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Insertion loss values of earplugs

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The purpose of this study was to investigate the feasibility of measuring the insertion loss (attenuation) provided by common earplugs using the Knowles Electronics Manikin for Acoustic Research. Five earplugs were tested in a 90 dB sound field at discrete frequencies matching those published by NIOSH. Each plug was exposed to 10 trials. Results indicate an approximation between values obtained in the study and NIOSH published values. Some consistent differences tend to indicate that NIOSH values may overrate low frequency attenuation.

INSERTION LOSS VALUES OF EARPLUGS

by

MICHAEL FAIRCHILD

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CHAPTER I

INTRODUCTION

Hearing loss resulting from work place noise has been reported since the beginning of industrialization (Temkin, 1973; Ward, 1979). Noise was one of many occupational health hazards that led to the passage of the Williams-Steiger Occupational Safety and Health Act of 1970 (Suter, 1979). The Williams-Steiger Act established, within the Department of Labor, the Occupational Safety and Health Administration (OSHA) and charged it with the duty of promulgating and enforcing rules relating to worker health and safety (Suter, 1979).

In 1974, OSHA published rules delineating allowable work place noise exposure and acceptable methods of noise control. Among these methods, though least preferred, is the use of personal hearing protection (Department of Labor, 1974). Personal hearing protectors have been characterized by Zwislocki (1958) as the most versatile, effective and economical means of noise control. Of the three major types of personal hearing protection, helmets, earmuffs and earplugs, earplugs are the most popular (Coles, 1969).

A number of methods have been proposed and used to assess the effect of earplugs and other personal hearing protectors, eventually culminating in the American Standards Association Standard ASA STD 1 - 1975 (ANSI 53 19 - 1974), "Method for the Measurement of Real-Ear Protection of Hearing Protection and Physical Attenuation of Ear-Muffs"

(American National Standards Institute, 1975). The standard assessment method, while unrealistically relying on threshold level intensities, was chosen because other methods had proven either unworkable or unreliable (Camp, 1979).

Unfortunately, attenuation characteristics measured at low intensity levels under ideal laboratory conditions may have limited application to real-life (work place) conditions. Recent progress in the development of artificial ears and acoustically accurate manikins, such as the Zwislocki artificial ear drum and the Knowles Electronics Manikin for Acoustic Research (Burkhard and Sachs, 1975) may have paved the way for further and more realistic research into the effectiveness of earplugs and other hearing protectors.

CHAPTER II

HISTORY OF THE PROBLEM

ACOUSTICS OF THE EAR

The outer ear (pinna) and canal (external auditory meatus) are acoustically reactive organs. The pinna consists of a thin plate of cartilage covered with skin and connected to the surrounding parts by ligaments and muscles (Goss, 1973). The deep central portion of the pinna, the concha, is 10 mm to 20 mm in diameter and leads to the 5 mm to 7 mm opening of the external auditory meatus (ear canal). The external auditory meatus is open to the concha portion of the pinna on one end and closed by the tympanic membrane on the other. The outer one-third of the external auditory meatus is composed of cartilage, the inner two-thirds of bone; both are covered with a thin layer of skin continuous with that of the pinna and tympanic membrane (Goss, 1973).

The acoustics of a tube, like the external auditory meatus, may be predicted, given the walls of the tube are rigid and the diameter is great enough to negate viscosity effects. When sound is transmitted into a tube, a node (area of minimal molecular movement) or an antinode (area of maximum molecular motion) occurs at any closing or opening of the tube respectively (Wood, 1966). In a tube open on one end and closed on the other, as is the external auditory meatus, a node occurs at the closed end and an antinode at the open end (see Figure 1). The simplest node-antinode configuration represents one-fourth of a wave length. The wave length of the fundamental resonant frequency is, therefore, four times the length of the tube (4L). The frequency of the fundamental may be expressed as f = C/4L where f = frequency in Hz and C = the speed of sound. While this formula is most applicable to tubal resonances, it is also relevant to the acoustic function of the human ear. The mean length of the adult ear canal is 25 mm, or one fourth the wavelength of a 3500 Hz tone (Helmholtz, 1954).

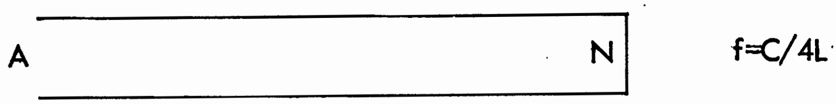

Figure 1. Node-antinode formation in a tube closed on one end.

A number of studies have measured actual ear resonances. Djupsland and Zwislocki (1973) found differing peak resonances at several points near the ear. At 1 cm from the entrance of the canal, they found a peak of about 8 dB at 7000 Hz; at the entrance of the canal, a 10 dB peak predictably at 3500 Hz; and at a point 1 cm from the tragus, a 15 dB peak at 3000 Hz with a secondary peak at 6000 Hz. In the latter, the peaks were nearly fused creating an increased sound pressure between 2000 Hz and 8000 Hz. In a study of the effects of azimuth, Shaw (1969) found the ratio of sound pressure level measured at the ear canal entrance and at the tympanic membrane to be substantially unaffected by variations in the acoustic field. Bekesy (1960) showed the ear to be an imperfect resonator due to acoustic damping from partial vibrational patterns of the tympanic membrane. As a result, ear resonances are neither as efficient in intensity nor as well defined in frequency as resonances from an equivalent hard walled coupler.

As described, the external auditory meatus may be viewed as a single open-end tube that selectively amplifies the acoustic signal. The external auditory meatus is also the final point at which an acoustic barrier may be placed in order to protect the inner ear from unwanted sound. Aural insert hearing protectors (earplugs), when placed in the meatus, provide varying amounts of protection depending on complex interactions between resonance and acoustic impedance.

AURAL INSERT HEARING PROTECTORS (EARPLUGS)

A wide variety of earplugs is available. The most common are made of premolded plastic and are furnished in several sizes (Swift, 1975). The best known of these is the single flanged V.51-R type, so named for its World War II designation (Coles, 1968). Other premolded plastic earplugs are most often multiflanged (tree shaped). Malleable earplugs are available that are shaped or molded into the ear (Swift, 1975). A recently introduced malleable earplug is made of polymer foam. When compressed and placed in the ear, the plug material attempts to regain its original shape, causing the plug to swell, thus closing the ear canal (Camp, 1979).

Earplugs both interrupt the natural resonance of the ear canal and introduce an acoustic impedance to air-born sound. The acoustic impedance of earplug material is composed of both resistance and reactance factors. Resistance is defined as the tendency of a material to absorb sound energy (Yost and Nielsen, 1977). Porous, sound absorbing materials, introduce many small paths to air-born sound creating, in the sound wave, a condition of high viscosity (Cheever, 1975). Reactance is defined as the tendency of a material to reduce the ability of a sound wave to oscillate at maximum efficiency (Yost and Nielsen, 1977). In general, reactance is dependent on the density and stiffness of the material (Cheever, 1975).

Sound passes through the plugged meatus reaching the inner ear, by any of three paths. The plug may vibrate as a whole creating a piston effect in the cavity between the plug and tympanic membrane; it may be deformed and vibrate in parts; and/or there may exist non-ear paths such as bone and eustachian tube conduction. Because of the alternate paths available to sound, Zwislocki (1958) determined the maximum attenuation available from an ideal plug to be about 25 dB at low frequencies and 55 dB at high frequencies (see Figure 2).

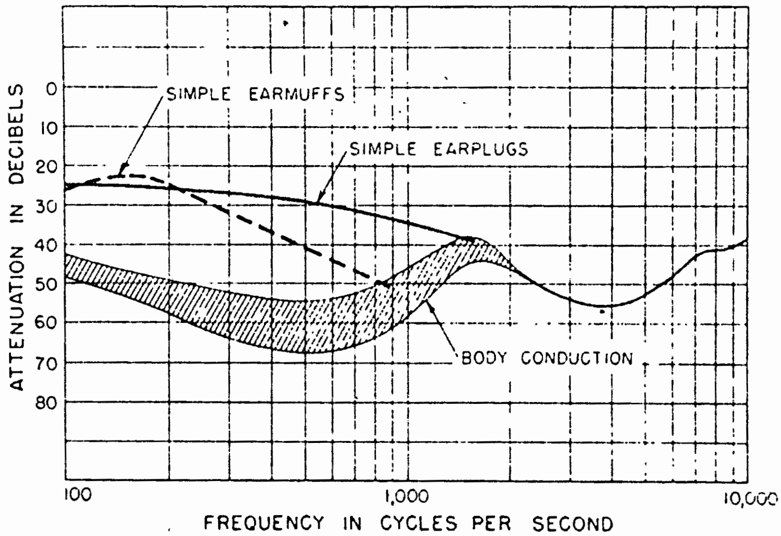

Figure 2. Theoretical maximum attenuation available from personal hearing protectors (Zwislocki, 1958).

Since the attenuation provided by earplugs may not match Zwislocki's ideal all earplugs must be tested. Numerous methods have been proposed and utilized in the study of earplug attenuation.

HEARING PROTECTOR RESEARCH

The intensity of the test stimuli used in attenuation studies ranges from low, near the threshold of hearing, to high, exceeding typical industrial noise levels. In general, the use of low sound pressure levels result in greater attenuation values than do the use of higher sound pressure levels (Hershkowitz and Levine, 1957; Weinreb and Touger, 1960; Nixon, Sommer and Cashin, 1963). Thus, an earplug may fully attenuate a threshold level sound of 10 dB but provide less attenuation when exposed to a superthreshold level sound of 60 dB. Both pure tones and narrow bands of noise are among the signals used to study the attenuation provided by hearing protectors. Webster, Thompson and Beitscher (1956), in comparing attenuation values taken using pure tones and narrow bands of noise, found the types of stimuli resulted in about equal means. Waugh (1974) found the use of narrow band noise stimuli resulted in less variation in attenuation values than did the use of pure tones.

Attenuation studies may be categorized into subjective studies (those requiring some decision by a subject) and objective (those requiring no subject participation). Most published attenuation values are taken using a subjective method called the threshold shift procedure. In the threshold shift procedure, the subject's hearing thresholds are taken both with the ear unoccluded, resulting in some threshold value X dB, and with the ear occluded, resulting in some threshold value (X + Y) dB (see Figure 3). The increased sound pressure level required to achieve threshold while the ear is occluded

(Y dB) is assumed to represent the attenuation provided by the hearing protector (Michael and Bolka, 1971).

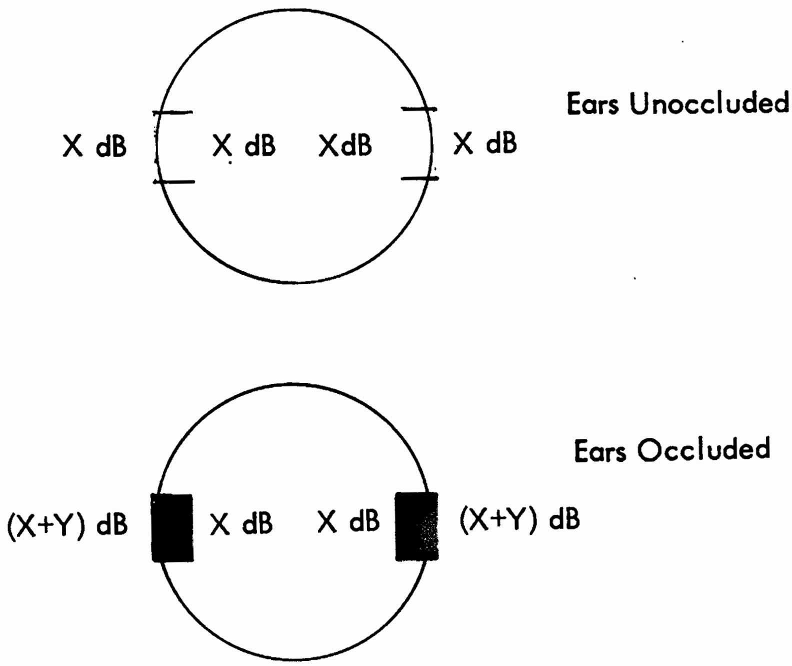

Figure 3. The threshold shift procedure of testing attenuation. The increased sound pressure level needed to produce X dB behind the earplug (Y dB) represents the attenuation provided by the earplug.

Because it is a comparison of threshold measures, the above method relies on a low sound pressure level signal with the accompanying overestimation of attenuation values. In an attempt to provide a more realistic measure, Hershkowitz and Levine (1957) used a loudness balance procedure, a subjective method using high sound pressure level signals. Their subjects were asked to compare the loudness of a signal presented to one (the reference) ear by earphone with the loudness of the same signal presented to the other (test) ear from a speaker a short distance away. Loudness comparisons were made with the test ear both occluded and unoccluded. The differences in matched loudness values between the occluded and unoccluded conditions were thought to represent the attenuation value of the hearing protector (see Figure 4).

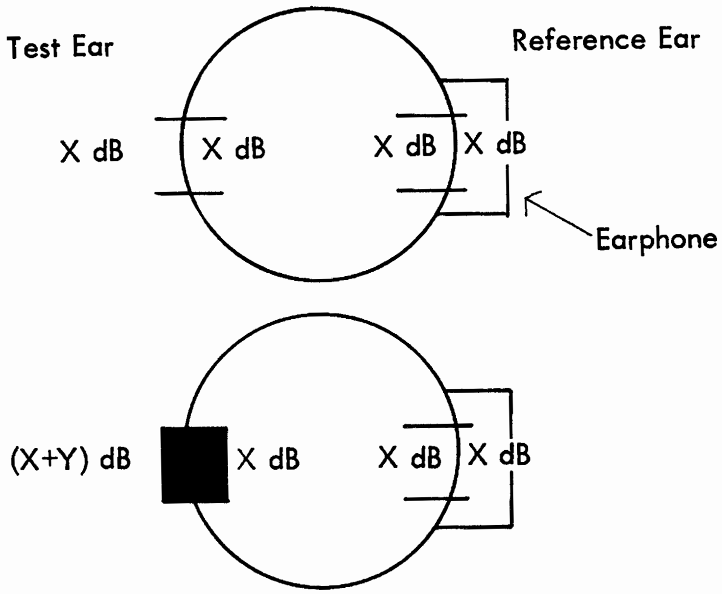

Figure 4. The loudness balance procedure of testing attenuation. The increased sound pressure level needed to produce X dB behind the earplug, causing the subject to perceive equal loudness between ears (Y dB), represents the attenuation provided by the earplug.

Fletcher and Loeb (1962) attempted to measure attenuation using temporary threshold shift; a temporary hearing loss resulting from exposure to intense sound (Ward, 1973). Subjects' hearing thresholds were obtained before and after exposure to measured amounts of sound, over a constant time period, in order to determine the minimum sound pressure level necessary to produce a temporary threshold shift. The differences in minimum sound pressure level necessary to produce temporary threshold shift through protected and unprotected ears were then compared (Figure 5). These differences were assumed to represent

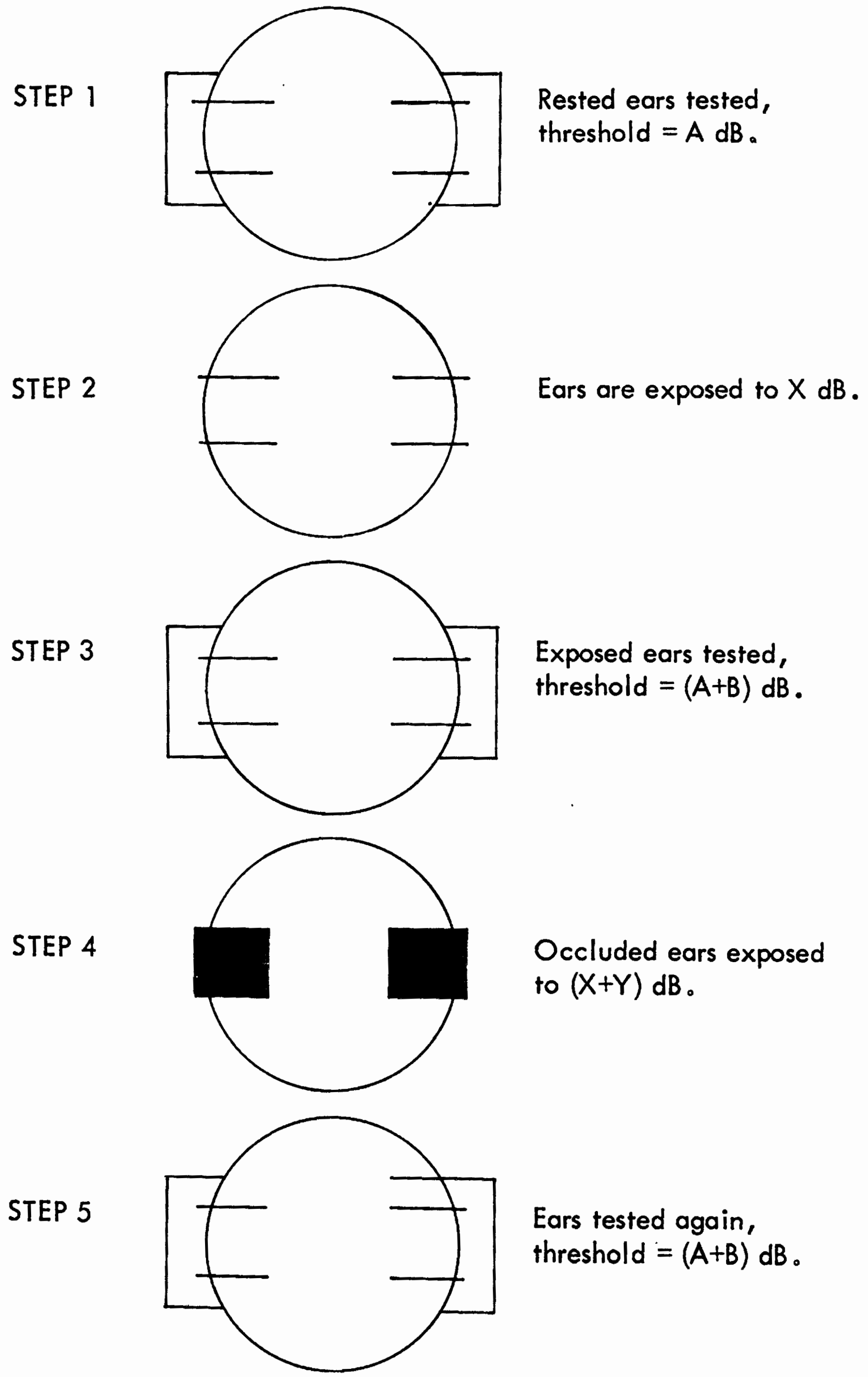

Figure 5. The temporary threshold shift procedure of testing attenuation. The increased sound pressure level (Y dB) required to produce a temporary threshold shift (B dB) represents the attenuation provided by the earplug.

an objective measure of hearing protector attenuation.

Nixon, Sommer and Cashin (1963) used the acoustic reflex, an objective response, to measure attenuation. The stimulus speaker was placed two inches from the test ear. A muff containing an impedance bridge (an instrument capable of detecting the slight movement of the ear drum caused by the stapedius reflex) covered the other ear. Acoustic reflex thresholds were obtained for the test ear, both while open and occluded (see Figure 6). The result, a high sound pressure level objective measure of attenuation, varied considerably from subject to subject.

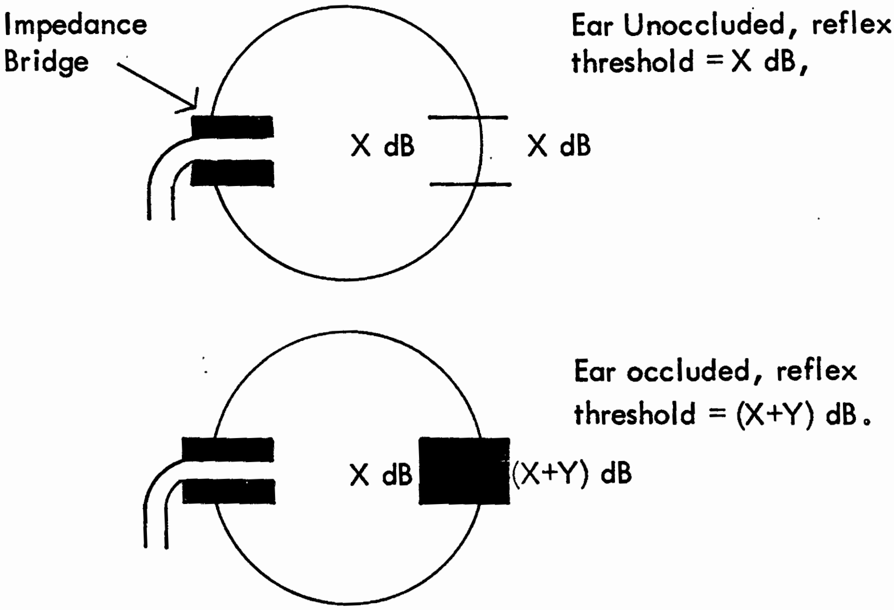

Figure 6. The acoustic reflex procedure of testing attenuation. The increased sound pressure level required to elicit the acoustic reflex (Y dB) represents the attenuation provided by the earplug.

Coles (1968) noted a number of attempts to measure attenuation objectively using no physiological response. Among the methods he listed were the use of artificial heads and ears and cadaver heads and ears.

Weinreb and Touger (1960) inserted a probe microphone into the ear canal of a live subject, between the hearing protector and the tympanic membrane. Attenuation was considered equal to the difference between sound pressure levels outside and inside the ear canal. Burkhard (1976c) and Marraccini and Burkt (1977) used a Knowles Electronics Manikin for Acoustic Research (KEMAR). Attenuation was defined as the difference in sound pressure levels measured at the ear of the KEMAR with and without the protection in place.

KNOWLES ELECTRONICS MANIKIN FOR ACOUSTIC RESEARCH (KEMAR) AND THE ZWISLOCKI ARTIFICIAL EAR DRUM

The KEMAR is made of one quarter inch fiberglass-reinforced polyester. The interior is coated with lead pellet-filled resin to provide mass and reduce acoustic coupling. The pinnae are soft tear-resistant appendages that snap into recesses in the head. The head and torso dimensions of the KEMAR are averages taken from several anthropometric studies. Particular attention was given to those dimensions thought to be acoustically important. The dimensions of the pinnae are means of 24 subjects, 12 female and 12 male. The KEMAR may be opened from the top of the head and back of the torso. The neck is hollow to allow passage of instrumentation. The head may be removed from the torso. The KEMAR system may be used with the head and torso, the head only or the ear and coupler only (Burkahrd and Sachs, 1976).

The Zwislocki artificial ear drum or coupler provides an acoustic load like that of the tympanic membrane (Helle, 1976). The coupler consists of a one half inch diameter chamber threaded to accept a one half inch microphone on one end. Near the microphone are four chambers branching at right angles. Each chamber provides acoustic resistance at different frequency bands (Burkhard, 1976a). The Zwislocki coupler was altered for use with the KEMAR. More robust materials were used inside the resistance chambers and the coupler was threaded to fit onto the KEMAR ear canal. The diameter and length of the KEMAR ear canal, including the Zwislocki coupler are 7.5 mm and 21.5 mm respectively. The length of the KEMAR ear canal is shorter than the mean human measurement for two reasons: (1) The velocity of sound is less at room temperatures than at body temperature. The KEMAR, having no internal heating mechanism, operates at room temperature while the human ear is warmed by body heat. The shorter KEMAR canal compensates for the disparity in velocity. (2) Microphone compliance adds effective length (Burkhard and Sachs, 1976). The shorter ear canal and microphone compliance produce resonance values similar to those of the human ear canal.

Burkhard (1976b) proposed the terms "insertion gain" and "insertion loss" for the acoustic effects of devices when measured on the KEMAR. Two methods of measuring insertion gain and loss have been reported. In the substitution method, the signal is measured through the unimpeded ear, then through the impeded ear, with the device in place. In a variation of this procedure, the KEMAR is removed and the microphone is placed in the head center position to take the unimpeded readings. The comparison method utilizes the symmetry of the sound field. An unimpeded microphone is placed either outside the KEMAR or in the KEMAR's unimpeded ear while the primary microphone reads the signal through the impeded test ear. The experimenter takes sound level readings at different times when using the substitution method and simultaneously when using the comparison method (Beck, 1976).

While procedures have been developed and tested for the use of the KEMAR in evaluating hearing aids and other acoustic devices, there remains a paucity of information concerning the use of the manikin to study hearing protection.

PURPOSE

The purpose of the study was two-fold: First to establish procedures for the testing of earplug attenuation on the KEMAR; and, second, to obtain sample attenuation (insertion loss) values for selected types of earplugs.

The questions the study sought to answer were:

1. Are earplug insertion loss studies feasible using the KEMAR?
2. How do insertion loss values obtained using the KEMAR relate to values obtained using other methods and instruments?

CHAPTER III

METHOD

PROCEDURE

An anthromorphic acoustic manikin was placed in a double wall, sound treated room 30 inches from, and facing (0° azimuth) a sound field speaker system. Distance and angle from the sound field speaker system were established by direct measurement.

The test signal was generated in a clinical audiometer and fed through a variable filter in order to obtain a narrow band of noise such that the lower frequency cutoff or half power point was located at a frequency of less than 0.75 times the center frequency, and the upper frequency half power cutoff was defined by at least 1.25 times the center frequency. The filters provided an attenuation of at least 30 dB per octave for the lower skirts and at least 29 dB per octave for the upper filter skirts (see Figure 7). This filtered noise test signal was channeled through a linear amplifier and into the sound treated room where it was converted from an electrical to an acoustic signal by a sound field speaker system.

Three equipment systems were used to measure the sound pressure level of the signal: (1) The sound field coincident system was used to measure the signal sound pressure level at a point coincident with the manikin's head center position. (2) The reference system was used to measure the signal sound pressure level at a point 12 inches from

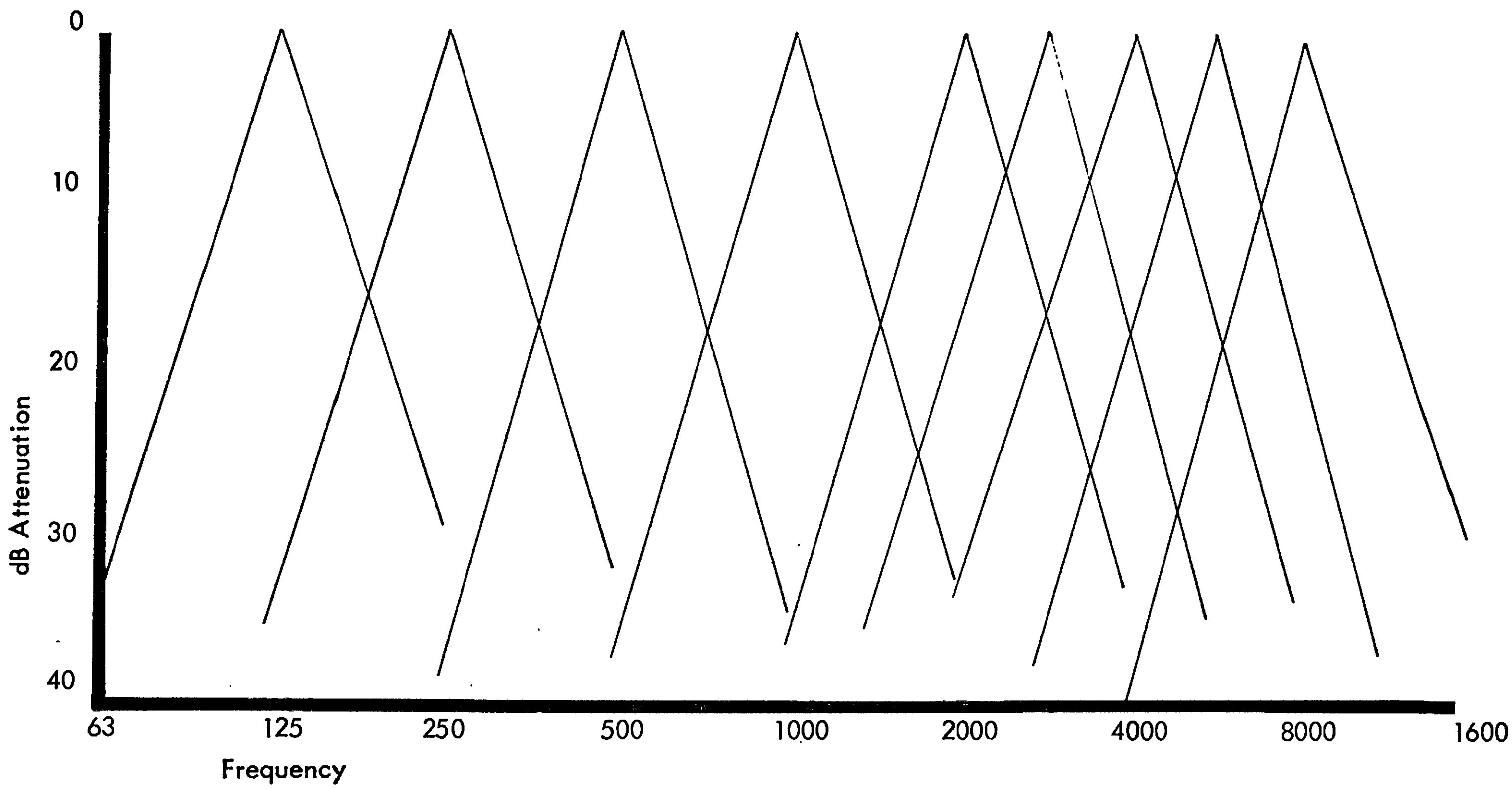

Figure 7. Octave band frequency response of the experimental narrow band noise signal measured at the output of the filter.

the face of the sound field speaker system. (3) The test system was used to measure the signal sound pressure level through the manikin's left ear.

A fourth system, used to check the sound field frequency response, consisted of a one half inch sound field microphone (fixed at a 90° azimuth from the speaker system) attached to an input stage, flexible cable, impulse precision sound level meter and one third octave filter set.

The sound field coincident system consisted of a one half inch sound field microphone attached to an input stage, flexible cable, impulse precision sound level meter and one third octave filter set. The microphone was attached five inches above the neck of the headless manikin torso at a 90° azimuth relative to the sound field speaker system (see Figure 8).

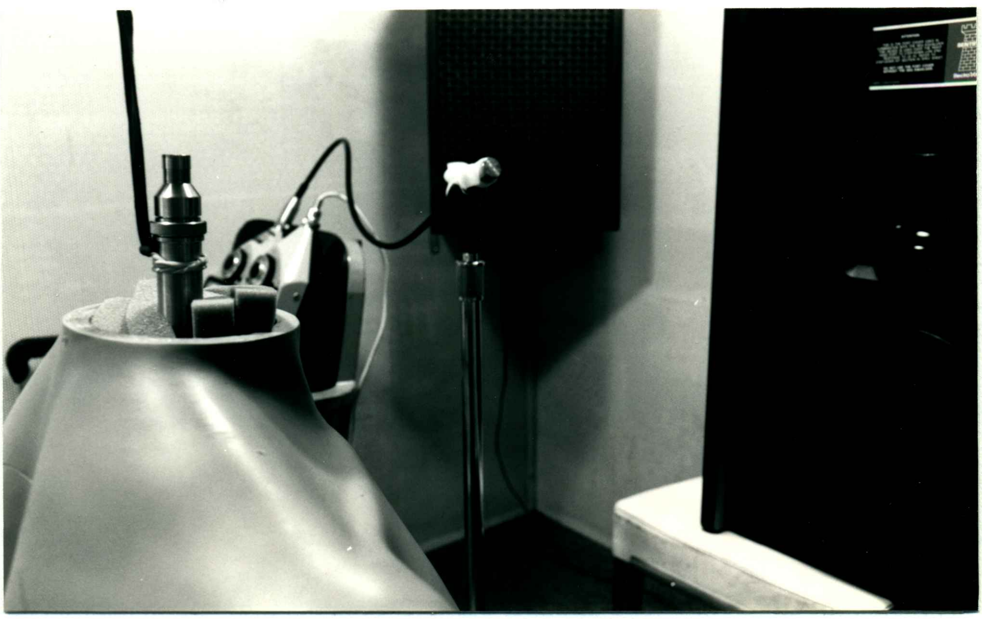

Figure 8. The sound field system used to measure signal sound pressure level.

The reference system incorporated a one inch sound field microphone, flexible microphone cable, precision sound level meter and octave filter set. The reference microphone was angled at 90° relative to the sound field speaker system (see Figure 9).

Figure 9. The reference system microphone in place

The test system included a Knowles Electronics Manikin for Acoustic Research (KEMAR) pinna and ear canal, a Zwislocki artificial ear drum or coupler, a flexible cable, an impulse precision sound level meter and a one third octave filter set. The latter four components were shared alternately with the sound field coincident system as depicted in Figure 10.

The earplug insertion system consisted of a clinical impedance audiometer with the bridge probe inserted into the back of the Zwislocki coupler and sealed with a one half inch foam plug (see Figure 11).

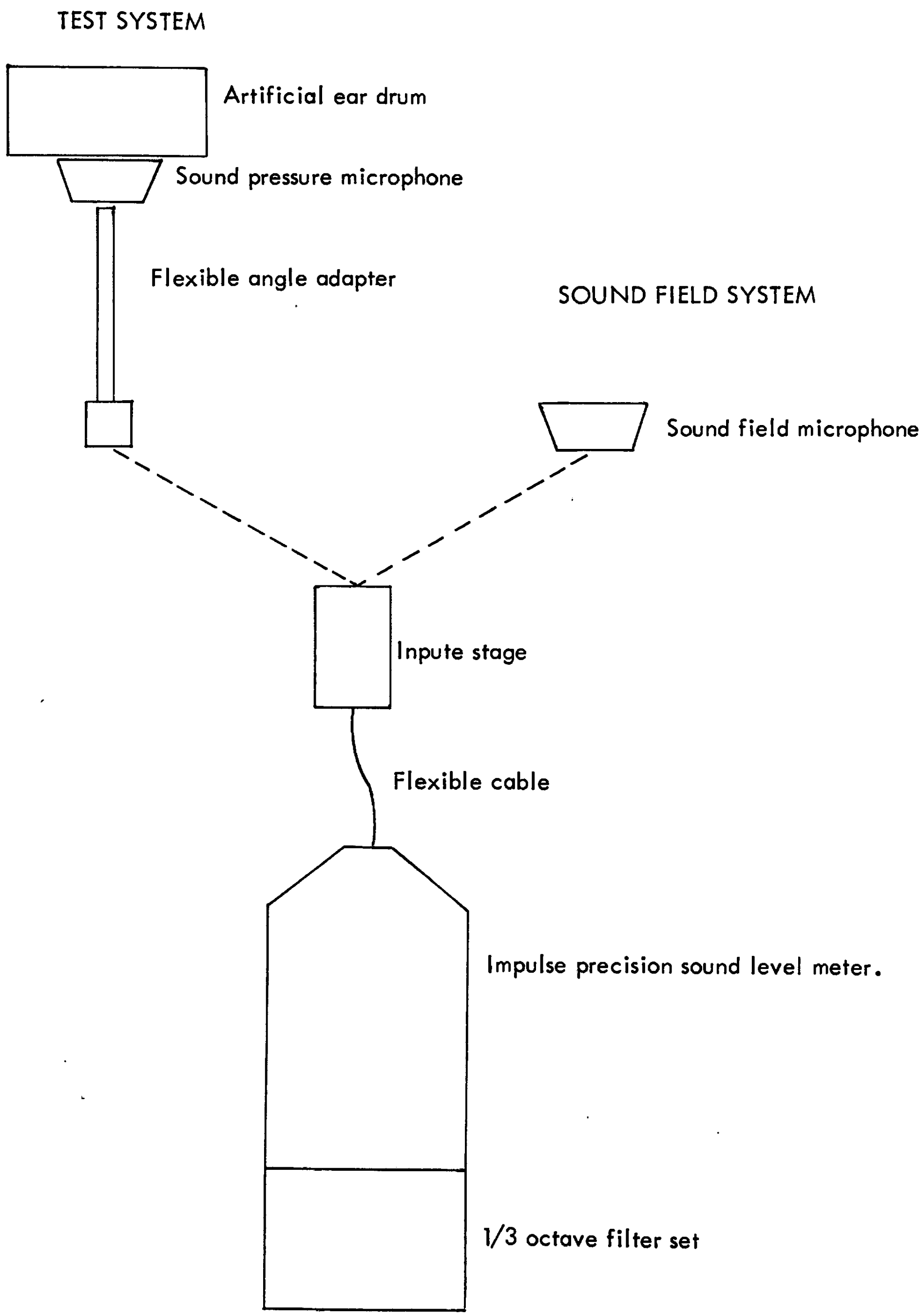

Figure 10. The experimental test system and coincident sound field system showing shared components (dotted line).

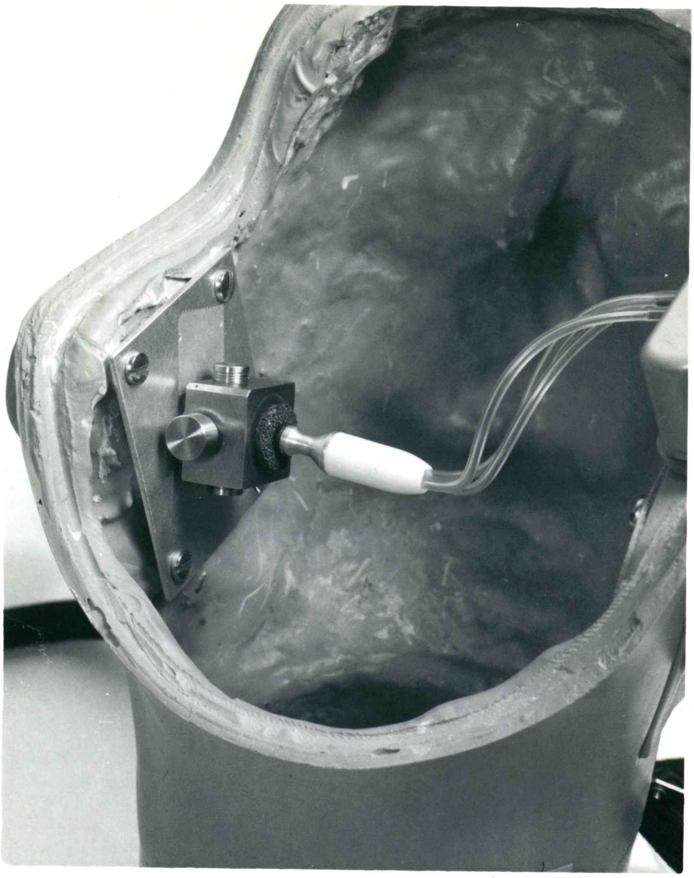

Figure 11. The earplug insertion system

The experimental signals consisted of narrow band noise, as previously described, with center frequencies of 125 Hz, 250 Hz, 500 Hz, 1000 Hz, 2000 Hz, 3000 Hz, 4000 Hz, 6000 Hz and 8000 Hz. All signals were presented sound field at 90 dB sound pressure level. All sound pressure level measures were made utilizing the appropriate filter settings in order to accomodate the center frequency of the experimental signal.

There were ten experimental sessions. The procedure for each experimental session was identical and consisted of three steps:

(1) The sound field sound pressure level was adjusted to 90 dB sound pressure level using the coincident measurement system. The sound pressure level measured at the reference system was noted and recorded. The process was repeated for each of the nine experimental signals. (2) The sound pressure level of the sound field experimental signal was measured through the unoccluded left ear of the manikin using the test system. The sound pressure level at the reference microphone was recorded prior to and after each experimental measurement. Step two was repeated for each experimental frequency. (3) An earplug was inserted into the manikin left ear using the earplug insertion system. The earplug was adjusted to produce a reading from the impedance meter of 1.70 cc to 1.75 cc equivalent volume. Five sound pressure level readings were made and recorded through the occluded ear for each of the nine experimental signals. Each reading was preceded and followed by a reading from the reference system. Step three was repeated for each of the five test ear plugs (see Figure 12).

CALIBRATION

Electronic calibration of the frequency response of the test signal was done by routing the signal from the variable filter through an insert voltage adapter to a precision sound level meter and one-third octave filter set. One-third octave readings of the signal, from 63 Hz through 16000 Hz were made.

Checks of the acoustic calibration of the frequency response were accomplished using the sound field system previously described. Acoustic measures were taken at one third octave intervals from 63 Hz through 16000 Hz.

Acoustic calibration of the microphones and sound level measuring systems was done using a precision piston phone with adaptors to fit the various sizes of microphones.

Calibration of the impedance audiometer used to control earplug insertion was done using a built-in 5 cc hard wall coupler.

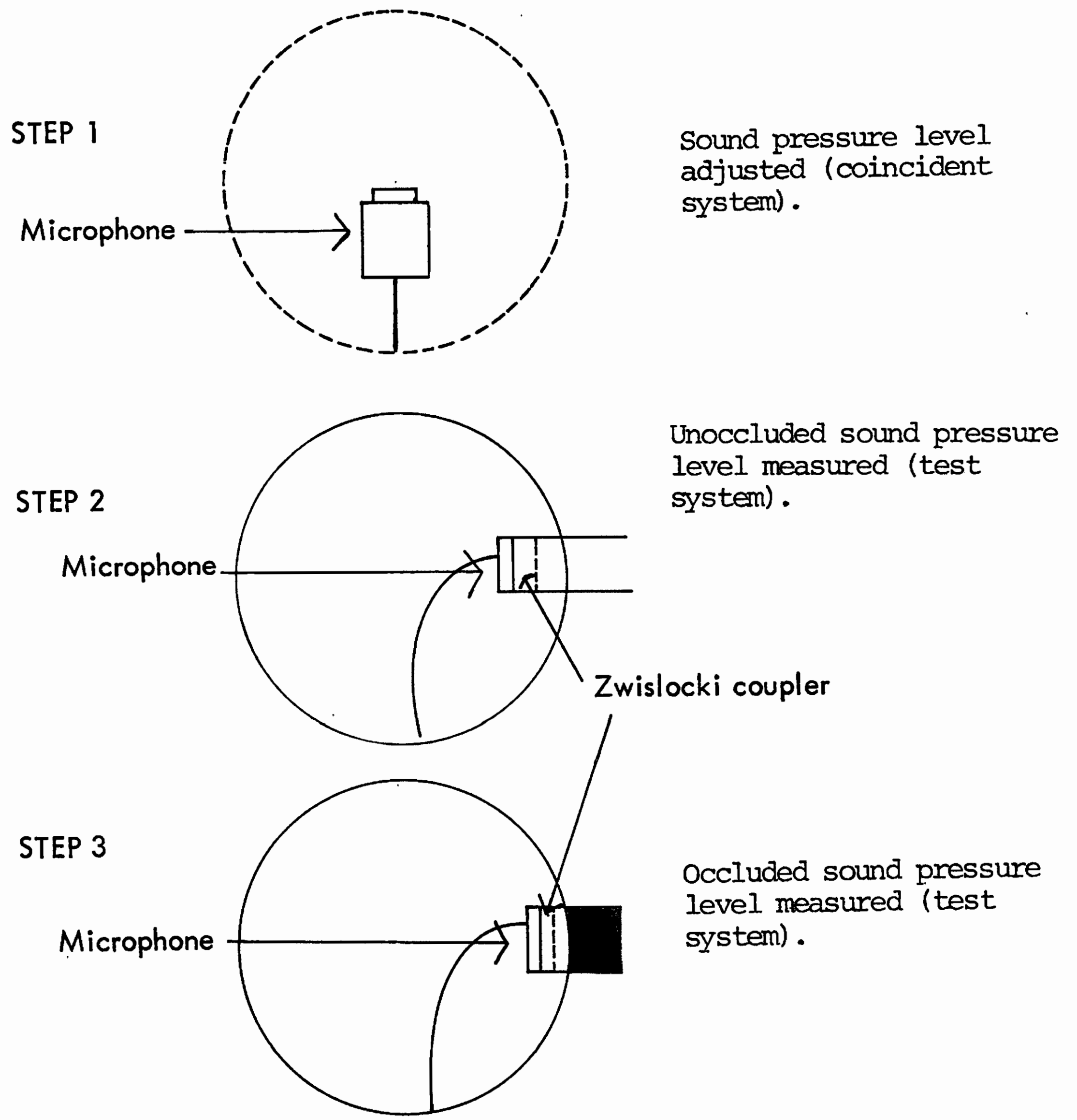

Figure 12. Diagram of an experimental session.

Electronic calibration checks were completed before and after the study. Acoustic calibration checks were accomplished before and after each experimental session. Calibration of the impedance audiometer was accomplished before each earplug fitting.

INSTRUMENTATION

The study was conducted in an acoustically treated room (International Acoustics Corporation, Model 1403) located at Portland State University.

Signal generating equipment consisted of an audiometer (Maico, Model 24-B) which was channeled through a variable frequency filter (Krohn-Hite, Model 32R) then through an amplifier (McKintosh, Model 50) and finally to a sound field speaker system (Electrovoice, Model Century).

Test sound pressure level measuring equipment included an artificial ear drum (Knowles adaption of the Zwislocki Artificial Ear Drum), a one half inch pressure microphone (Bruel and Kjaer, Model 4136), a flexible adapter (Bruel and Kjaer, Model UA 0122), an input stage (Bruel and Kjaer, Model ZCO 007), a low impedance shielded cable (Bruel and Kjaer, Model AO 0027), an impulse precision sound level meter (Bruel and Kjaer, Model 2209) and a one-third octave filter set (Bruel and Kjaer, Model 1616).

Sound field sound pressure level measuring equipment included a one half inch sound field microphone (Bruel and Kjaer, Model 4133), an input stage (Bruel and Kjaer, Model ZCO 007), a low impedance shielded cable (Bruel and Kjaer, Model AO 0027), an impulse precision sound level meter (Bruel and Kjaer, Model 2209) and a one third octave filter set (Bruel and Kjaer, Model 1616).

Reference sound level measuring equipment included a one inch sound field microphone (Bruel and Kjaer, Model 4144), a microphone extension cable (Bruel and Kjaer, Model AO 0059), a precision sound level meter (Bruel and Kjaer, Model 2203) and an octave filter set (Bruel and Kjaer, Model 1613).

Earplug insertion equipment included an impedance audiometer (Teledyne, Model TA3D), and a sponge rubber one half inch ear insert (Teledyne).

Calibration equipment included an insert voltage adapter (Bruel and Kjaer, Model UAO 322), a precision sound level meter (Bruel and Kjaer, Model 2203) and a one-third octave filter set (Bruel and Kjaer, Model 1616). A precision piston phone (Bruel and Kjaer, Model 4220) was additionally included.

Earplugs used in the study (see Figure 13) were a moldable type (Marion, Model Silent Partner), a polymer type (E.A.R. Corporation, Model EAR), a single flanged type (Curtis Safety Products, Model V.51-R), a multiflanged type with heavy flanges (Wilson Products, Model Sound Silencer) and a multiflanged disposable type with relatively light flanges (Three M Company, Model 3-M).

DATA ANALYSIS

The study resulted in 2700 experimental readings, each interposed between two reference readings. The pre-and-post-reference readings were averaged and the average compared to a baseline reference reading noted earlier when the sound pressure level was established using the coincident sound field system. If the averaged differences varied more than $\pm$ 0.5 dB from baseline, the experimental reading was adjusted by the amount of the variation. Five adjusted readings from a single plug/frequency condition within a session were averaged. The averaged value of each of the pluged conditions was then subtracted from the average of the open condition, which had been corrected and averaged in the same manner as the pluged condition. The result was the insertion loss provided by an earplug at one frequency during one session. These insertion loss measurements were the raw data of the study (see Appendices).

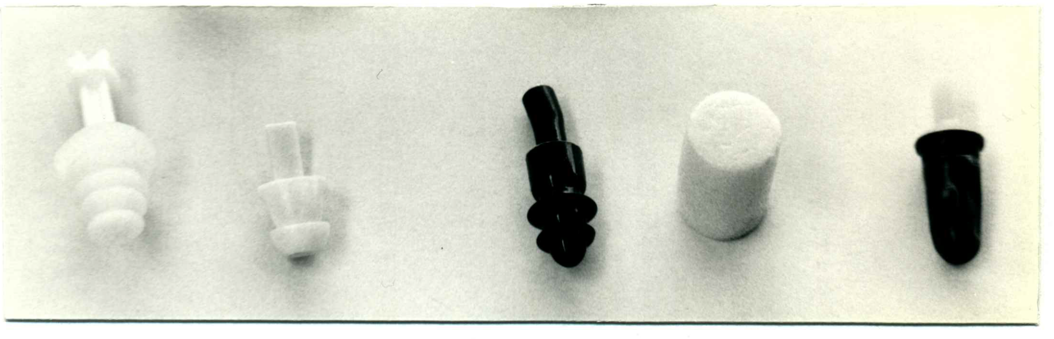

Figure 13. Earplugs used in the study. Left to right; Multiflanged disposable, single flanged, multiflanged, polymer foam and moldable.

CHAPTER IV

RESULTS

An anthromorphic acoustic manikin (KEMAR) was used to evaluate the protection provided by five common types of aural insert hearing protectors (earplugs). Protection was defined as the average insertion loss resulting from an earplug being placed in the manikin's ear canal. That is, a comparison of the sound pressure levels measured through the manikin ear while unoccluded and again occluded by the earplug.

Ninety dB narrow band noise signals, with center frequencies of 125 Hz, 250 Hz, 500 Hz, 1000 Hz, 2000 Hz, 3000 Hz, 4000 Hz, 6000 Hz and 8000 Hz were used in the study. Center frequencies were chosen to correspond to those reported by the National Institute of Occupational Safety and Health or NIOSH (Kroes, Fleming and Lempert, 1975). The minimum band rejection of the filter skirts was 27 dB per octave.

The intensity of the experimental signal was established at 90 dB at a point coincident with the head center position of the manikin. Insertion of the manikin head and ear into the sound field, with measurement taken through the manikin ear, resulted in a frequency response much like that of the human ear (see Figure 14). The manikin head and torso were placed at 0° azimuth, or facing the sound source.

The earplugs were inserted into the manikin ear canal only far enough to insure mechanical stability. Insertion depth was controlled by using a clinical impedance audiometer, inserting each plug to

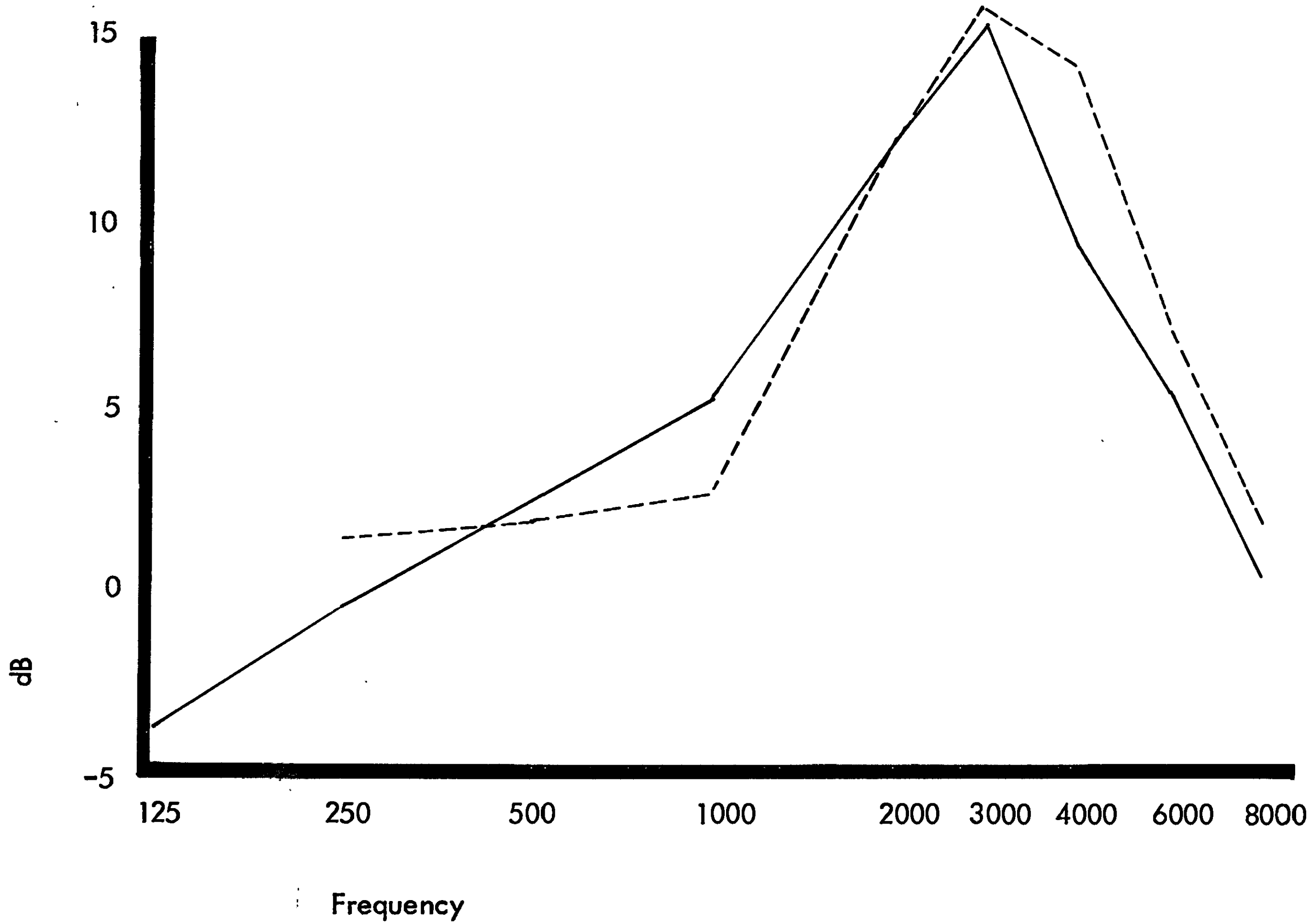

Figure 14. Frequency response of the left ear of the manikin, solid line, with comparative data from human subject research, broken line (Shaw, 1974).

achieve identical equivalent volumes in the manikin ear canal. The result was a highly repeatable measure (± 2mm) of insertion depth. Use of the impedance meter also insured against the possibility of air leakage or consequent break in the acoustic seal.

The mean insertion loss values for the five earplugs studied are illustrated in Figures 15 - 19.

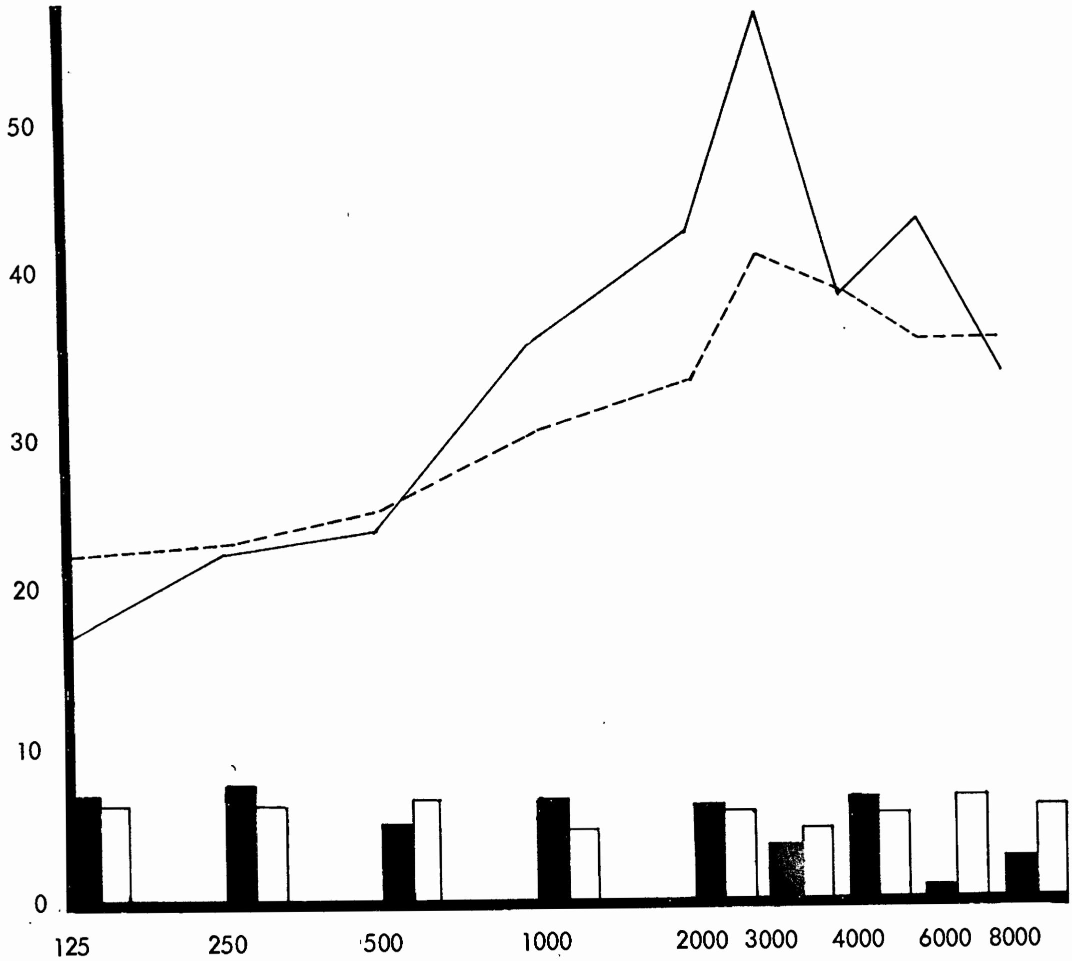

Figure 15. Means (solid line) and standard deviations (solid bars) of insertion losses resulting from placement of the single flanged V.51-R earplug in the manikin left ear. Comparative means and standard deviations (Broken lines and open bars) of attenuation values of the same earplug reported for NIOSH by Kroes, et. al. (1975).

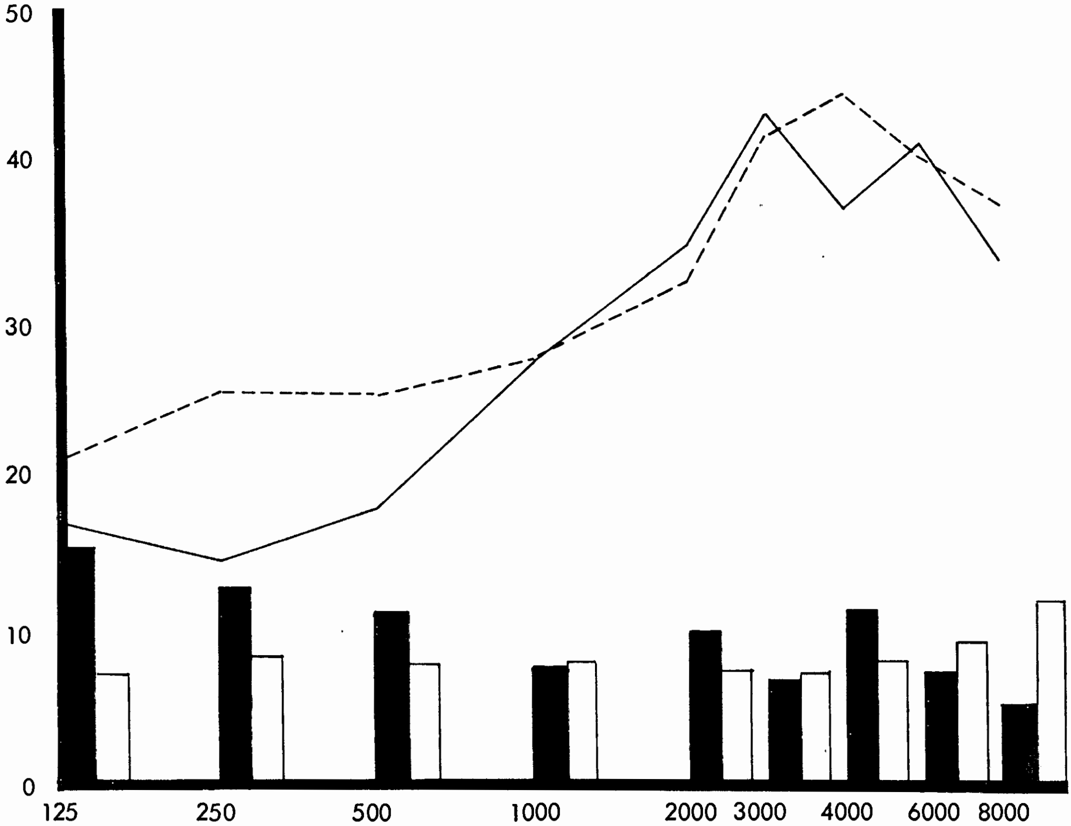

Figure 16. Means (solid line) and standard deviations (solid bars) of insertion loss resulting from placement of the multi-flanged earplug in the manikin left ear. Comparative means and standard deviations (broken lines and open bars) of attenuation values of the same earplug reported for NIOSH by Kroes, et. al. (1975).

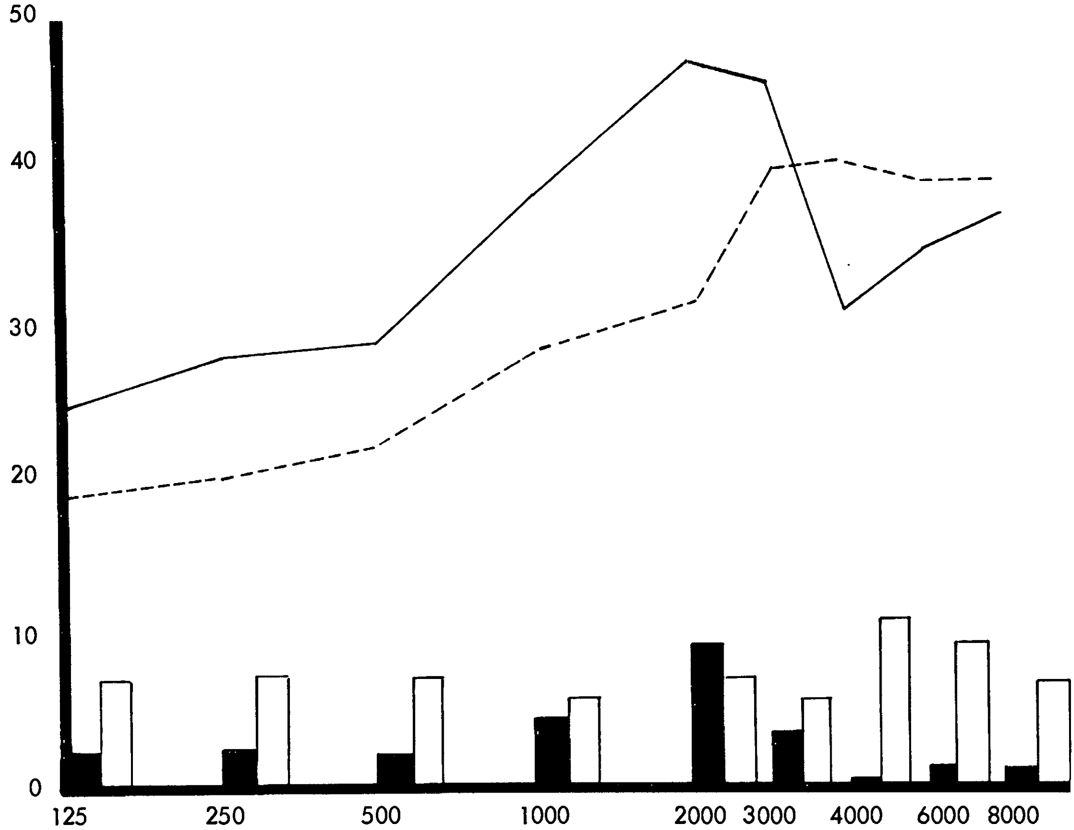

Figure 17. Means (solid line) and standard deviations (solid bars) of insertion loss resulting from placement of the multi-flanged disposable earplug in the manikin left ear. Comparative means and standard deviations (broken line and open bars) of attenuation values of the same earplug reported for NIOSH by Kroes, et. al. (1975).

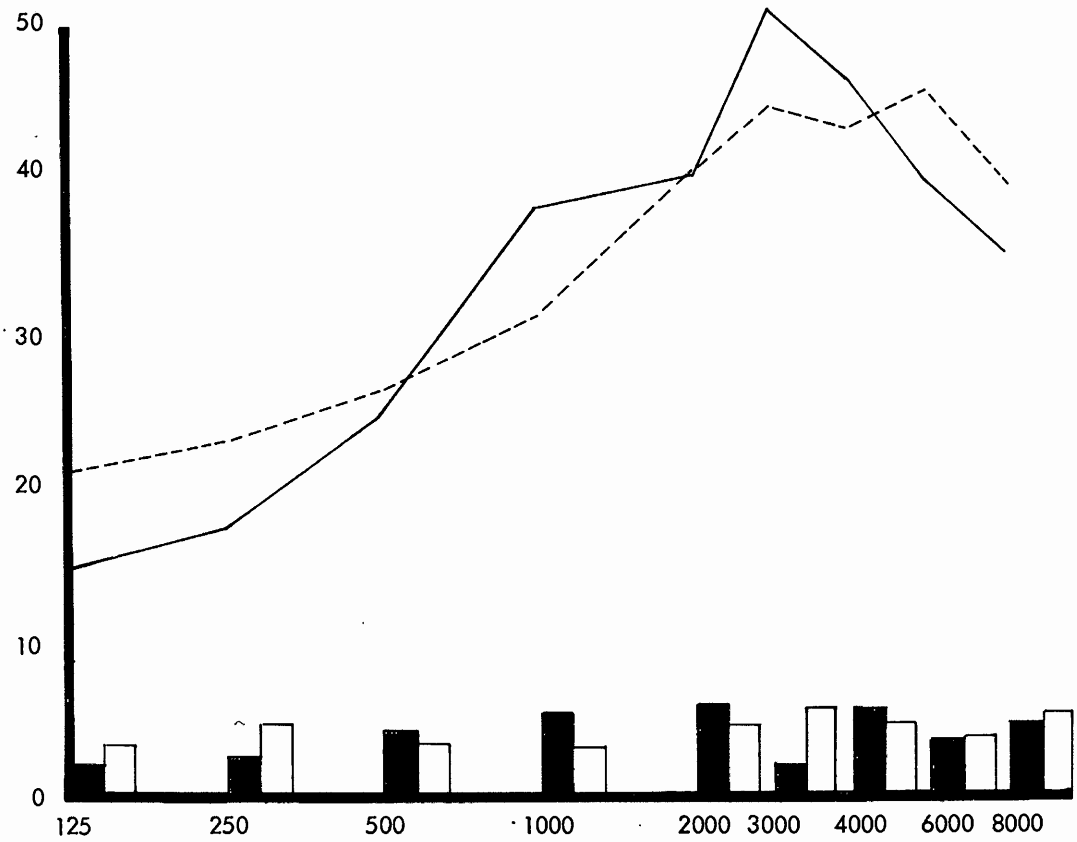

Figure 18. Means (solid line) and standard deviations (solid bars) of insertion loss resulting from placement of the moldable earplug in the manikin left ear. Comparative means and standard deviations (broken line and open bars) of attenuation values of the same earplug reported for NIOSH by Kroes, et. al. (1975).

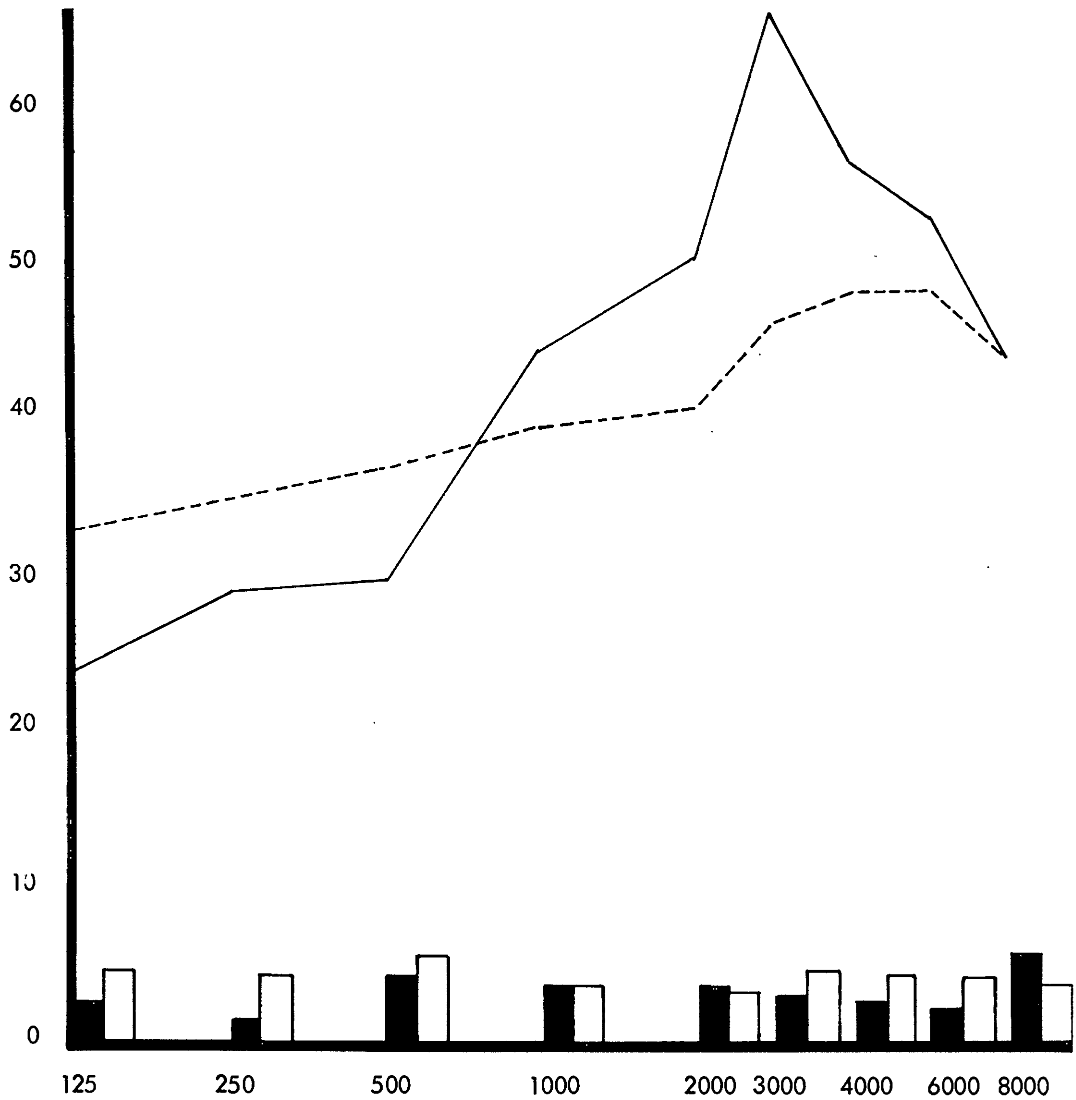

Figure 19. Means (solid line) and standard deviations (solid bars) of insertion loss resulting from placement of the polymer earplug in the manikin left ear. Comparative means and standard deviations (broken line and open bars) of attenuation values of the same earplug reported for NIOSH by Kroes, et. al. (1975).

The values reported in figures 15 - 19 are ten-trial means. The range of observed insertion loss values of ten trials for earplug and frequency is reported in Table I. Ranges are also reflected in the standard deviation values which are shown in Figures 15 - 19.

TABLE I

RANGES OF INSERTION LOSS MEASURED ON FIVE EARPLUGS

EARPLUGS	FREQUENCY IN KHz								
	.125	.250	.500	1.0	2.0	3.0	4.0	6.0	8.0
V.51-R	25.2	27.3	17.1	18.0	21.1	12.1	21.9	3.4	6.7
Multiflanged	33.5	28.6	25.7	17.4	22.7	18.5	25.2	24.0	8.5
Multiflanged Disposable	6.6	7.5	5.7	11.7	20.5	17.0	0.9	3.9	3.5
Moldable	7.4	10.0	11.9	16.3	19.9	5.5	18.6	8.0	8.6
Polymer	8.6	5.0	10.8	14.2	12.5	9.0	9.1	6.0	17.6

CHAPTER V

DISCUSSION

CONCLUSIONS

Comparison of Figures 15 - 19 shows considerable variation between the insertion loss values obtained using the manikin and the attenuation values reported by NIOSH. Since comparisons are between studies done using identical types and brands of earplugs, differing results are likely due to variations in the design and/or instrumentation of the studies. There is a general tendency for the insertion loss values to be smaller than comparable NIOSH attenuation values in the three lowest frequencies. The comparative curves of four of the five earplugs studied are consistent with this low frequency characteristic. At frequencies above 500 Hz, the comparative curves show a unanimous tendency among the five earplugs studied for the insertion loss data to be greater than attenuation (human subject) data.

It appears plausible that the difference in low frequency response may be due to the differing sound pressure levels used in the two types of studies. Attenuation studies of the type reported by NIOSH use very low sound pressure level signals, at or near the threshold of human hearing. The present study exposed the manikin ear to sound pressure levels at the limit of allowable workplace noise (90 dB). Zwislocki (1958) suggested that a piston effect occurs when earplugs are exposed to high intensity, low frequency sound. In the Zwislocki model, the entire plug oscilates in the ear canal creating a piston effect between the earplug and tympanic membrane (or in the case of the present study, microphone diaphragm). Pressure changes, created by the moving earplug causes a sound pressure to occur at the tympanic membrane, or microphone.

The manikin ear canal is round, as are the earplugs fitted into it. The human ear, on the other hand, is usually oval (Goss, 1973). An earplug should fit the manikin ear canal better than a human ear canal. A more complete acoustic seal should result. Zwislocki (1958) noted the limits of ear protection were set by the alternate paths available for the conduction of sound. The manikin has none of these however. The unnaturally precise acoustic seal possible on the manikin, coupled with the lack of alternate paths for sound, probably causes the insertion loss to be overestimated in frequencies above 500 Hz.

Earplug insertion loss appears to closely follow the frequency response of the open manikin ear (see Figure 20). Dalsgard (1976) indicated any device placed in the ear canal would provide some insertion loss due to the interruption of natural resonances that provide amplification. To further test the hypothesis that resonance interruption plays a major part in insertion loss, a Pearson's Product Moment Correlation was computed comparing the insertion loss values of each earplug by frequency with the open ear frequency response of the KEMAR manikin (see Table II). The correlations were significant in all cases, the highest correlation, r = .90, being consistent with a .001 level of significance. Resonance interruption played an important part in the insertion loss provided by the five types of earplugs used in this study. The contribution of resonance interruption also followed the frequency response of the open ear, peaking in the 3000 Hz to 4000 Hz range.

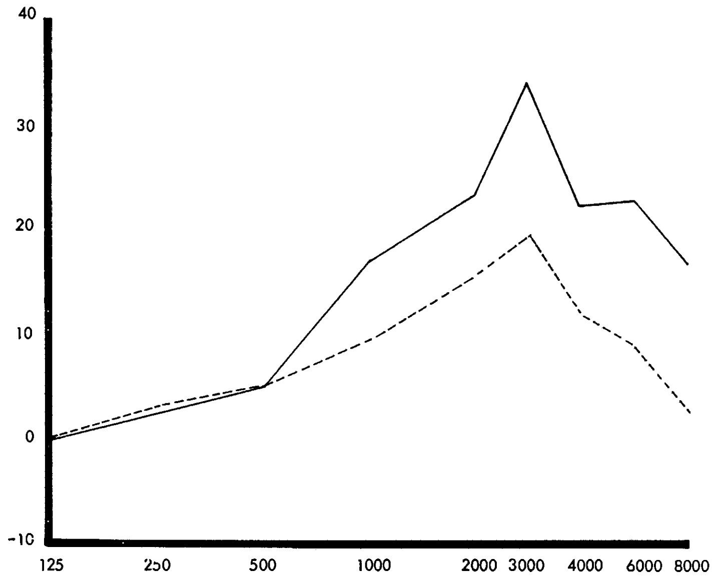

Figure 20. Adjusted insertion loss means of five earplugs (solid line) compared with the manikin open ear frequency response (broken line).

TABLE II

CORRELATIONS COMPARING INSERTION LOSS AND OPEN EAR FREQUENCY RESPONSE

EARPLUG	r-VALUE	LEVEL OF SIGNIFICANCE
V. 51-R	.90	.001
Multiflanged	.75	.02
Multiflanged, disposable	.82	.01
Moldable	.87	.001
Polymer	.87	.001

IMPLICATIONS FOR FURTHER RESEARCH

There are ample precedents for the use of mechanical couplers that approximate, but do not match, the human ear. For example, the 2cc coupler used to evaluate hearing aids and the 6 cc coupler used to calibrate audiometer earphones, both represent artificial cavities that only approximate real ear conditions. More research is necessary to standardize insertion loss values obtained on the KEMAR manikin with the Zwislocki coupler in order that the manikin may become a reliable tool for the assessment of hearing protection.

The insertion loss data obtained at lower frequencies tended to be lower than data reported by NIOSH. As previously discussed, this effect may be an artifact of the sound intensity used in the studies. Fruitful research might be directed at comparing values obtained on human subjects at high intensities and manikin values obtained at low, threshold level, intensities. If, in fact, a low frequency piston effect does occur at high intensities on both human and manikin ears, then most earplugs may fail under normal high noise working conditions.

The apparent relationship (see Figure 20 and Table II) between the protection provided by an earplug and the resonance of the ear canal should be studied carefully. If, in fact, resonance interruption is a major portion of the protection provided by earplugs, then the design of earplugs should take this fact into account. New designs might be lighter and more comfortable, relying on acoustic design rather than on mass and volume.

Considerable variability was noted in insertion loss between fittings of flanged earplugs. For example, the variability of insertion loss, as reflected in the maximum ranges (from Table I) was 27.3 dB for the single flanged V.51-R, 28.6 dB for the multiflanged plug and

20.5 dB for the multiflanged disposable earplug for an average maximum range among the flanged earplugs of 25.5 dB. In contrast, the average maximum range of the unflanged earplugs was 17 dB. Given that insertion depth was nearly identical, the differences are likely due to differing angles formed by the flanges when the plug was inserted into the canal. Further research might reveal patterns of insertion loss dependent on angular changes in flanges during insertion and reinsertion.

The manikin may be exposed to noise conditions that are undesirable for human subjects. One such application is the testing of non-linear ear protection, hearing protectors that do not become effective until a relatively high intensity is achieved. Such protectors are untestable using the standard low intensity threshold shift procedure. Using human subjects in procedures requiring high sound pressure level signals may measure attenuation, but at some risk to the subjects. The manikin, on the other hand, may be exposed to any level of sound pressure.

The KEMAR manikin offers a number of advantages over human subject research in the study of hearing protection. The most important of these is the ability to withstand noise environments unsuitable for human subjects, a high level of precision in objective measurement and its replicability.

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APPENDIX A

SAMPLE OF DATA ANALYSIS

Given the following hypothetical data from one earplug at one frequency during one fitting:

Reading #	Prior Reference SPL	Experimental SPL	Post Reference SPL
1.	93.5	72.0	93.0
2.	93.0	71.5	94.0
3.	93.5	72.5	94.0
4.	94.5	73.0	94.5
5.	93.5	73.0	94.5

Step 1. Prior and post reference readings are averaged.

Reading #	Calculation	=	Mean reference SPL
1.	(93.5 + 93.0)/2		93.25
2.	(93.0 + 94.0)/2		93.50
3.	(93.5 + 94.0)/2		93.75
4.	(94.5 + 94.5)/2		94.50
5.	(93.5 + 94.5)/2		94.00

Step 2. The reference SPL noted during the coincident sound field procedure is subtracted from the averaged values of step 1.

Reading #	Calculation	=	Correction Value
1.	93.25 - 93.50		-0.25
2.	93.50 - 93.50		0
3.	93.75 - 93.50		0.25
4.	94.50 - 93.50		1.0
5.	94.00 - 93.50		0.5

Step 3. Correction values are rounded to the lowest .5 dB and added to experimental SPL measures.

Reading #	Calculation	=	Corrected value
1.	72.0 + 0		72.0
2.	71.5 + 0		71.5
3.	72.5 + 0		72.5
4.	73.0 + 1		74.0
5.	73.5 + 0.5		73.5

Step 4. Corrected values are averaged to obtain the corrected mean.

(72.0 + 71.5 + 72.5 + 74.0 + 73.5)/5 = 72.70

Step 5. The mean corrected value is subtracted from the mean corrected open reading, obtained in steps similar to those described above, to obtain one item of raw data:

Mean corrected open		Mean corrected occluded		Insertion loss
93.5	-	72.7	=	20.8

APPENDIX B

CORRECTED MEAN INSERTION LOSS
V.51-R EARPLUG

TRIAL	FREQUENCY IN KHz								
	.125	.250	.500	1.0	2.0	3.0	4.0	6.0	8.0
1.	32.4	40.0	36.6	44.1	52.8	63.5	53.5	44.5	35.9
2.	14.0	19.9	23.2	40.0	43.6	60.0	36.8	43.0	35.9
3.	19.8	22.8	21.5	32.8	47.1	52.5	43.9	43.5	29.2
4.	7.2	12.7	19.5	27.3	36.1	54.7	31.9	45.0	35.6
5.	12.8	17.9	22.1	38.4	42.9	59.3	37.7	42.5	35.8
6.	20.8	26.5	28.4	43.6	46.2	57.4	42.7	42.3	38.1
7.	19.6	22.3	24.2	37.1	39.1	51.4	36.2	43.1	33.0
8.	19.0	23.9	20.8	26.1	31.7	54.0	31.6	44.0	30.3
9.	16.6	20.6	22.0	35.9	44.7	58.3	40.2	43.2	32.1
10.	9.8	15.3	21.4	32.9	39.6	57.2	34.9	41.6	34.3

APPENDIX C

CORRECTED MEAN INSERTION LOSS
MULTIFLANGED EARPLUG

TRIAL	FREQUENCY IN KHz								
	.125	.250	.500	1.0	2.0	3.0	4.0	6.0	8.0
1.	32.9	27.8	28.5	34.8	45.9	34.0	49.7	46.0	39.0
2.	0	-0.8	6.5	20.0	26.6	48.5	27.3	48.5	35.4
3.	25.4	22.2	23.9	27.2	36.0	52.5	40.6	24.5	30.6
4.	-0.2	-0.2	5.1	19.4	26.1	47.5	27.2	41.4	33.0
5.	18.6	15.9	16.7	20.9	26.7	36.6	28.8	43.2	22.8
6.	27.8	22.8	25.8	31.5	40.1	46.2	42.9	38.5	33.6
7.	30.5	27.1	28.2	35.0	45.2	44.1	47.6	39.6	37.1
8.	-0.6	0	5.5	20.9	24.1	35.5	26.8	42.4	33.8
9.	-0.4	0.4	6.1	18.5	24.0	39.9	24.9	47.4	30.5
10.	32.7	27.0	30.8	35.9	46.8	42.4	50.1	44.4	38.6

APPENDIX D

CORRECTED MEAN INSERTION LOSS MULTIFLANGED DISPOSABLE EARPLUG

TRIAL	FREQUENCY IN KHz								
	.125	.250	.500	1.0	2.0	3.0	4.0	6.0	8.0
1.	27.4	30.7	30.2	33.2	37.0	40.5	30.8	36.0	35.4
2.	22.7	25.5	26.4	41.8	53.3	36.5	30.3	32.5	38.9
3.	27.5	30.8	30.1	32.9	37.0	43.4	30.7	33.7	35.7
4.	22.2	25.6	26.5	40.7	53.3	49.4	30.3	35.1	36.6
5.	23.2	27.7	27.2	43.7	55.6	48.3	31.0	34.3	38.5
6.	22.7	26.0	27.6	40.9	52.8	52.5	31.1	35.7	35.9
7.	25.5	29.4	31.0	32.0	35.1	45.5	31.0	32.1	45.5
8.	27.7	32.1	31.8	34.4	37.5	39.9	30.5	33.1	35.6
9.	22.8	25.5	26.1	40.8	53.7	48.6	31.2	34.6	35.4
10.	21.1	24.6	25.9	40.1	54.9	53.5	30.9	33.8	35.5

APPENDIX E

CORRECTED MEAN INSERTION LOSS MOLDABLE EARPLUG

TRIAL	FREQUENCY IN KHz								
	.125	.250	.500	1.0	2.0	3.0	4.0	6.0	8.0
1.	13.3	16.0	23.2	36.8	37.6	48.0	43.4	41.0	33.4
2.	16.8	20.7	30.5	46.0	47.1	49.0	54.9	36.2	38.0
3.	13.9	17.3	24.8	38.8	41.2	47.5	47.8	35.0	36.8
4.	11.6	13.2	19.0	31.2	31.6	51.5	37.3	34.5	30.7
5.	15.2	16.7	24.2	36.9	37.6	53.0	45.7	40.3	36.6
6.	14.5	15.8	22.0	35.1	37.7	52.5	44.2	42.5	30.2
7.	19.0	23.2	32.4	45.8	51.5	49.0	55.9	41.0	38.6
8.	13.0	15.2	20.5	29.7	33.1	48.7	40.5	43.5	30.6
9.	15.0	18.8	24.7	37.6	39.9	35.3	46.4	42.0	35.3
10.	14.6	17.6	25.7	39.8	41.0	52.1	46.1	35.5	38.8

APPENDIX F

CORRECTED MEAN INSERTION LOSS
POLYMER EARPLUG

TRIAL	FREQUENCY IN KHz								
	.125	.250	.500	1.0	2.0	3.0	4.0	6.0	8.0
1.	20.8	27.8	35.5	43.3	49.0	64.0	56.7	56.5	48.3
2.	23.6	30.5	28.6	45.9	53.9	61.5	60.9	50.5	52.4
3.	29.4	30.1	25.8	40.2	48.5	70.5	54.6	51.5	34.8
4.	23.0	27.6	35.7	43.0	49.6	66.3	56.7	52.3	47.1
5.	24.4	29.6	31.9	44.3	51.6	64.2	57.6	55.4	46.4
6.	21.6	27.3	26.0	39.1	48.1	67.2	51.8	55.3	40.5
7.	21.7	26.5	27.2	41.8	44.2	68.5	55.2	52.1	41.1
8.	27.6	31.1	27.4	53.3	56.7	66.3	59.8	54.3	51.4
9.	22.0	28.7	33.1	44.3	47.1	62.3	55.2	52.1	39.9
10.	25.7	31.5	24.9	45.4	53.3	69.7	58.4	55.7	41.2