" (Webster & Rubin, 1962). The experience they referred to was related to the ability to insert a plug or wear a muff properly and what the feeling of a "good-fitting seal" was like. Much subsequent experience has confirmed their findings.

It has been proposed that audiometrically experienced subjects have more consistent responses and that this reduces intra-laboratory testing variability as well, if it could be assumed that all the subjects in all laboratories were equally experienced (Michael, 1982). In this author's experience, subjects quickly learn to take reliable audiometric tests so that after a maximum of five test sessions, audiometric experience is no longer a critical issue.

Berger, Kerivan and Mintz (1982) reported an inter-laboratory variability experiment that was sponsored and coordinated by the Environmental Protection

Agency (EPA). Four HPDs, representing a wide range of currently available types, were tested. Seven laboratories participated directly; data obtained separately from an eighth laboratory also were included in the evaluation. The results showed significant variation in both mean values of attenuation and standard deviation. The European data were obtained prior to publication of the study done by Berger, Kerivan, and Mintz (1982), but too late for inclusion in their report. They stated that the variability between laboratories was probably attributable to differences in HPD fitting practices, as well as subject selection, training, and motivation procedures. In addition, a portion of the variability seemed to be due to different data reduction techniques employed by the facilities. A data analysis experiment involving raw data provided by one laboratory and analyzed by that laboratory, as well as the EPA and two other laboratories, resulted in a range of computed noise reduction ratings (EPA, 1979) of 7.2 dB.

Berger, Kerivan and Mintz concluded that data for a group of devices could not be meaningfully compared unless all tests were conducted in one laboratory. In this regard, reports from the National Acoustic Laboratories (1984), Martin, Mozo and Patterson (1982), and Tobias (1975, cited in Berger, 1986) are very useful, in that each present REAT data for a large number of HPDs all tested in one laboratory, each using essentially equivalent test-to-test procedures, equipment and protocols.

The variability problem is not solely the province of REAT or even of subjective testing. As pointed out above, one of the most crucial variables influencing attenuation is the protector / subject or protector / test fixture interface. To the extent that this control data variability, experimenter fitting practices and subject selection criteria will be the governing parameters with regard to intra and interlaboratory variability regardless of the test method utilized (Tobias, 1975, cited in Berger, 1986).

An example of inter-laboratory variability with REAT data for three HPDs, measured over spans of 4 to 6 years in our laboratory using various groups of ten subjects for each test. Testing was in conformance with ASA STD 1 -1975, except that fitting was accomplished “consistent with a reasonable degree of comfort”. Of particular interest is the change in the standard deviation in 1982 on the earmuff tests. This was principally attributed to one subject who could not get her bushy hair totally out from under the earmuff cushions, and yet it was decided that our laboratory’s protocol did not permit excluding her from the tests.

Subject Fitting and Selection

Efforts to reduce interlaboratory variability have involved attempts to more stringently define experimenter fitting techniques and subject exclusion procedures, so that optimal HPD performance can be measured. The assumption is that there will be less variability among laboratories in the measurement of optimal performance than in the measurement of typical real-world data. The validity of that assumption has not been resolved at this time. Moreover, some standards writing groups (BS 5108:1974, 1983; ISO 4869-1981) recognize that optimal attenuation values are of little use to designers, purchasers, or users who need an indication of the protection that HPDs can normally be expected to provide. They have incorporated in their fitting procedures statements such as “test subjects shall be instructed to adjust the hearing protector for the best attenuation consistent with reasonable comfort. Regardless, all of the standards do permit exclusion of certain test subjects if they appear to provide abnormal data. This is probably a necessity due to the limited numbers of subjects (10 to 15) that are typically tested.

In recognition, of the importance of subject selection and fitting techniques, the most recent American HPD standard explicitly states in both its abstract and introductory sections that "The methodology (of this standard) is intended to yield optimum performance values which will depict the noise-reducing capabilities of hearing protectors only to the extent that users wear the devices in the same manner as did the test subjects" (ANSI S12.6-1984, section 1.2). The crucial question of the ability of a laboratory attenuation test to accurately reflect the protection afforded in real-world applications is one that transcends all of the test methodologies discussed in this review regardless of their seeming accuracy and sophistication, and at the same time is one that cannot be decided on its technical merits alone.

Procedural Difficulty and Experimental Accuracy

REAT methods have been criticized for being too time consuming. Since each open or occluded ear test will require a minimum of approximately five minutes (for nine frequencies), the three open and three occluded ear tests for each subject will usually last 45-60 minute protector, so that there will be 8 to 10 hour of testing per device for ten-subject experiment. However, other subjective methods will be equally time consuming if multiple frequencies are also to be tested. An equivalent number of subjects will also be required due to variability in fitting HPDs and variability in subjective judgments. The equipment requirements for REAT tests are no more stringent than for other subjective tests and often less so, except for the room requirements. Some laboratories do have problems meeting the ambient noise and / or diffusivity criteria.

When evaluating a test method, an important consideration is whether or not all the relevant sound paths to the inner ear are considered. At least four sound paths to the occluded ear have been discussed in the literature (Berger, 1986; Nixon, 1979 and Veneklasen, 1955). These include air leaks, piston like vibration of the HPD, transmission through the material of the HPD. Khanna, Thondroff and Queller (1976) have described three different bone conduction paths or components of outer ear (vibration of ear canal walls), middle ear (excitation of ossicular motion), and inner ear (direct mechanical distortion of cochlea). In general, all the subjective methods account for these various sound paths since the protector is being tested in situ and, since the judgments are based on the stimulation of the inner ear, the final link in the acoustical chain.

Headphone REAT

Headphone REAT tests are identical to the ASA STD 1-1975 protocol except that the sound field is established inside a set of circum aural enclosures outfitted with small loudspeakers (Michael, Kerlin, Bienvenue, Prout & Shampman, 1976). This makes the testing considerably more portable and also less sensitive to ambient noise, since the headphones provide attenuation during both the open and occluded ear tests. The main disadvantage is that only insert HPDs can be tested. The method is ideally suited to in-field measurement of HPD attenuation to determine real-world performance. Experiments that have been conducted (Crawford & Nozza, 1981; Edwards, Hanser, Broderson, Green& Lempert, 1983; Hachey & Roberts, 1983 and Padilla, 1976) are discussed and summarized by Berger (1983).

Under laboratory conditions with identical subjects, ASA STD 1 testing and a suitable headphone REAT evaluation yield comparable data. Small correction factors (generally less than 5 dB) are adequate to predict one result from the other. But when the headphone procedure is applied in typical industrial environments, the results indicate that HPDs perform considerably better in the laboratory than in the real world. This is primarily due to the way in which laboratory subjects wear HPDs compared to the manner in which they are worn in actual use conditions (Berger, 1983).

A variant of the (circum aural) headphone REAT paradigm for insert device evaluation consists of measuring IL using (supra-aural) audiometric earphones such as TDH-39 drivers mounted in MX-41/AR cushions. The reliability of this approach is questionable because: (a) the earphone may actually rest on the protector causing it to either break its seal or to be pushed further into the canal, depending upon the design of the HPD and the particular fit being tested; (b) resting an earphone on the pinna can distort the concha and / or canal, changing the fit of the insert; (c) possible plug/earphone contact can create structural instead of acoustical excitation of the insert; and (d) the acoustical excitation of the plug inside the small occluded volume under the supra-aural device is not representative of sound-field excitation and may actually change as the occluded volume is reduced by the application of the HPD.

These potential problems notwithstanding, users in occupational settings are tempted to implement such a measurement in order to estimate the protection being obtained by their employees in practice. Only one author has provided comparative data to assess the accuracy of such an earphone-measurement technique, and those data are only from a pilot study of limited extent (Berger, 1983). The results demonstrated that for certain earplug/ear canal combinations the audiometer test yielded large overestimates (greater than 10 dB) of the sound-field attenuation. However, for 84% of the comparisons, the agreement between the sound-field and audiometeric REAT methods was within 10 dB at 500 Hz, a degree of accuracy unsuitable for laboratory testing, but one that could yield useful estimates of the protection of wearers under field conditions.

REAT with Hearing-Impaired Subjects

This technique was first suggested by Thunder and Lankford (1979). Their specific purpose was to investigate HPD attenuation in high sound level environments. They hoped to accomplish this by comparing HPD attenuation for five sensory neural hearing-impaired subjects (average hearing threshold levels greater than 40 dB at all frequencies) to that for five normal-hearing subjects. They found poorer attenuation at all frequencies (250 Hz to 8 kHz) for the hearing-impaired subjects. They concluded that HPD efficiency was reduced at high sound levels.

No other authors have reported a study with the sole intention of replicating the preceding results, but data from other studies can be analyzed with this in mind. For example, Berger (1986) investigated HPD performance on a large number (100) of naïve subjects. The subjects were unselected, except for an otoscopic inspection to exclude outer ear infection or impacted cerumen. Comparing attenuation for the worst hearing listeners (four to nine subjects, depending upon frequency) to the remaining normal-hearing subjects showed no evidence of level-dependent effects. Results are only shown above 2 kHz since only at those frequencies was the average hearing threshold level of the subjects significantly poorer than normal: 38 dB at 3 kHz, 41 dB at 4 kHz, 46 dB at 6 kHz, and 42 dB at 8 kHz re: ISO R226-1961 corrected for diffuse-field listening.

Another study conducted by Abel, Alberti, Haythomthwaite and Riko (1982) that concerned HPD effects on speech intelligibility also yields useful data. They conducted REAT evaluations of HPD attenuation on eight groups of 12 subjects each. Three groups had normal hearing, three had bilateral noise-induced high-frequency losses (5 to 25 dB at 500 Hz with a slope in loss of 35 to 65 dB between 500 and 4000 Hz), and two groups had bilateral flat losses (30 to 50 dB at 500 Hz and 45 to 65 dB at 4000 Hz). One circum aural and two insert HPDs were evaluated, although not all groups were tested on all devices. Comparisons between the attenuation achieved by the hearing-impaired versus normal-hearing groups were variable, being dependent both upon frequency and the device tested. The hearing-impaired groups demonstrated significantly more attenuation at 3 and 4 kHz for the earmuff, whereas, the normals had significantly more attenuation at 6 and 8 kHz for the earplug. This latter result may have been an artifact due to limitations in the equipment used to generate the test stimuli.

The wide divergences among the data suggest that using hearing-impaired subjects introduces additional variables into an REAT test, which at this time, may not be fully explainable. Certainly, the data do not make a strong case for the dependence of HPD attenuation on sound level.

In 1979, Thunder and Lankford tested hearing protector attenuation at threshold in five-normal hearing subjects and five with flat sensorineural hearing losses. They used pre-molded, triple flanged earplug and subject fit procedure. The results revealed that hearing impaired subjects received significantly less real ear attenuation threshold than individuals with normal hearing.

Berger (1986) reported the results of an experiment in which hearing threshold level was correlated with real ear attenuation at threshold for an expandable earplug. Using seven subjects with losses at 3 kHz and above, hearing threshold levels of 30 dB HL or greater, Berger found no significant correlation between hearing threshold level and attenuation at threshold. In fact, the attenuation at 3150 to 8000 Hz was virtually the same for hearing impaired and normal hearing subjects.

Frank, Murphy, Johnson and Simon(1996) studied real ear attenuation of ear muffs in ten normal hearing and ten hearing impaired individuals using testing procedures followed the specifications of ANSI SI 2.6 – 1984. Measurement was done for the one-third octave bands. The results showed that the hearing impaired subjects received slightly more attenuation than the normal hearing subjects at all frequencies, but these differences were not statistically significant. These results provide additional support to the finding that hearing protection devices are capable of providing as much attenuation to hearing impaired users as they do to normal hearing individuals.

B. Real Ear Attenuation using above threshold procedures.

It is always reassuring when a quantity can be measured by alternative technique and similar values result. The above threshold procedures offer this capability. Furthermore, they permit investigation of (a) the possibility of level depend attenuation effects (2) REAT errors arising from masking due to amplification of physiological noise and additional methods of measuring the performance of HPDs under field conditions. The above threshold procedure can generally eliminate the need for expensive test chambers necessary to ensure acoustical environments with sub threshold noise levels. The above threshold procedures will account for all acoustic transmission paths to the occluded and unoccluded ear since the final subject responses are based on the excitation of inner ear.

(i) Masking

Earliest description for a masking technique for attenuation measurements was given by Webster,Thompson and Schroter, (1956). He described placing an active earphone over the hearing protective device while presenting masking noise via loud speakers in the test chamber. The subject's noise masked threshold for earphone signal was then found for both the protected and unprotected conditions. He assumed that the change in the masked threshold corresponded to the attenuation provided by the HPD as long as the noise elevated the threshold for the earphone stimuli by at least 20 dB in the protected condition. This method is only suitable for evaluating circum aural protectors.

(ii) Loudness Balance

In its most simple form, the loudness balance procedure requires a subject to adjust a suprathreshold test stimulus for equal loudness under two conditions, with and without HPDs. Another version requires the subject to adjust sounds, presented alternatively to the two ears via headphone, for equal loudness. One ear is then occluded by an insert and the loudness balance readjusted. In either case, the difference in signal level that is required to reestablish the balance is a measure of the HPDs attenuation. Loudness balance procedures are deceptively simple in concept, but as Rudmose (1982) and Theile (1985) have discussed, they are subjected to many experimental artifacts like:

a. Mechanical coupling of subject's chair to the loud speaker
b. The far or near loud speakers source problem
c. The earphone and loud speakers distortion problem
d. The formal procedure for performing balancing
e. For the monaural ears the problem of successfully occluding the nontest (or transfer) ear and these can affect the validity of the results.

(iii) Midline Lateralization

In addition to threshold and loudness balance decisions, human subjects are also adept at making lateralization judgments. Lateralization occurs with headphone presented acoustic stimuli, it describes the sensation that arises when the sound source appears to be inside the head. If sounds of similar intensity and pitch are presented to the two ears via head phones, the lateralized location is the middle of the head that is mid line lateralization. When subjects perform midline lateralization at the same position in the head, with and without an ear plug in the test ear (no earplug in the reference ear) the difference between the sound levels in the test ear for the two conditions is a measure of the HPD attenuation.

The midline lateralization procedure was described by Fleming and Cudworth (1979) and Fleming (1980) cited in Berger (1986). The midline lateralization paradigm offers no advantages in speed, accuracy or implementation relative to REAT testing, except that it can be conducted in higher ambient noise levels (approximately 60 dBA).

(iv) Temporary Threshold Shift

Any auditory phenomenon that is dependant on the intensity of the acoustic stimulation ear in theory be used to infer the attenuation provided by an HPD. The only aural after effect that seems to have actually been used in temporary threshold shift (TTS) is the change in threshold sensitivity at a particular frequency, measured at some designated time after a specified exposure. The difference between the SPLs necessary to produce particular TTS in the protected and unprotected conditions respectively is the effective protection. For e.g., suppose that a 5-minute exposure to a 1000 Hz tone at 100 dB produce a TTS_2 (TTS measured 2 minutes after exposure) at 1500 Hz of 20 dB in a particular listener's unprotected ear. If with the HPD in place, it is necessary to raise the SPL from 100 dB to 130 dB in order to produce the same TTS_2, then the HPD has provided 130 – 100 = 30 dB of attenuation.

The TTS method has some serious limitations other than the obvious fact that several exposure with the HPD in place will generally be necessary in order to find the SPL that will produce the target TTS. Use of these high levels means that, if the HPD is at all effective, the protected ear will be given exposures that are increasingly hazardous to the unprotected ear as the SPL is gradually raised during successive exposures.

Therefore, great care must be taken to fit the HPD consistently. In addition, the TTS developed must be limited to 20-30 dB in order to ensure complete recovery and no permanent damage. Because of the relative inefficiency of the TTS method and the unavoidable hazard associated with its use, a more common implementation has been the measurement of the reduction in TTS generated by a given particular exposure.

(v) Speech Intelligibility

Like the TTS reduction method, speech intelligibility test can be used to rate the relative performance of hearing protectors. For e.g. the speech reception threshold, the level necessary for 50% correct identification of bisyllable word lists can be evaluated with and without hearing protection. The difference in dB between the speech reception thresholds is then a measure of the HPD attenuation. The draw back of this approach is that it lacks frequency specificity. There are other psychophysical methods developed such as cross modality loudness scaling, magnitude estimation and reaction time

II. OBJECTIVE METHODS

As with the subjective above threshold procedures, the objective methods provide the ability to measure HPD performance at levels above threshold. Furthermore, the first two of the methods to be discussed, The Acoustical Test Fixture (ATF) and miniature microphone in real-ear methods (MIRE), and can expedite data acquisition, especially with today’s computerized signal analysis systems. The other objective methods, microphone in cadaver ears and aural reflex threshold shift

(ARTs) do not save time and in fact, create procedural problems like intense sound pressure level required to elicit reflex with ear protection, limited frequencies that can be used and laboratory method

The objective methods (with the exception of ARTs) do not directly account for all of the sound paths to the occluded ear, and bone conduction is either incorporated via post measurement computerized adjustments, or ignored altogether. Even the cadaver ear method does not fully account for bone conduction.

A. Acoustical Test Fixtures (Artificial Head / Ear) (ATF) method

The ATF measurement is conceptually the most appealing of the test methods. Ideally, it would eliminate the need for subjects, provide accurate and repeatable results, reduce test times, accommodate a wide variety of acoustic test signal and be suitable for product design, automated testing, and quality control monitoring. The perfect ATF has not yet been developed, but much literature exists describing efforts in that direction.

The preceding discussions concerning sound field parameter for standardized. REAT tests pertain equally to ATF tests. Free or diffuse sound fields and direction of incidence will affect the results in similar ways, whether a real or artificial head is being protected. The only advantage in this regard that the ATF offers is the ability to tolerate higher test room noise levels, providing that the test signals are sufficiently amplified. An ATF will, of course, be a model of a real head or an average real head. The degree to which it must mimic the mechanical and acoustical behaviors of real heads is one of the first problems to be addressed. The most basic approach would be to simply mount a microphone in a box of suitable dimensions with its diaphragm either flush mounted or slightly recessed behind one surface. The insertion loss is then measured by monitoring the SPL with and without the HPD. This model ignores the possible importance of head geometry, skin, bone, cartilage dynamics, pinna and concha effects, and the eardrum and ear canal impedance. Different acoustics test fixtures have been developed by Acoustic Society of America (ASA), International Organization for Standardization (ISO) and Knowles Electronics Manikin for Acoustic Research (KEMAR). However, these ATFs have not physically modeled the dynamic structure of the human skull. The models of skull vibration are a function of both frequency and method of excitation and couple (via) multiple pathways to the cochlea. Thus far, this complex vibratory system has eluded successful analytical or mechanical modeling.

B. Microphone in Real Ear (MIRE) method

An alternative to using an artificial head as a test fixture in which to place a measurement microphone is to use a real head. This procedure is similar in speed and capability to the artificial head method, while offering the advantage of a more accurate test fixture that exhibits all of the anthropometric features and leakage paths that real world HPD users do. Unfortunately, this method neglects the important BC paths, as does the artificial head approach, although post-measurement corrections can be applied.

When measuring in real ears, either insertion loss (IL) (using one microphone) or noise reduction (NR) (using two microphones) measurements are possible. The IL measurements are more relevant to actual user protection but in this case; tend to limit the usable test SPLs since, in the unprotected condition, a real ear will be exposed to approximately the same levels as the microphone. Thus the NR measure offers more flexibility.

In the mid-1950s, researchers began investigating the use of microphone in real ear method (Dickson, Hinchcliffe & Wheeler, 1954, and Webster, 1955). Until recently this technique has been limited to measuring circum aural and supra aural HPDs due to difficulty of mounting a microphone or probe to be in the canal in conjunction with the insertion of an earplug (Berger & Kerivan, 1983).

Weinreb and Touger (1960) found close agreement between microphone in real ear and loudness balance data. Further more, the microphone – IL and loudness balance values agreed reasonably well with REAT data for the same devices, with the REAT averaging 3 to 5 dB higher except at 2 kHz when they were lower. This latter feature can explain those devices whose attenuation starts to approach BC thresholds. It is most likely to occur at 2 kHz where BC is most sensitive. For that condition the REAT value will be limited by the bone conduction paths, where as the microphone in the canal will not sense energy conducted to the ear via that path, and therefore, will measure a lower sound level hence a higher IL.

Another interesting feature of the Wienreb and Touger's (1960) study was that the variability they found for the objective data was no smaller than for the subjective data. The data of Berger and Kerivan (1983) also confirm this latter observation to a significant extent as is the reported results of Dickson, Hinchcliffe, and Wheeler (1954). This suggests that the primary variability in the REAT paradigm is the placement of the HPD or perhaps variations in the test stimuli, and not uncertainty in the determination of the subject thresholds.

Villchur (1972) measured the IL of earphone devices mounted in MX – 41/AR ear cushions using a probe tube microphone assembly mounted in the concha. He compared his data to ANSI 724-22 – 1957 attenuation values provided by Copeland and Mowry (1971) for the same type of device. The differences were less than or equal to 3 dB from 500 to 8000 Hz but increased to 4.3 and 6.9 dB at 250 and 125 Hz respectively. The REAT values exceeded the microphone-measured values at all frequencies. The author suggested that the error was in the REAT procedure and that it was due to masking arising from physiological noise. He provided confirmation of his hypothesis by comparing earphone and free field thresholds for the five subjects. The earphone thresholds were masked by the same amount that the REAT exceeded the IL measured values.

The most comprehensive comparison of REAT and microphone in real ear data, and the only one to include semi aural and insert HPDs was conducted by Berger and Kerivan (1983). They limited their investigation to frequencies up to and including 2 kHz, since only up to that frequency was the SPL measured in ear canal substantially independent of the microphone position. This assured that IL would be unaffected even if the microphone moved slightly during application or removal of the HPD their data confirms the occlusion effect/ physiological noise hypothesis, as well as the general accuracy of the REAT methodology. Shenoda, Fischer and Ising (1987) measured attenuation of earmuffs for high-level impulsive noise in anechoic room and reverberation room. A ¼ inch condenser microphone was used to carryout acoustic measurements. The microphone was held by the test person at the entrance of the ear canal in order to measure the impulse noise without the earmuff. During this measurement, the test person used an earplug to protect his ear against the impulse noise. To measure the inside noise level, the same microphone was fixed in the centre of the 'Bilsom Blue' earmuff through a hole, which had been closed with plastic sealing. A loudspeaker, which was fed with continuous noise from a random noise generator through a band pass filter, was fixed at a distance of 1.5 m and directed towards the microphone. Insertion loss was measured in both anechoic and reverberation room. The results showed that the noise attenuation of the measured 'Bilsom blue' earmuff was found to be very low at lower frequencies for the human head in either an anechoic or a reverberation room. Theoretically, the attenuation increases above the resonance frequency with a slope of 12 dB/octave but, in practice, it is limited by the air leakage and partial vibration of the cushion.

The sound attenuation of earmuffs under conditions of high impulsive noise decrease with increasing peak levels due to a non-linearity. The cushion causes this nonlinearity. At high peak levels the attenuation of earmuffs is higher for perpendicular sound incidence than for frontal sound incidence. The sound attenuation in the diffuse field of a reverberation room is higher than the mean attenuation in the free field for comparable peak levels. Due to secondary pathways, the attenuation of an earmuff on a human head is much lower than that measured on an artificial head. Thus, the artificial head can only be used for model studies, for instance of non linearity.

Dillon and Murray (1987) have demonstrated quite convincingly, the equivalence of functional and insertion gain. Hawkins and Stevens (1950) have shown that real ear measures of insertion gain are highly reliable if care is exercised in positioning the probe tube at a constant location within the ear canal.

Due to the success of the probe tube microphone system in real amplification measurement, Gerling, Metz, Roemer, Bonko and Rowsey (1989) used it in measurement of attenuation of hearing protectors. Specifically they used it in comparing foam insert hearing protectors using probe tube microphone method and threshold method. The results revealed that the functional attenuation and real ear attenuation lies within 1 dB to 7 dB of each other through 3000 Hz. Above 3000 Hz the probe results show a progressively marked decrease in the amount of occlusion loss when compared to the behavioral attenuation obtained in this study. These results can again be supported by the explanation that comes from Berger and Kerivan's (1983) study.

In another study by Traynor, Ackley and Wierbowsky (1989) the mean attenuation in dB SPL (functional measurement) was compared with those obtained from the probe tube microphone conditions. Insertion held condition, where the subject was made to hold the HPD in place with their finger to increase the hermetic seal afforded by the HPD. Insertion condition, where the HPD was not held in position but simply inserted as far as possible. A difference in mean attenuation in dB of about 17.94 dB between the real ear at threshold condition and the simple insertion condition was obtained while a difference of 6.25 dB from real ear at threshold and insertion held condition was obtained. Although this difference was smaller than the insertion condition the variability was much higher.

The difference in average dB SPL attenuation noted between the real ear at threshold and the probe tube technique were thought to be created by the loss of the HPDs hermetic seal due to the introduction of the probe tube between the HPD and ear canal.

Nixon (1979) in a review of several methods available for assessment of attenuation characteristics of hearing protectors cited five criteria that a procedure must satisfy in order to make it suitable for standardization. The five criteria were that it:

1) Was relatively simple
2) Had universal application
3) Yielded results that could be generalized to the total population

4) Was not too costly

5) Was not time consuming

C. Insertion Loss method

It appears that the probe technique of assessing hearing protection device of attenuation does hold promise in satisfying the above five criteria. In addition, it also provides information over a large array of frequencies and changes in attenuation in 1 dB step can be determined. There is also elimination of subject's threshold response variability. There is no contamination of thresholds by room noise. Room noise could be a problem with REAT measurements, especially, if the subjects have normal thresholds; but this method neglects the importance of bone conduction path.

The popular method of evaluating attenuation characteristics of a hearing protective device is the absolute threshold shift technique or real ear attenuation at threshold. However, the HPDs are required to be used in high noise environments and not at thresholds. Measurement of attenuation at threshold may not give the same attenuation at high noise levels.

Martin (1979) utilized a small microphone to measure attenuation in cadaver ears for 1/3 octave bands of noise and impulses spanning an intensity range from 75 to 175 dBSPL. He found that attenuation was linear over the range examined. Rudmose (1982) also found similar results using a technique incorporating miniature microphones placed in ear canals of living human subjects in protected and unprotected conditions.

Webster, Thompson and Schroter, (1956) and Weinreb and Touger (1960) used miniature microphone and loudness balance techniques. They found that attenuation provided by hearing protective devices at high levels of noise is considerably less than that indicated by real ear attenuation at threshold data. Results showed an overestimation of attenuation by 2 to 5 dB at frequencies below 500 Hz by the real-ear threshold procedure. Specifically, increased physiological noise resulting from occlusion of the ears can artificially inflate protected thresholds by 2 to 5 dB for test frequencies below 500 Hz.

Jay (1977) observed non-linearity in hearing protector attenuation at sound levels of 110 dB SPL. A cross-modality matching paradigm was used in a manner similar to the way the magnitude-estimation procedure was implemented in the study. Pure-tone signals of 500 and 1750 Hz were delivered to subjects in a free-field environment. The average unprotected and protected loudness-growth functions converged as sound level increased from 50 to 110 dBSPL. This convergence was interpreted as a decrease in attenuation of the hearing protector at high sound levels. An alternate explanation, however, for the convergence of the two loudness growth functions is as follows. Because relatively low-frequency signals were utilized in the study by Jay (1977), the unprotected loudness-growth functions at high intensities are not actually an unprotected function. Rather, contraction of the middle-ear muscles at high sound pressure levels (110 dBSPL) resulted most likely in some amount of attenuation at the signal frequency in the unprotected conditions. In the protected condition, however, the attenuation provided by the hearing protector at 110 dB SPL most likely reduced the sound pressure level of the test signal to a level below reflex threshold so that the protected loudness-growth function was uncontaminated by the attenuation produced by the acoustic reflex. Thus, the convergence of the unprotected and protected loudness-growth functions at high sound levels observed by Jay (1977), may be due to attenuation being present at the highest levels for the unprotected conditions rather than attenuation of the hearing protector increasing at high intensities.

Damongeot and Lataye (1973) cited in Berger (1986), utilized a masked bone conduction procedure somewhat more complex than utilized in this study to investigate the linearity of hearing protector attenuation. The discrepancies in attenuation at low and high sound levels observed by these investigators were confined primarily to low frequencies. Physiological noise may again play a role in this discrepancy. The particular procedure developed by Damongeot and Lataye (1973), moreover, is most likely influenced by other contaminating factors as pointed out by Brinkmann and Brocksch (1976) cited in Berger (1986). Specifically, with any masked bone conduction procedure at least 45 dB of masking must occur in the unprotected condition. Otherwise, the observed reduction in masking associated with placement of the hearing protector on the subject will not reflect appropriately the amount of attenuation provided by the protector. If, for example, only 15 dB of masking is produced in the unprotected condition, then the maximum possible reduction in masking is also 15 dB. It was for this reason that only the highest noise level possible, which always produced more than 45 dB of masking, was utilized in the masked bone-conduction procedure of this study.

The question that arises then is which method to use. A real-ear psychophysical procedure of some type is preferred in that one is attempting to estimate the protection provided to the wearer of the protector. More objective procedures making use of miniature microphones, whether on manikins, cadavers, or live humans, invariably involve some assumptions about the role of the bone-conduction pathways that are not assessed directly with these procedures. Of the options available that make use of miniature microphones, the use of living humans as subjects would be preferred so as to provide more realistic indications of the variability in attenuation estimates. Even in this case, however, measuring sound levels with a microphone in the ear canal of the subject misses any contributions that the middle-ear inertial and compressional components of bone conduction hearing might make toward a decrease in attenuation at high sound levels.

METHOD

The present study aimed at comparing the attenuation of ear plugs measured using a subjective procedure Real Ear Attenuation at Threshold (REAT) and an objective procedure Insertion Loss (IL). In order to investigate this, the following method was used.

Subjects:

Thirty normal hearing adults, 15 females and 15 males, in the age range of 18 to 40 years, with a mean age of 23years, participated in the study and the subjects satisfied the following criteria.

1) No significant history of any auditory disorders.
2) Normal hearing threshold of less than 15 dB HL at audiometric frequencies from 250 Hz to 8 kHz.
3) With spoken knowledge of Kannada.

Instrumentation

1) A calibrated diagnostic audiometer (Madsen OB922, version 2) was used for ensuring normal hearing and REAT measurements.
2) A calibrated immittance meter (GSI-TS) to ensure normal middle ear function.
3) A calibrated hearing aid analyzer (FP 40, version 3.5) was used for Insertion Loss (IL) measurement.
4) Flanged earplugs- Tricheck (triple flanged) - were used for attenuation measurement.

Stimuli Used:

Warble tones (at octave frequencies from 250 Hz to 8 kHz) for REAT measurement and speech identification test in Kannada (Vandana, 1998) for subjective evaluation of earplug performance.

A composite signal of 50 dB, 70 dB and 90 dB SPL was used for IL measurement. Composite signal is a complex signal, consisting of seventy nine speech weighted frequencies, which are presented simultaneously.

Test environment: A sound treated test room with ambient noise levels within permissible limits. (re: ANSI 1991, cited in Wilber (1994).

Procedure: Measurement of attenuation of earplug was carried out using the following procedures. Real Ear Attenuation at Threshold (REAT)

1. Insertion Loss (IL)

1. Procedure for REAT:

REAT measures are based upon determination of the difference between the minimum level of sound that a subject can hear without wearing an HPD (open threshold), and the level needed when the subject is wearing an HPD (occluded threshold). The difference between these two thresholds, the threshold shift, is a measure of the REAT afforded by the device (Berger, 1986).

The subjects were seated comfortably on a chair, in the patient room. The ear selection was such that the right ear was tested for 15 subjects and left ear for the other 15 subjects. IL measures were also done on the same ears for which REAT was carried out.

Unoccluded Measures

a) Unoccluded threshold for the test ear under earphones was established for warble tones at 250, 500, 1 k, 2 k, 4 k and 8 kHz.

b) Unoccluded speech identification score (SIS) was measured at a presentation level of 50 dB HL in quiet, and in the presence of speech noise at 50, 70 and 90 dB HL. Live speech was presented through ear phone and speech noise also was routed through the same ear phone.

c) For speech identification scores, the subjects were instructed to repeat the list of 25 PB words (of Vandana's PB list, 1998). Each word correctly repeated was given a score of one. Speech identification scores were not converted to percent.

Occluded Measures

The test ear was fitted with flanged earplug by the tester by pulling the pinna back wards and upwards to ensure good fitting.

a) Occluded threshold (with flanged earplug) for the test ear under earphones were established for warble tones at 250, 500, 1 k, 2 k, 4 k and 8 kHz.

b) Occluded speech identification score at 50 dB HL in quiet, and in the presence of speech noise at 50 dB, 70 dB and 90 dB HL were also obtained. Live speech was presented through earphone and speech noise also was routed through the same ear phone.

c) For speech identification scores, the subjects were instructed to repeat the list of 25 PB words (of Vandana's PB list, 1998). Each word correctly repeated was given a score of one. Speech identification scores were not converted to percent.

Differences in decibels between the occluded and unoccluded threshold measures were computed for each frequency for each subject to compute the REAT of the flanged ear plug. The REAT for all the subjects were tabulated for further analysis.

2. Procedure for Insertion Loss (IL) measurement

Insertion Loss is the difference between the sound pressure levels (SPLs), or sound intensity levels, measured at a reference point before and after a particular noise reducing treatment is applied.

a) The subjects were positioned with the test ear at a distance of 1 foot and 45 degrees azimuth from the speaker of FP40 hearing aid analyzer equipment. The speaker of FP40 was located on the speaker stand as for real ear measurement. The test ear was the same ear (right or left) for which the REAT measurement was carried out.

b) The sound field was equalized by leveling.

c) The probe tube was placed in ear canal 25-30 mm past the tragal notch.

d) Composite signal was selected to record the real ear unoccluded responses (REUR), at 50, 70 and 90 dB SPL. REUR was recorded separately for each level of the input signal.

e) The flanged ear plug was placed in the subject's test ear canal along with the probe tube and adjusted for adequate fitting by pulling pinna backward and upward; by the tester.

f) Real ear occluded (with flanged ear plug) responses(REOR) of the test ear, for the same signal at the above presentation levels, i.e., 50, 70 and 90 dB SPL, were measured by keeping the probe tube at the same location in the ear canal as was used in real ear unoccluded measurement.

g) Real Ear Insertion Loss (REIL) was automatically displayed by subtracting the Real-Ear Occluded Response (REOR) from Real Ear Unoccluded Response (REUR). REIL at different frequencies for the test ear for each subject was tabulated.

The tabulated REAT and REIL measures were subjected to statistical analysis.

RESULTS

The aim of this study was to compare the REAT and IL procedure. For this purpose 15 males and 15 females in the age range of 18 to 40 years were taken and data was collected.

SPSS for windows (version.10) was used to analyze

1) The difference in REAT and IL measures
2) Mean, standard deviation (SD) and range of REAT, SIS, IL
3) Linearity of IL

Independent sample 't' test was carried out for both IL and REAT measures to find out, if there was any gender difference in the data. There was no significant difference between the genders (significance value 'P'>0.005) in the data collected. Hence, the data from 15 male subjects and 15 female subjects were combined to form a single group to form a total of 30 data points.

REAT measure:

From Table 4.1, it is observed that the mean attenuation and variability at higher frequencies is greater than at lower frequencies.

Table 4.1: Mean, SD and range of attenuation for REAT Measurement

	Attenuation (dB)		
Frequency	Mean	S.D	Range
250 Hz	21.33	12.45	0 -50
500 Hz	22.83	12.43	0-45
1 kHz	23.16	11.02	0- 40
2 kHz	29.17	10.59	10-50
4 kHz	29.5	16.63	0-75
8 kHz	40.32	16.42	0- 65

Speech identification scores in occluded and unoccluded conditions:

Table 4.2: Mean, SD and range for speech identification scores in unoccluded and occluded conditions.

	Speech Identification Scores	
Condition	Mean(SD) unoccluded	Mean(SD) occluded
In quiet	24.87 (0.35)	19.87(3.33)
In 50 dB noise	22.9 (1.95)	17.63 (3.23)
In 70 dB noise	9.47 (6.37)	8.03 (6.01)
In 90 dB noise	1.57 (3.93)	1.6 (4.14)

Table 4.2 shows that as the signal-to-noise ratio reduced SI scores worsened in both occluded and unoccluded conditions. The variability in SIS scores among subjects is also higher at lower SN ratios.

From repeated measures, it was found that there was highly significant difference in SI scores between all the conditions (in quiet, in 50, 70 and 90 dB SPL noise levels) tested.

Paired sample t test was carried out to compare unoccluded and occluded speech identification scores. From the tables 4.2 and 4.3, it is clear that only in quiet and in the presence of 50 dB noise there is highly significant difference (P less than .001) between occluded and un occluded conditions. In the presence of 70 and 90 dB SPL noise levels there was no difference in speech identification scores.

Table 4.3: Results of paired sample t test for unoccluded and occluded conditions.

Conditions	df	t value	Significance
In quiet	29	7.90	.000
In 50 dB	29	7.54	.000
In 70 dB	29	1.47	.151
In 90 dB	29	1.43	.143

IL measure:

Table 4.4: Shows that IL increased with increase in frequency. At higher frequencies, IL increased with increase in input levels.

Table 4.4: Mean, SD and range of IL measurements at input levels of 50 dB, 70 dB and 90 dB SPL.

Frequency	Insertion Loss (dB)					
	50 dB SPL		70 dB SPL		90 dB SPL	
	Mean (SD)	Range	Mean (SD)	Range	Mean (SD)	Range

200 Hz	4.89(5.60)	0 to 18.2	3.92 (4.40)	0.2to30.3	2.81(4.46)	0.1to35.4
500 Hz	6.77(9.10)	0.1to32.6	7.35 (97.38)	0.1to27.0	5.89(7.28)	0.3to14.3
1 kHz	9.18(6.97)	0.1to27.5	10.12 (6.99)	0.6to25.0	8.66(6.96)	0.3to28.1
2 kHz	13.52 (8.19)	0.3to33.7	17.61 (8.40)	0.7to30.1	15.04(9.27)	0.4to33.5
4 kHz	15.96(95.08)	2.6to25.3	19.63 (9.32)	6.5to36.4	15.80(7.48)	0.7to37.1
8 kHz	10.54(8.61)	0.4to30.3	13.71(10.63)	0.1to35.40	15.5(10.05)	4.5to37

Linearity of IL measure.

Multiple Analysis of Variance (MANOVA) was performed to see the effect of different levels of stimulus (50, 70 & 90 dB SPL) on insertion loss at different frequencies. The results showed that there was no significant difference in IL measured at different levels (P>0.05). i.e.; the IL was linear with increase in presentation levels at all frequencies (Table 4.5).

Table 4.5: Results of MANOVA to evaluate linearity of IL.

	Frequency	df	‘F’ value	Level of Significance
Between the levels 50, 70 and 90 dB.	200	2, 87	1.38	0.258
	500	2, 87	0.28	0.753
	1 kHz	2, 87	0.34	0.712
	2 kHz	2, 87	2.02	0.139
	4 kHz	2, 87	2.79	0.067

There was no significant difference in IL at different presentation levels (50, 70 & 90 dB SPLs), hence the IL at 50, 70 and 90 dBSPL were averaged at each frequency.

Comparison of REAT and IL procedures:

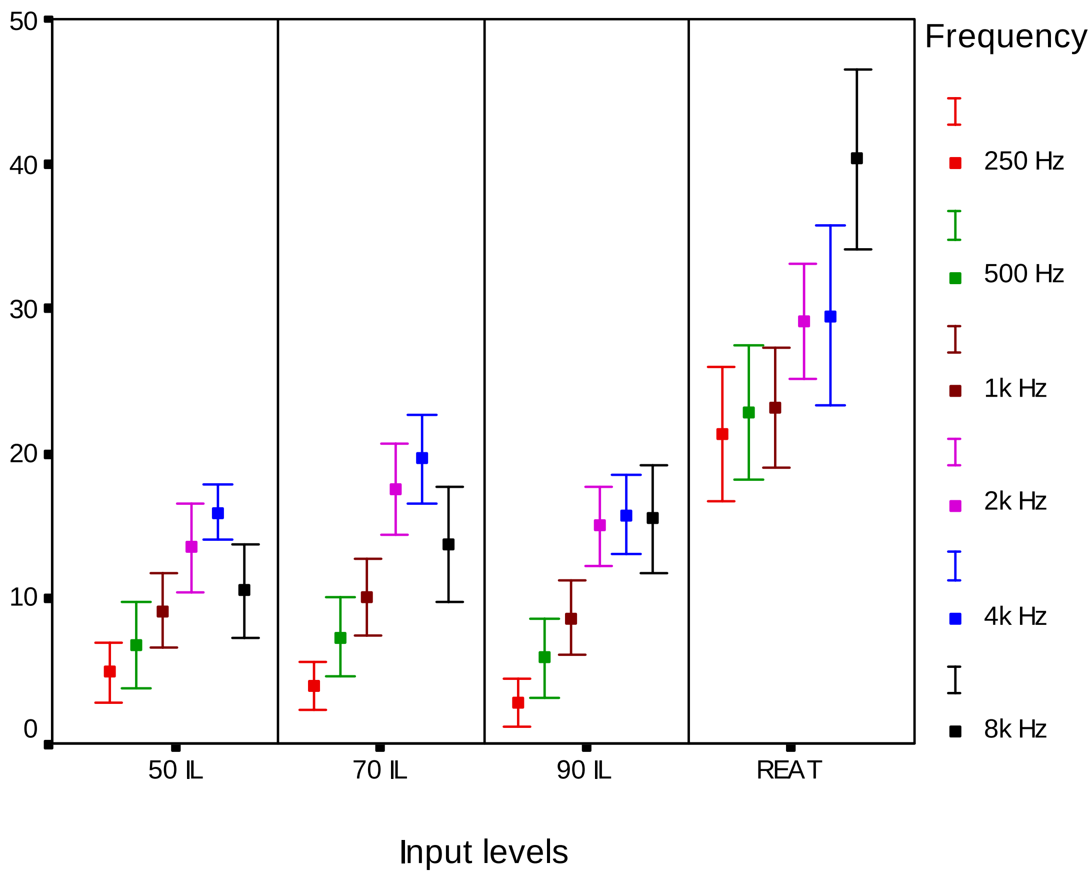

Figure 4.1: Graph shows error bars for comparison of REAT and IL at different input levels.

From the Figure 4.1, the error bars reveal that at 50, 70 and 90 dB SPL, IL values overlap and do not vary much. The REAT values are much higher.

Table 4.6: Shows the mean, SD and range of REAT and IL.

From the table 4.6, it can be observed that both REAT and IL increased with increase in frequency and the variability was more with REAT values.

Table 4.6: Combined mean, SD and range of REAT measure and mean, SD and range of IL.

Frequency	REAT (dB)			IL (dB)		
	Mean	SD	Range	Mean	SD	Range
200 Hz	21.33	12.45	0to 50	3.87	4.87	0to19.30
500 Hz	22.83	12.43	0to45	6.67	7.53	0.1to32.6
1 kHz	23.16	11.02	0to40	9.32	6.91	0.1to28.3
2 kHz	29.17	10.59	10to50	15.38	8.06	0.2to33.7
4 kHz	29.50	16.63	0to75	17.12	7.23	0.4to36.4
8 kHz	40.32	16.42	5to65	13.25	9.90	0.1to37

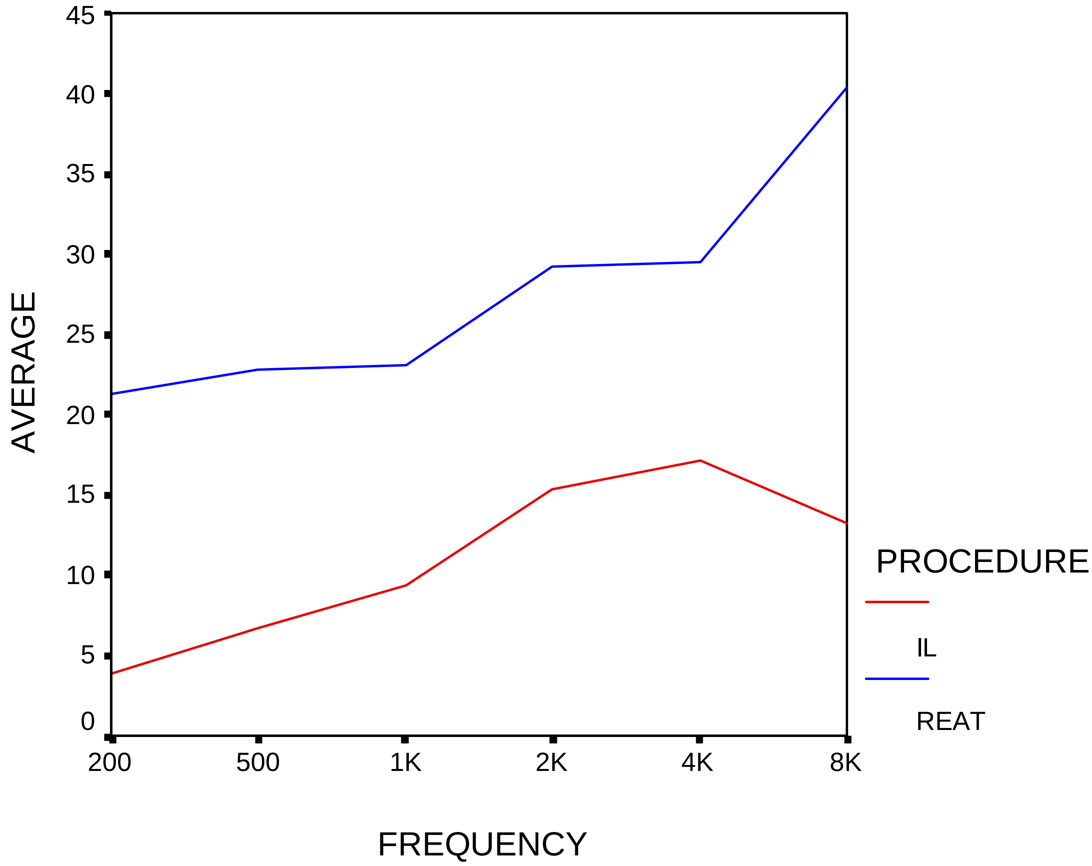

Figure 4.2: The graph showing REAT and IL for flanged earplugs. From the Figure 4.2, it is clear that attenuation values given by REAT procedure is higher compared to that of IL attenuation values. In the IL procedure, attenuation values reduced at higher frequencies, but in REAT attenuation values increased with frequency.

Comparison was done using independent sample t test between REAT and IL measure.

Table 4.7: Independent sample t test to compare REAT and IL at different frequencies.

Frequency	df	't' value	Mean difference (IL - REAT)	Level of significance
250 Hz	119	11.063	-17.45	0.000
500 Hz	119	8.528	-16.16	0.000
1 kHz	119	8.088	-13.84	0.000
2 kHz	119	7.468	-13.78	0.000
4 kHz	119	5.662	-12.37	0.000
8 kHz	119	10.338	-27.06	0.000

Table 4.7 shows comparison between REAT and IL measures at each frequency. There was highly significant difference between values obtained through REAT and IL procedures at all frequencies with REAT being higher than IL. This difference was more at 250 Hz, 500 Hz and 8 kHz ($P<0.001$).

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DISCUSSION

In the present study the objective was to compare earplug attenuation measured through Real Ear Attenuation at Threshold (REAT) with Insertion Loss (IL) procedures, effect of earplug on speech identification scores and also to study the linearity of insertion loss as levels of the signal increased.

The results indicated that there was a significant difference between REAT and IL measures at the tested frequencies, 250, 500, 1 k, 2 k, 4 k and 8 kHz. This finding agrees with that of the earlier studies carried out by Weinreb and Touger (1960), Villchur (1972), Berger and Kerivan (1983) and Traynor, Ackely and Wiebowsky (1984).

The difference between REAT and IL measures were more at low frequencies. Berger and Kerivan (1983) and Villchur (1972) suggested that the error was in the REAT procedure and that it was due to masking arising from physiological noise. The gap in the seal affected when a probe tube was inserted into the occluded ear canal also explains the difference. Traynor, Ackley and Wierbowsky (1989) have also explained that the difference between REAT and IL measures were due to the loss of the HPDs hermetic seal by the introduction of probe tube between the HPD and ear canal.

Another finding with IL measure is that the maximum attenuation was seen at frequencies 2 kHz and 4 kHz. Weinreb and Touger (1960) attributed this to the fact that at 2 kHz, bone conduction is most sensitive. Microphone in the ear canal will not sense the energy conducted via bone conduction pathway, and therefore, will measure a lower sound level hence a higher IL.

The results showed that there was a linearity of insertion loss for the input levels of 50, 70 and 90 dBSPL. Martin (1979) and Rood (1982) have also found similar results. This can be attributed to the fact that the noise levels investigated here were not sufficiently intense. That is, a noise level of 90 dBSPL was the maximum level investigated in this study. Also the effect of middle ear muscles is not taken into consideration in IL, which is one of the factors to induce non- linearity in attenuation of HPDs.

Another point that deserves mention is the generalization of the present findings. Although linearity of attenuation was confirmed for the flanged earplug in this study, it cannot be generalized for all the other types of hearing protective devices. Certainly the physical characteristics of the protector under evaluation can potentially influence the linearity of attenuation.

In the present study, speech identification scores were evaluated with and without earplugs in the ear canal and in four conditions, in quiet and in the presence of composite noise at 50, 70 and 90 dB SPL levels. Results showed that there was highly significant difference between protected and unprotected conditions in quiet and 50 dB level of noise. At high noise levels speech identification performance deteriorated in both occluded and unoccluded conditions. The variability in SIS scores among subjects is also higher at lower signal- to- noise ratio. This affects the communication at high noise levels.

The finding of the present study agrees with that of Chung and Gaunon (1979). Their results showed that at high signal- to- noise ratio (of +10), subjects with normal hearing obtained higher word recognition scores with the ear protectors. Conversely at a low S/N (-5), speech identification scores deteriorated with the use of hearing protective devices.

The results of the present study disagrees with that of earlier studies done by Kryter (1962), Berger (1980), Abel, Alberti, Haythomthwaite and Riko (1982) and Pekkarinen, Salmivalli and Suonpa (1990). They have reported better speech identification at high levels of noise when hearing protectors were used.

The results of the study imply that the attenuation measured by the two different procedures, i.e., REAT and IL procedures, differ significantly with REAT being higher than IL. Also there is linearity of IL that was noted. Further, the SIS scores reduced at high noise levels, in both unoccluded and occluded condition.

SUMMARY AND CONCLUSIONS

There are a number of methods for measuring attenuation of hearing protective devices. They are both objective and subjective methods. The popular method of evaluating attenuation characteristics of a hearing protective device is the absolute threshold shift technique or Real Ear Attenuation at Threshold (REAT). However, the HPDs are required to be used in high noise environments and not at thresholds. Measurement of attenuation at threshold may not give the same attenuation at high noise levels. There are studies which show both linearity and non-linearity of attenuation at high input levels. In this study, linearity of attenuation in Insertion Loss (IL) measure was studied by finding attenuation values at different presentation levels. Linearity of attenuation was seen for the IL measure in different presentation levels 50, 70 and 90 dBSPL. Since there was no significant difference between these levels, combined IL at different levels were taken for further comparison of IL with REAT.

Other main objectives of this study were to compare subjective procedure REAT with objective procedure IL measure. For this comparison, combined measure of IL (i.e., mean attenuation at all presentation levels 50, 70 and 90 dBSPLs for different frequencies) and REAT were taken. Results of the present study suggest that there was a significant difference between REAT and IL. REAT values were much higher than that of IL values. This can be attributed to the fact that REAT procedure is greatly affected by masking by the physiological noise and IL is affected by the hermetic seal due to insertion of probe tube for its measurement.

Speech identification performance of the subject's with and without earplug, in quiet and at high noise levels, was also studied. Speech identification scores showed no significant difference between both conditions, with and without the earplug at high levels of noise 70 and 90 dB. At high noise levels speech identification performance deteriorated in both occluded and unoccluded conditions. These results indicate that in the presence of high levels of noise speech identification is very poor with the use of earplug, thus may affect communication in such high noise environment.

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