MANIKIN MEASUREMENTS

Proceedings
of a
Conference
Organized By
M.D. Burkhard

Industrial Research Products, Inc.
A Knowles Company

TABLE OF CONTENTS

Preface.

Knowles Electronics through its subsidiary Industrial Research Products, Inc., developed the KEMAR Manikin as a tool for improving the measurement and reporting of performance of hearing aids. Since its first description and introduction in 1972, a number of hearing aid manufacturers, research audiologists and others have investigated the characteristics of hearing aids when mounted on the KEMAR Manikin, and also have sought out other applications in which it is desired to simulate actual use of acoustical devices on people or the interaction of an average person with the acoustical environment. During the intervening time a number of useful techniques have evolved which greatly simplify the reporting of *in situ* characteristics of hearing aids. The results have been sufficiently useful and worthwhile to lead several hearing aid manufacturers to begin reporting hearing aid performance in terms of data obtained on the KEMAR Manikin.

As with many new measurement tools when they are first introduced, several philosophies of use evolved and it was considered appropriate that a forum be created wherein the interested persons could share their experiences, compare procedures and results and, thereby, achieve a uniformity of use and reporting. Ultimately this might lead to standardization of simulated *in situ* measurements of hearing aids. Two conferences were organized and sponsored by Industrial Research Products, Inc., one in Zurich on March 4, 1976 and one in Washington, D.C. on April 5, 1976 for this purpose.

Several persons accepted our invitation to talk about their experience with the KEMAR Manikin. Following the conferences the contributors who had made formal presentations, were asked to provide a manuscript summarizing material they presented. This volume contains those manuscripts, as submitted, and also contains a summary of my remarks as the organizer, which were not otherwise covered by the contributed manuscripts. The conference at which the respective contributors presented their material is indicated with each chapter. My own contributions are a summary of material presented at both conferences and no distinction as to the date is made, for these chapters.

Material was presented on some applications of a non-hearing aid nature. This information is helpful because it indicates the uses and limitations of manikins but the main emphasis was on applications to hearing aid characterization.

An important result of a conference of this type is the discussion that is stimulated. The conference proceedings were recorded and the discussions that followed the various presentations are included to indicate some of the concerns and some of the points of agreement that existed at the conclusion of the conferences.

It is hoped that, in a small way, these proceedings will provide a base from which the delivery of hearing aids to the hard of hearing can be further improved; and that by virtue of the participation of engineers, audiologists and researchers who are working with auditory prosthesis, the forum has provided an interchange of ideas of benefit to the community. We also anticipate that the information exchanged will find its way into the deliberations that are necessary to achieve standardization in this area.

As organizer I am indebted to my colleagues at Industrial Research Products, Inc., Knowles Electronics, Inc. and Knowles Electronics Limited, for their help in making arrangements for the conferences and for their assistance in preparation of material. I would particularly like to acknowledge the assistance of R.J. Maxwell in the collection of some of the data and evaluating some of the procedures, presented in Chapters 10, 11, 12 and 13. The conferences and the publication of this proceedings came about through the support and encouragement of Hugh S. Knowles.

Elk Grove Village, Illinois USA
February 1978.

Mahlon D. Burkhard.

Contributors

L.B. Beck, M.A.
Biocummunications Laboratory
University of Maryland
College Park, Maryland
USA

Burkhard, Mahlon D., M.S.
Manager of Research
Industrial Research Products, Inc.
321 N. Bond St.
Elk Grove Village, IL 60007
USA

Burnett, Edwin D.
Sound Section
Institute for Basic Standards
National Bureau of Standards
Washington, D.C. 20234
USA

Causey, G.D., Ph.D.
VA Hospital
50 Irving Street
Washington, D.C. 20422
USA

Dalsgaard, S.C., M.Sc.,
Head of laboratory
Laboratoriet for Teknisk-Audiologisk Forskning,
Odense Sygehus,
DK-5000 Odense.
Denmark

Helle, Roland, Dr. Ing.,
Manager of the Hearing Aid Laboratory
Medical Engineering Group
Siemens AG.
Postfach 500
D-8520 Erlangen
Germany

Kuhn, George F., Ph.D.,
Electronics Engineer, Sound Section,
Institute for Basic Standards
National Bureau of Standards
Washington, D.C. 20234
USA

Lauridsen, Ole, M.S.,
Topholm and Westermann
Ny Vestergardsvej 25,
DK-3500 Verlose
Denmark

Preves, David, EE.,
Vice President - Engineering
Starkey Laboratories, Inc.
700 Washington Ave., South
Eden Prairie, Minn. 55343
USA

Chapter 1.

Introduction

Summary of Introductory Remarks by Hugh S. Knowles, President, Knowles Electronics, at the Conference held April 5, 1976, in Washington, D.C.

I'd like to take a moment to put this conference into a larger context. We first felt the need a number of years ago of calling people's attention to the difference between IEC 118 standards measurements, which at that time were also ASA [now ANSI] hearing aid measurements, and what occurs in the real world of *in situ* performance of hearing aids on listeners. In a group of three-day workshops started by the Hearing Aid Industry Conference in 1959, we attempted what we thought was Messianic work of trying to interest hearing aid manufacturers and dispensers in the differences by pointing out what was known about the sound transmission into an ear from a remote source and from a hearing aid. Acoustical measurements people were doing research with physical measures which they understood thoroughly as being representative of the performance of the hearing aid. They did not concern themselves with subjective measurements. Everyone that was involved in such measurements knew the limitations. The people in the field who were trying to apply the results found that any relation between, for example, the response vs frequency graph as plotted and its behavior in real life was purely coincidental. The people making the physical measurements, and who were involved in much of the standards committee work, were well versed in the relevant phenomena and were concerned that measurements could be compared worldwide among laboratories. But somehow the relationships and limitations escaped the people who were trying to apply them. As a result, we found thousands and thousands of hours of research in the behaviorial field which was misguided because of misunderstanding of the limitations of the physical measurements.

We became increasingly concerned over this dilemma and, finding that we had somehow not communicated our enthusiasm to organizations that we thought should carry out the necessary work, we in Industrial Research Products undertook some of the developments and research ourselves. Two considerations were dominant in our thinking: The solution we might come up with would be done more promptly because we would be willing to tolerate larger errors than would prestigious institutions such as a National Bureau of Standards or a National Physical Laboratory. Secondly, we could get the results into the hands of persons who need the information quickly.

The acoustic behavior of what appears to be a simple thing, an ear, was incompletely understood and comprehensive physical measures are amazingly lacking in the literature. We have Edgar Shaw to thank for very much of the recent work that's been done in this field. It is well known that 2-cm^3 couplers do not represent the ear at all well. The 2-cm^3 coupler was standardized thirty-odd years ago, and I remember the committee that prepared the standard expected it to have a five-year life. It was, and still is, an extremely useful device, if one recognizes its limitations, for hearing aid and hearing aid receiver measurements. Recent developments that increase the frequency range of receivers for hearing aids have emphasized the need for circumventing these limitations. But a more important need was the push to use directional microphones in hearing aids. Conventional hearing aid measurement soundboxes and other production control-type systems were completely unsuitable for evaluating these devices. They simply must be measured or referred to the *in situ* conditions. This requires either a jury of reasonable size or development of an objective instrument method, and this suggested the manikin approach.

Another need resulted from the introduction of the CROS fitting for hearing aids, or putting it more generally, the leaky type of insert or earmold. Sound reaches the ear or the tympanum directly as well as by amplification through the hearing aid. Because diffraction plays such an important role in the sound arriving at the tympanum, it must be accounted for. One cannot just make a cavity. We cannot hide behind the simplistic approach that has been used in IEC Committee SC-29C, for many years, which was that we should be satisfied with a coupler having a driving point impedance that corresponds to the acoustic termination of the particular earphone, be it circumaural, supra-aural or an insert receiver. The transfer characteristics of the ear in normal *in situ* use must be duplicated.

These considerations are what finally got us launched on this comprehensive program of developing an acoustic measurements manikin primarily for hearing aid research and engineering. It was not our intent, nor our expectation, that we could, in any reasonable length of time, turn out something that would not be subjected to considerable comment on the part of purists, but we were hopeful and have been very gratified to find that people who recognize the need for this type of device have been cooperative in making constructive criticisms and suggestions about the KEMAR manikin. We hope we have left the device sufficiently flexible so that these improvements can be made.

I would like to add comments which are a byproduct of preparation for a recent presentation to the AAAS [Boston, February 1976]. There are in the world, depending on your criteria on hard-of-hearing, approximately 200 million people who need some kind of hearing prosthesis. This includes, of course, all of the under-developed countries. Psychological research has shown that hearing is very important, not only specifically to speech communication in the narrow sense of the word, but also to the level of performance of the individual in his social environment. Research keeps pushing back the age at which lack of normal hearing should be determined in a very small child. Unless corrective measures are taken very early, a child never attains the asymptotic performance, as we measure it in our modern scholastic system, of which he may be capable. Hearing is an extremely important function in the performance of the individual in our culture.

The emotional trauma that goes with lack of hearing is a difficult thing to assess quantitatively, but it is readily observable by persons working with the deaf. The hard-of-hearing see visual stimuli around them: They see people talking; they see people laughing; they are continually aware of things that they are missing the response to, and they can easily become paranoid and feel that, for example, laughter is at their expense.

Thus, whatever this group here today can achieve in the way of improving our knowledge of protheses will have tremendous value to the vast group of hearing impaired. That is why it is so important for people who are professionally involved with the problems, primarily audiologists, otologists and otolaryngologists, and engineers, to sort out and improve upon the various aspects of prothesis design and specification.

Much of the basis for hearing aid engineering and specification, today, is from studies that are quite old, such as the Medical Research Council report (London, 1947) and the work at the Harvard laboratory. Now, many decades later, technology has moved ahead so that it is possible to do things that were impossible then. For example, size which contributes to vanity is an important factor in people's selection of hearing aids. Technology has permitted many of the cosmetic requirements to be attained with the design of small devices that have broad frequency range and a good signal-to-noise ratio. In short, the technology has gotten considerably ahead of our understanding of what the optimum characteristics of hearing aids should be. The clinical work to tell the engineers what to build, lags behind what the industry is capable of providing technologically. We hope very much that a seminar of this type will stimulate persons working in the field to pursue vigorously the measurement and interpretation of hearing aid performance in ways that will provide useful prosthesis to that 200 million population of hard-of-hearing people.

Chapter 2.

Anthropometric Manikin For Acoustic Research

M.D. Burkhard
and
R.M. Sachs
Industrial Research Products, Inc.

The following description of the KEMAR manikin is reprinted with permission from the Acoustical Society of America. It was the basis of a presentation by Mr. Burkhard in both the Zurich and Washington Conferences.

For reference in other parts of the proceedings, figures in this Chapter will be preceded by 2—.

A manikin for hearing aid and related acoustic research was designed with median human adult dimensions. Ear simulation matches the acoustic response with an auricle, an ear canal, and an eardrum that equal the median ear in dimensions, acoustic impedance, and modes. Dimensions of torso and head are based on published anthropometric data, but the auricle is based on data obtained for this development. The ear canal and eardrum are adapted from the earlike coupler by Zwislocki. The ear entrance sound pressure was found to be relatively insensitive to surface or skin impedance of the head. Validating measurements show the manikin, designated KEMAR, to be like a median human in acoustic response to free fields.

Subject Classification: 65.22, 65.80, 65.35, 65.82.

INTRODUCTION

Head and body diffraction effects encountered when fitting hearing aids have been recognized and evaluated a number of times. However, the advent of head-worn hearing aids equipped with directional microphones and the newer open ear hearing aid fitting techniques, such as CROS and vented earmolds, create a new need to determine the performance under more lifelike conditions. The various parameters needed to convert a standard free-field hearing aid response to the equivalent performance on an average individual may be determined for each fitting method, but a realistic estimate usually requires a number of observations on a number of individuals. Physical conditions vary enough among the various fittings, e.g., microphone type and position on the head or body, hearing aid location, and ear canal closure conditions, that only a few parameters apply to all situations. As a further experimental constraint, it is sometimes difficult to vary only one or two parameters at a time with an individual to determine their effect on the overall acoustic response of the hearing aid. An appropriately proportioned and designed manikin would provide lifelike test conditions and experimental flexibility. KEMAR (Knowles Electronics Manikin for Acoustic Research) was constructed as a test and evaluation tool that gives wearer simulation of all types of hearing aid fittings using the following criteria.

(1) Average anthropometric dimensions of an adult human.

(2) Ear canal and eardrum to match real ears in open, partially closed, and closed ear use.

(3) Acoustically and dimensionally average pinna.

(4) Easily exchangeable pinna to permit study of ear size effects.

(5) Reproducibility.

Romanow (1942) showed how sound diffraction around the body of a hearing aid wearer altered a hearing aid response and should be taken into account. A test was proposed by Carlisle and Mundel (1944) to include body diffraction, the so-called body baffle effect, in a body-worn hearing aid response measurement. More recently, Wonsdronk (1959) studied the diffraction problem for head-worn hearing aids by recording over-the-ear hearing aid responses on ten men and ten women. He then attempted to simulate the diffraction effects with two simple head models: a sphere and a box each with simple auricles. His main conclusion was that diffraction around these simple models could not duplicate that of the human body, especially below 2 kHz. A large minimum in over-the-ear pressure at 1300 Hz measured on his subjects was not reproduced. A plaster cast of a human head was unexplainably inadequate. An attempt to take into account torso diffraction by adding absorbing material between the head models and the anechoic chamber floor did not change the conclusions significantly, probably because a human torso is acoustically reflective rather than absorbent.

Lifelike heads without torsos have also been used for acoustic measurements by Nordland (1962), Kasten and Lotterman (1967), Damaske and Wagener (1970), Mellert (1972), Von Wilkins (1972), and Muldoon (1973). Bauer *et al.* (1966) constructed a reproducible head and torso. This manikin has a $\frac{1}{8}$-in. -thick Plastisol "flesh" overlaid on a polyester fiberglass skull, artificial pinna, ear canals and eardrums (with 1-in. microphones at the eardrum locations). Its dimensions are larger than the mean or average human adult, matching the average dimensions of the original seven male astronauts of the NASA Mercury Program.

Only Bauer *et al.* (1966) appear to have given attention to duplicating ear canal sound transmission. Open ear canal sound pressure transfer ratios on their manikin agreed with the Wiener and Ross (1946) data to within 3 dB for frequencies below 5 kHz. Damaske and Mellert (1969) and Von Wilkins (1972) matched "dummy head" open ear response characteristics to the subjective loudnesses perceived by listeners in a free field with various angles of sound incidence. Nordland (1962) was only concerned with interaural phase and amplitude differences as a function of azimuth angle, and therefore made no attempt to duplicate the frequency response characteristics of an ear. Unlike others, Kasten and Lotterman (1967) were concerned chiefly with head diffraction effects on hearing aids connected to earmolds that closed the ear canal, and thus did not consider the details of the ear canal simulation. Similarly, Muldoon was concerned with diffraction effects in the monitoring of occupational noise exposure of

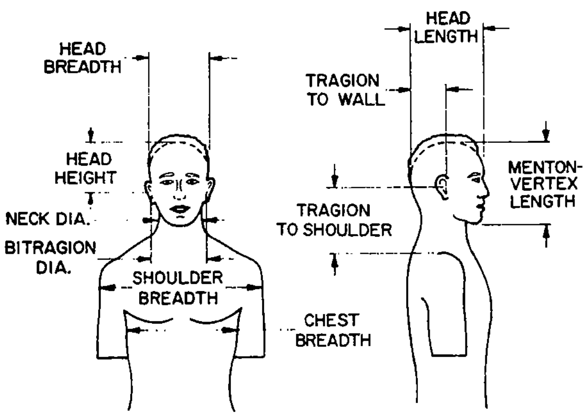


FIG. 1. Anthropometric measures used in design of KEMAR.

workers and did not take account of the ear canal.

I. KEMAR DESIGN DIMENSIONS

A. Torso and head

Each dimension of the manikin was chosen to represent an average human adult. Particular attention was given to head and torso dimensions believed to be critical acoustically. Principal references were Churchill and Truett (1957), giving data on head and face dimensions from a 1950 survey of over 4000 male flying personnel (mean age 28) and 852 WAF trainees (mean age 20), and "The Measure of Man" portfolio by Henry Dreyfuss (1967). Important dimensions are shown in Fig. 1: The bitragion diameter (head diameter at notch above tragus, i.e., at the anterior notch), head size (length, breadth, and chin to head top length), location of ears on head, neck diameter, shoulder and chest breadth, and the distance from shoulder to ear. The final dimensions of KEMAR are compared to male and female median values in Table I. In all cases KEMAR dimensions are within 4% of the average.

B. Ear canal and eardrum

Design of the ear canal and eardrum simulator to match acoustic data of real ears was based on the coupler design of J. J. Zwislocki (1970, 1971). Figure 2 is a cross-section drawing of the eardrum portion of the

TABLE I. Dimensions for KEMAR and average human adults, in centimeters.

	Median male	Median female	Average human	KEMAR
Head breadth	15.5	14.7	15.1	15.2
Head length	19.6	18.0	18.8	19.1
Head height	13.0	13.0	13.0	12.5
Bitragion diameter	14.2	13.5	13.85	14.3
Tragion to wall	10.2	9.4	9.8	9.65
Tragion to shoulder	18.8	16.3	17.55	17.5[a]
Neck diameter	12.1	10.3	11.2	11.3
Shoulder breadth	45.5	39.9	42.7	44.0
Chest breadth	30.5	27.7	29.1	28.2
Menton vertex length	23.2	21.1	22.15	22.4

[a]Adjustable over ±1.27 cm.

coupler. The Zwislocki coupler has a central cylindrical hard wall cavity with diameter (7.5 mm) close to an average human adult ear canal. Four side branches R1 through R4, located near the microphone, synthesize the acoustic impedance variation with frequency that has been observed on ears. Each side branch consists of a series acoustic network with inertance, resistance and compliance. The KEMAR ear canal length from the entrance to the ½-in. microphone (eardrum)

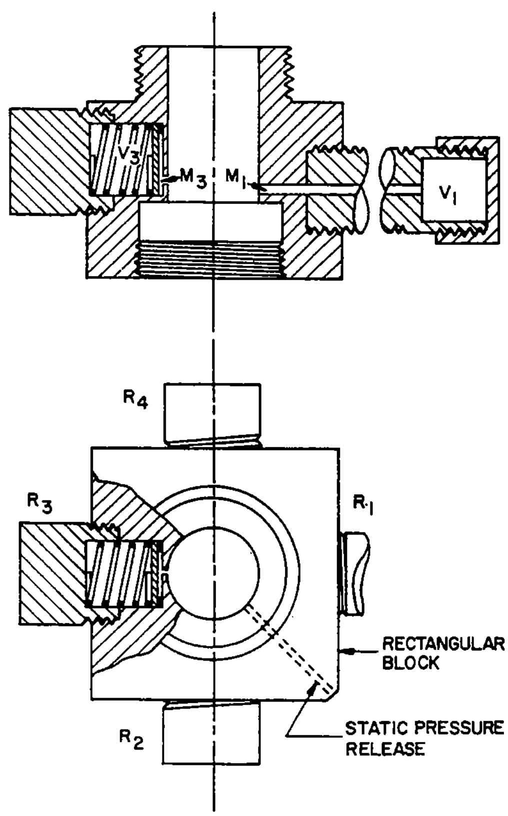


FIG. 2. Schematic drawing of Zwislocki eardrum simulator.

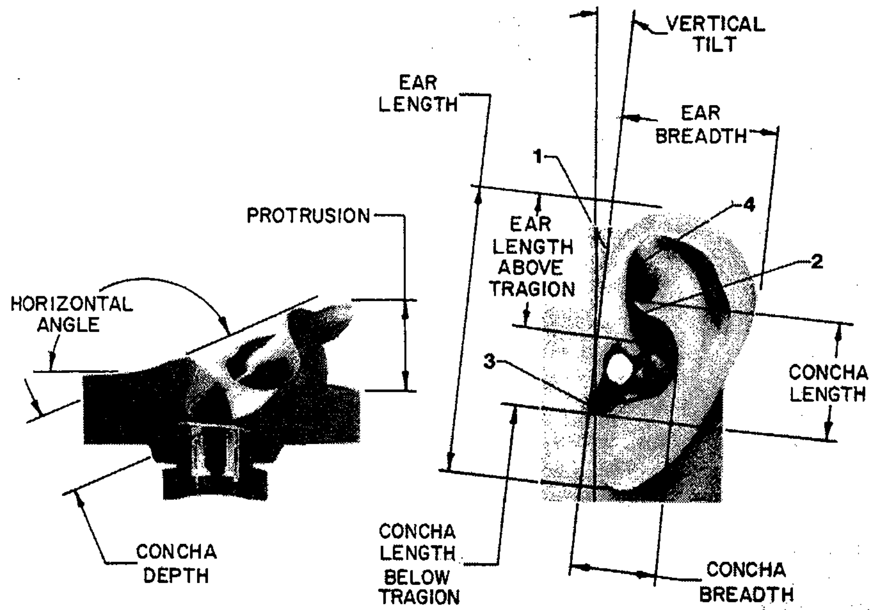


FIG. 3. Auricle measures used in design of KEMAR. (1) Upper pinna-skull notch; (2) antihelix; (3) intertragal notch; (4) crus of helix.

termination is 2.15 cm. This length is less than on the average person for two reasons: The sound velocity is greater at body temperature than at room temperature, and the microphone compliance adds effective length. Thus, the canal resonance frequency of the coupler matches the average ear. Most calibrating couplers have synthesized the acoustic impedance of the ear seen by an external source, such as an earphone, but lack geometrical reality and hence the transfer impedance synthesis which the Zwislocki coupler accomplishes and which is necessary in a general purpose acoustic test manikin.

C. Auricles

Auricles were selected to be dimensionally and acoustically average. Only the external dimensions of the adult pinna have been reported in the literature for a large sample (e.g., Dreyfuss, 1967; Alexander and Lauback, 1968), namely, ear length, ear length above the tragion, ear breadth and ear protrusion. Studies by Yamaguchi and Sushi (1956), Shaw and Teranishi (1968), and Shaw (1966, 1972) have emphasized the importance of concha shape, size, and orientation in determining the acoustic responses of the ear. Although Zwislocki deduced concha dimensions from the ear impedance data of Delaney (1964), no quantitative measures were used that could be correlated to acoustic free-field response deemed essential to a satisfactory manikin. So the remaining auricle dimensions of concha depth, horizontal angle, concha length, concha length below the tragion, concha breadth, and vertical tilt were measured on 12 males and 12 females and are the basis of selection. All of these dimensions are given in Fig. 3 and Table II.

Concha opening breadth, length, and length below the tragion can be measured readily with calipers or rulers.

TABLE II. External ear dimensions.

Dimension		Averages 12 Male	12 Female	Overall	Standard deviation 12 Male	12 Female	Overall	KEMAR	50% Male[a]	50% Female[a]	Average
Ear length	cm	6.85	6.24	6.55	0.59	0.38	0.58	5.89	6.35	5.84	6.10
Ear length above tragion	cm	3.30	3.07	3.19	0.41	0.20	0.34	2.7	3.04		
Ear breadth	cm	3.77	3.36	3.57	0.24	0.27	0.33	3.1	3.55	3.3	3.42
Ear protrusion	cm	2.28	2.03	2.16	0.22	0.23	0.26	1.85	2.10		
Ear protrusion angle	deg	156.7	155.1	155.9	8.6	9.7	9.0	158			
Vertical tilt front view[b]	deg	3.0	2.7	2.9	3.2	3.6	3.1	7			
Vertical tilt side view[b]	deg	7.6	4.7	6.2	2.8	3.4	2.8	6			
Concha volume	cm^3	4.65	3.94	4.30	0.76	0.81	0.85	4.0			
Concha length	cm	2.73	2.53	2.63	0.23	0.20	0.24	2.45			
Concha length, tragion to lower notch	cm	1.74	1.62	1.68	0.16	0.16	0.17	1.82			
Concha breadth	cm	1.88	1.72	1.80	0.21	0.21	0.22	1.57			
Concha breadth tragion to helix	cm	1.82	1.65	1.73	0.27	0.22	0.25	1.39			
Concha depth	cm	1.29	1.29	1.29	0.12	0.08	0.10	1.33			

[a]Dreyfus (1967).

[b]Four males and four females.

The front edge of the ear is first identified by a line from the upper pinna–skull notch to the anterior edge of the intertragal notch. Breadth is the perpendicular distance between this line and a parallel line tangent to the antihelix. Concha length is the distance from the intertragal notch to the intersection of the crus of helix with the lower crus of antihelix. Both length and breadth were measured from photographs and are from projections onto a plane tangent to the head. Therefore, the tabulated breadth is about 10% smaller than if measured in a plane tangent to the pinna. Concha depth and volume were determined from ear impressions. Each impression was weighed immediately after removal and volume calculated. The impression was then sectioned and the depth measured as indicated in Fig. 3. Concha shapes varied considerably among the 24 subjects.

The pinna on an average person from our sample tilts inward from vertical at the bottom with an angle of 3°. This is determined by locating a line vertically on the side of the head that passes over the center of the ear canal entrance, and just touches the upper and lower

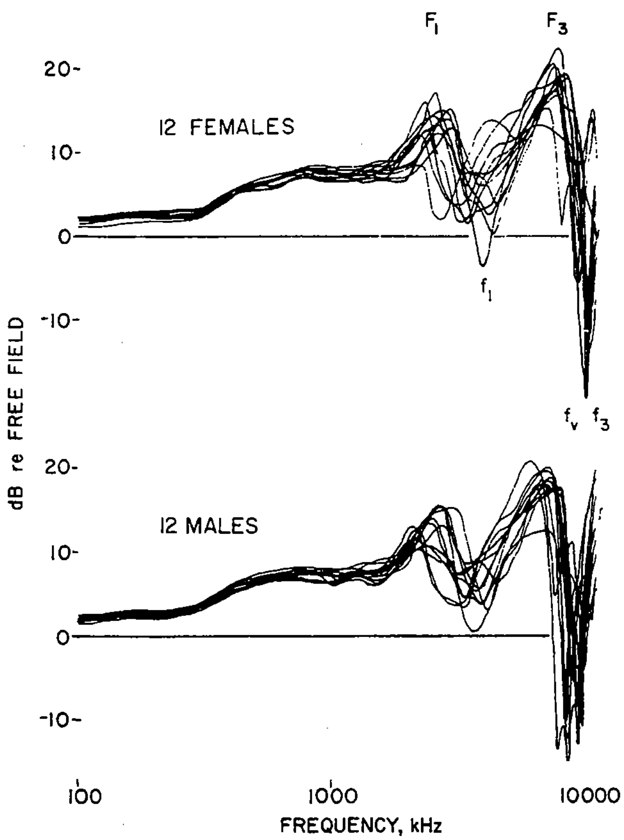


FIG. 4. Ear canal entrance sound pressures, 12 males, 12 females, sound source at 90° azimuth angle, 1m. Free field is the reference pressure condition.

edges of the pinna as viewed from the side. The angle this line forms with vertical when viewed from the front is the pinna tilt angle, front view, and is slightly larger, 7°, on KEMAR, to fit other average head dimensions judged to be acoustically more important.

The final selection of an appropriate average auricle also took account of frequency dependence of sound pressure at the ear canal entrance of the same 24 subjects. Pressure at the ear canal entrance and at the eardrum, although different from each other as shown by Weiner and Ross (1946) and others, exhibit the same dependence on concha and other external ear resonances, sound direction and diffraction around the head and torso. In other words, the variation of the eardrum to ear canal entrance sound pressure ratio depends only on the characteristics of the ear canal and eardrum. Thus it is unnecessary to evaluate the sound pressure at the eardrum of the subjects, if the ear canal acoustics studies can be accepted. For measurement of sound pressure at the ear canal entrance, an equalized Knowles Electronics BT-1751-type microphone was placed in the bottom of the concha with its sound port projecting over the open ear canal entrance. The fact that the external ears of people are acoustically different can be appreciated from the collected plots of sound pressure in the concha for 12 adult males and 12 females with 90° sound source azimuth angle shown in Fig. 4. Nevertheless, there are common features that have been identified by Shaw and Teranishi (1968) for a nearby point sound source as correlating with external ear excitation modes. Frequencies of pressure response minima f_1 and f_3 are identified with the $\frac{1}{4}$ and $\frac{3}{4}$ length modes of the ear canal, while the frequencies of pressure response maxima F_1 and F_3 give a measure of the total effective length of the combined concha and ear canal for $\frac{1}{4}$ and $\frac{3}{4}$ wavelengths. The two frequencies f_1 and f_3 exhibit good consistency in predicting the average effective length of the ear canal. Frequency f_3 could be identified for all subjects. Another response minimum at frequency f_v, which is attributed to the concha only, was identifiable on the response curves for ten males and only five females with 90° sound incidence and with nine males and five females with 0° incidence. In the nine persons exhibiting the minimum of f_v at both 0° and 90° incidence, the average frequency was 0.27 kHz higher for 90° incidence. According to Shaw and Teranishi (1968), equivalent lengths of the ear canal and the total ear have the following relationships to these singularity frequencies where c is the velocity of sound: $L_1=c/4F_1$, $l_1=c/4f_1$, $l_3=3c/4f_3$, and $L_3=3c/4F_3$.

In Table III, the average of these frequencies and the equivalent lengths are shown for the 12 male and 12 female ears. The total equivalent lengths of the ear calculated from F_1 and F_3 show surprisingly good consistency, while the ear canal equivalent length is somewhat less consistent, owing perhaps to some uncertainty in deciding whether a minimum was f_v or f_3. No equivalent length is given for f_v because it appears to result in part from other than simple standing wave resonance effects. Although most applications for KEMAR will be for frequencies below 7 kHz, the acoustics of the external ear for frequencies extending above 10 kHz adds validity to the ear simulator specification.

D. Acoustic horizontal

In view of the rather strong dependence of ear canal sound pressure on vertical or elevation angle of the sound source (Shaw and Teranishi, 1968), a well-defined horizontal is essential when KEMAR is used for open ear measurements. The final mounting and positioning is such that horizontal is defined by the line connecting the lower eyelid of the open eye and the upper pinna-skull notch, as viewed from the side.

TABLE III. Mean values of frequencies of pressure maxima and minima at the open ear canal entrance for 12 male and 12 female subjects; sound incidence from 90°; loudspeaker source at 1 m. Equivalent lengths based on $c = 354$ m/sec for l_1 and l_3, $c = 349.5$ for L_1 and L_3.

	Mean (kHz)	Std. dev. (kHz)	Equivalent length (mm)
F_1 Male	2.45	0.25	35.6
Female	2.56	0.26	34.1
F_3 Male	7.06	0.57	37.1
Female	7.66	0.58	34.2
f_1 Male	3.63	0.40	24.4
Female	3.69	0.53	24.0
[a]f_3 Male	9.81	0.59	26.1
Female	10.48	0.60	25.4
[b]f_v Male	8.66	0.59	
Female	9.10	0.51	

[a]11 males, 11 females. [b]10 males, 5 females.

II. KEMAR CONSTRUCTION

KEMAR, shown in Fig. 5, is fabricated of fiberglass-reinforced polyester from specially prepared molds. The head separates from the torso and is free to rotate at the neck. The top of the head is removable to provide access to the interior of the head. The neck is hollow and provides for passage of instrumentation cables and accessories from the head to the torso interior. The torso portion extends downward to below the waist where there is provision for mounting to a flat surface. The arms of the torso terminate at the elbow. Access to the interior may be made through a panel in the back of the torso.

The wall thickness of the manikin is approximately $\frac{1}{4}$ in., and the interior has been coated with lead-pellet-filled resin to provide additional mass and reduce the coupling of the manikin to acoustic fields.

Mounting support for KEMAR provides a vertical axis of rotation that bisects the line between the right and left ear canal entrances. When rotated, this head center point stays fixed in space.

As a practical matter, special consideration was given to details of mounting of the ears which are removable and replaceable. The two external ears were cast with a soft, tear-resistant RTV silicone rubber giving simulated pinna flexure, easy insertion of earmolds, and accommodation to headphones. Each external ear snaps into a recess in the side of the head and also into the inner connecting machined piece that provides an ear canal extension and forms a part of the

FIG. 5. Photograph of KEMAR with a wig.

ear canal—eardrum simulation. This can be seen in the ear construction cross section in Fig. 3. Median size pinnas have been developed, but the method of attachment permits fabrication and substitution of other sizes.

Two neck rings were constructed so that the shoulder-to-ear distance could be adjusted from male to female median values. The range of adjustment is 2.54 cm in 1.27-cm increments.

A view of the head interior, Fig. 6, shows the eardrum simulator in place. The Zwislocki coupler here used a Brüel & Kjaer $\frac{1}{2}$-in. condenser microphone with a flexible right angle adapter (B & K UA0122).

III. ACOUSTICAL CHARACTERISTICS

Acoustical measurements of KEMAR show good agreement with similar data on persons. This section summarizes the results of a number of validating measurements. Unless noted, all comparisons are relative to a free field and use the right ear. Most comparisons will be to the summary by Shaw (1974) which was a comprehensive review of published data on external ear contribution to directionality in hearing. All observations are for a sound source in the horizontal plane only.

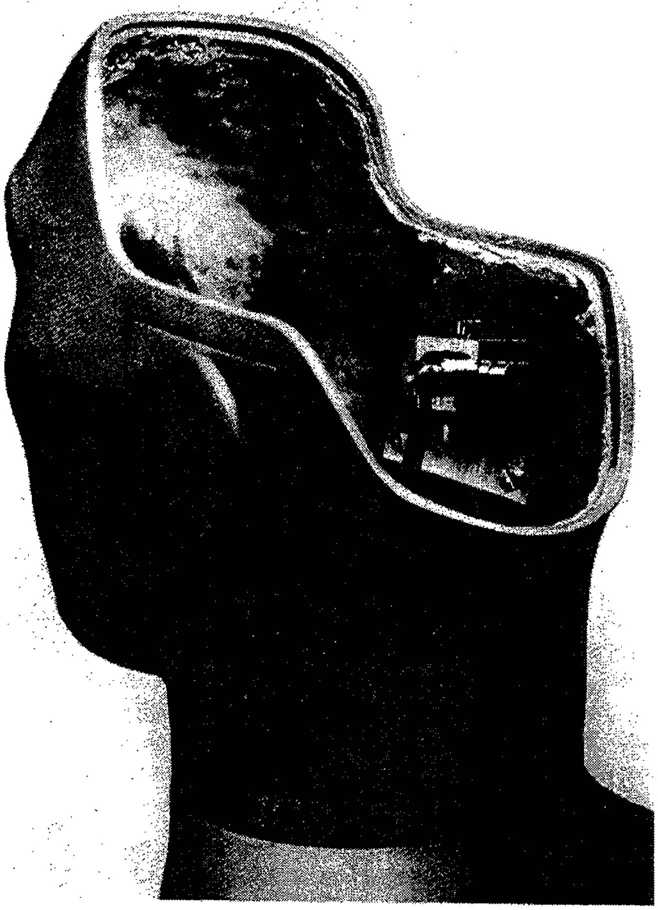


FIG. 6. Photograph of the interior of KEMAR showing eardrum simulator with a Brüel & Kjaer $\frac{1}{2}$-in. microphone and a right-angle adapter.

A. Eardrum pressure

Figure 7 shows the KEMAR eardrum pressure for the free-field sound source for several source azimuth angles. The free field was generated by driving the loudspeaker with the electrical signal that produces a constant free-field pressure on a microphone located at the center head location, sans manikin, as in a technique described by Wonsdronk. There is general agreement with Shaw at all angles. The main peak at 2.5 to 3.0 kHz, independent of azimuth angle in the horizontal plane, is due to the $\frac{1}{4}$-wavelength resonance of the combined ear canal and concha. Some deviations between the two curves may be expected because KEMAR's response was taken with pure tones, whereas, as Shaw pointed out, the composite is more typical of measurements on individuals with $\frac{1}{3}$-octave noise bands. Shaw attributes the response minimum near 10 kHz to a concha antiresonance mode that is poorly coupled to the horizontal sound field. The frequency of this mode in the KEMAR response shows more dependence on azimuth than Shaw assumes, varying from 8 kHz at 0° and 180° to 10 kHz at 90°. This minimum is an important acoustic reference frequency for selection of external ear design, as noted previously.

The curves in Fig. 7 display the important total diffraction effect for human listening in a sound field. They also show the directional influence of diffraction on frequency response and the resulting spectral colorations that give additional directional clues for complex sounds.

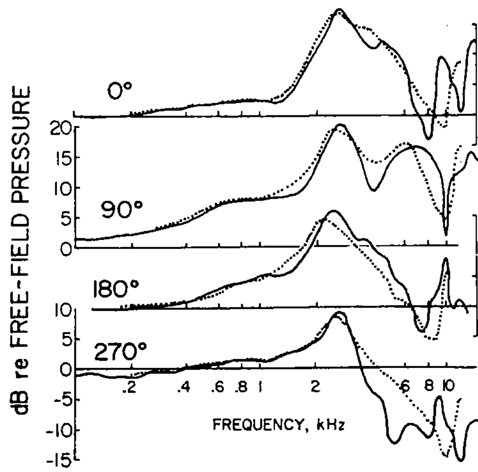


FIG. 7. KEMAR eardrum sound pressure for a free sound field with source azimuth angles of 0°, 90°, 180°, and 270°. ——KEMAR; ··· Shaw composite.

B. Ear canal pressure transfer ratio

Eardrum to ear canal entrance pressure ratio is shown in Fig. 8. Shaw's reference curve was derived from data of Wiener and Ross (1946) and Zwislocki (1970) and extended to 12 kHz with ear models (Shaw, 1971). Wiener and Ross and Zwislocki give standard deviations ranging from 3 to 5 dB for the ear canal transfer ratio in the 3–5-kHz region. The deviation of this ratio on KEMAR from Shaw's composite is less than 2 dB up to 10 kHz and is small compared to the range in the cited data. The ratio on KEMAR is independent of sound source location in the horizontal plane, similar to the conclusion of Shaw (1971). This ratio depends almost entirely on the acoustic impedance and dimensions of the eardrum and ear canal. Above 7 or 8 kHz, the ear canal–eardrum simulator acts as a hard wall tube closed with a rigid eardrum. Thus, the $\frac{3}{4}$-wavelength canal resonance in KEMAR at 12 kHz is highly underdamped compared to that found in typical real ears.

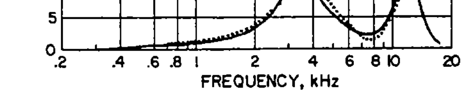


FIG. 8. Ratio of the eardrum to ear canal entrance sound pressure. ——KEMAR ···Shaw composite.

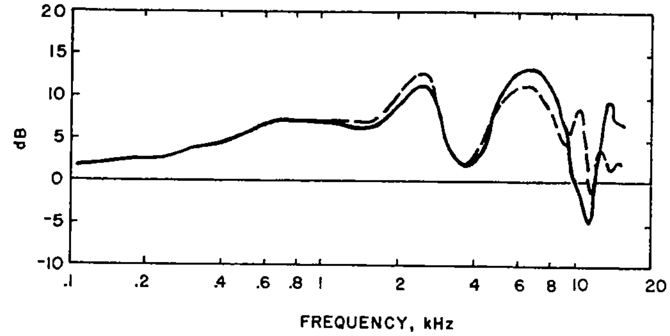


FIG. 9. Sound pressure at the entrance of the open ear canal on KEMAR, both ears, with sound source on the same side as the ear being measured. These curves should be compared to Fig. 4. Reference sound pressure is free field. ——right; – – – left.

C. Ear entrance sound pressure

Ear entrance sound pressure versus frequency for the two KEMAR ears when excited in a free field with source facing the respective ear are shown in Fig. 9. Sound pressure at the open ear canal entrance of KEMAR shows most of the details recorded in the 24 ears shown in Fig. 4 and may be considered typical of this population for frequencies up to 10 kHz, although the 6–7-kHz response peak is not as sharp as recorded in some ears. Closer agreement might be expected in view of the agreement between eardrum sound pressures of KEMAR and people shown above and the criteria for selection of the model ear pinna.

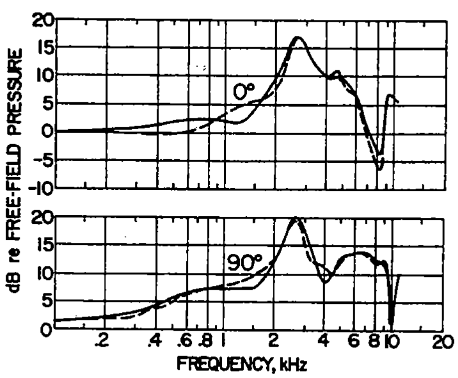


FIG. 10. Eardrum sound pressure in KEMAR showing effect of torso reflections, for sound from front and side. ——with torso; – – – no torso.

D. Body diffraction

The presence of the torso affects pressure in the vicinity of the ear below 2 kHz most noticeably at 0° azimuth. Referring to Fig. 10, the clothed torso alters the eardrum pressure by as much as 3 dB compared to the KEMAR head mounted alone in free field at 1200 Hz. At this frequency, sound waves reflected from the torso arrive out of phase and partially cancel the waves reaching the ear directly. The effective place of reflection results in a sound path difference of 14 cm compared to the sound wave that diffracts around the head. The interference due to reflection from the shoulder can also be seen in the 90° sound incidence curve. In this case, the effective place of reflection creates a path length difference of 11 cm compared to the direct sound.

TABLE IV. Frequencies of pressure maxima and minima at the open ear canal entrance on KEMAR right and left ears. Equivalent length based on $c = 345$ m/sec.

		kHz	Equivalent length (mm)
F_1	right	2.5	34.5
	left	2.5	34.5
F_3	right	6.6	38.0
	left	6.4	39.2
f_1	right	3.7	23.3
	left	3.6	23.9
f_3	right	11.0	23.5
	left	11.3	22.9
f_v	right		
	left	9.0	

The effect of neck length on eardrum pressure is also most noticeably below 2 kHz at 0° azimuth. As shown in Fig. 11, a pressure minimum at 1.3 kHz is observed when the shoulder to tragion height is 17.5 cm. When ear entrance to shoulder distance is varied from 16.3 to 18.8 cm, corresponding to the mean female and male neck lengths, this pressure minimum changes from 1.4 to 1.2 kHz. It is likely that reflections from the chest and shoulder are causing a partial cancellation of sound reaching the ear by a direct path, since the chest or shoulder to ear path difference is approximately $\frac{1}{2}$ wavelength at 1.3 kHz.

E. Need for flesh simulation

The head and torso of KEMAR were fabricated of a hard polyester–fiberglass material. Investigators have frequently made special efforts to simulate soft flesh-like properties of a human being. For example, the Plastisol skin used on the B. Bauer *et al.* manikin has a durometer of 10–15 Shore A and is about $\frac{1}{8}$ in. thick (A. DiMattia, personal cummunication). This material is softer than skin over the mastoid bone area but harder than soft human flesh (e.g., below the cheek bone). Does human flesh have an acoustic impedance low enough to alter pressure in the region of the ear compared to our hard KEMAR head and torso? Clothing applied to KEMAR decreases the magnitude of interference effects such as reflections from the shoulder. These can easily be tailored, then, to meet a desired test condition. Hair, in the form of a wig, shown in Fig. 5, makes the pressure minima at the eardrum at around

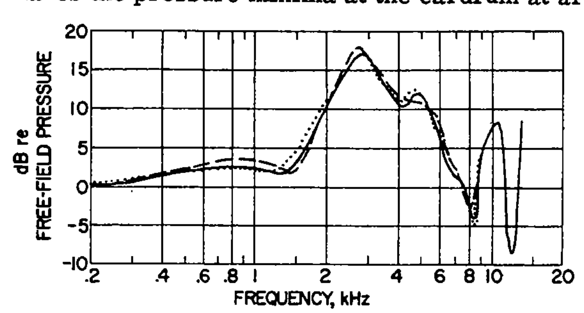


FIG. 11. Eardrum sound pressure in KEMAR showing effect of neck length on torso reflections. ••• 18.8 cm; ——17.6 cm; –·– –16.3 cm.

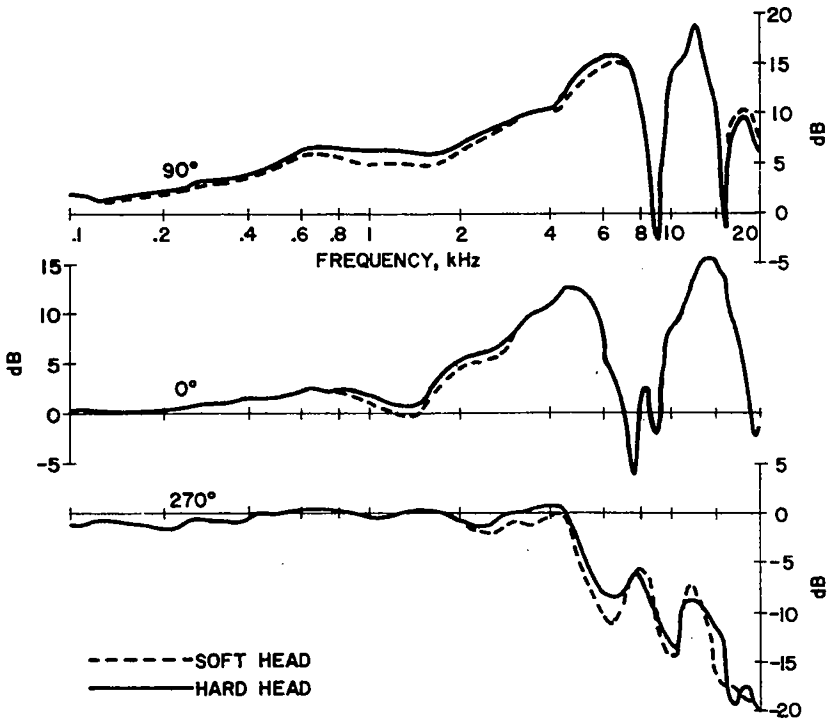


FIG. 12. Comparison of blocked ear entrance sound pressure for a hard fiberglass-reinforced polyester head and a soft sponge RTV head on KEMAR.

10 kHz less deep, but has little effect on eardrum and surface sound pressure at lower frequencies. It isn't easy, however, to alter the surface impedance of a head.

To answer the impedance effect question for the head, a duplicate KEMAR head was cast with General Electric RTV-7 silicone rubber foam. Compared to human flesh (Goldman and Von Gierke, 1961) this material is about twice as compressible, 10 vs 5×10^{-6} cm/dyn for a 1-cm^2 area, and has less than half the effective mass, 0.6 compared to 1.5 to 3.3 g/cm^2. A simple lumped parameter (mass-compliance) model thus predicts that the acoustic impedance of the RTV-7 head is the same order or less than soft human tissue. Blocked ear canal entrance sound pressures on the soft and hard head, for sound incident from 90° azimuth, Fig. 12, shows no more than 1-dB differences at a few frequencies over the range extending to over 8000 Hz. Being hard headed does not affect significantly the sound that will be incident on the ear canal entrance and hence the eardrum.

IV. CONCLUSION

The large amount of published data on the dimensions of adult humans and the acoustic fields around persons permits a good representation of an average human adult with a manikin for purposes of acoustical experiments. The details of the external ear which have been found critical in the listening process, particularly at frequencies of 2000 Hz and above, required additional data to arrive at a reasonable size and shape of auricle. Overall pinna dimensions, concha dimensions, angles of the pinna with the cheek, and angle of the pinna with vertical on the head were used with free-field sound measurements in the concha to determine a preferred external ear design. Because the auricle design is based on a relatively small number of people, future work with the manikin may lead to a preferred shape and/or design, or to the use of several sizes of pinna. To accommodate future external ear design refinements and give the option of experiments with different external ear sizes, the auricle for a KEMAR can be replaced relatively easily without altering the other acoustically important characteristics.

The difference between the male and female average neck length, i.e., the distance from the ear canal to the shoulder, is large enough that it was desirable to account for this variable in any simulation. Other mean male and female anthropometric dimensions for size and shape of the head had differences that corresponded to less than $\frac{1}{10}$ wavelength for frequencies less than 8 kHz, a frequency that was judged to be more than adequate for hearing aid work. The somewhat larger size differences between male and female upper torso may quite reasonably be of concern in some investigations. The larger male size, or female anatomy, can be simulated with padding. The sound pressure at the eardrum when compared to a free field is not critically sensitive to the flesh impedance of the head, a finding that simplifies the fabrication of KEMAR considerably. The recently developed Zwislocki coupler has proved to be useful for synthesizing an average ear canal in the manikin. The evaluation of KEMAR involves many parameters, all of which showed close agreement with various measures and studies of people.

Although for many applications one would be ill advised to design for an "average man" as pointed out by Hertzberg (1972), this anthropometric limitation is not of concern here since KEMAR is intended as a measuring tool which has acoustical diffraction and responses in the middle of the ranges for the adult population. Combining a number of dimensional parameters into the synthesis of a near-median person does give acoustic field interaction like an "average human adult" for frequencies up to 8 or 9 kHz. A more appropriate description is that KEMAR represents a *median individual* in the human adult population.

Hearing aid testing and research are the principal needs to be satisfied with KEMAR; but room acoustic evaluations by Mellert, Von Wilkens, and others, in which electrical signals from the two ears of a "dummy head" were played on "stereo" headphones for a panel of listeners, suggest an important additional use. Elimination of the added variables of control of subject location and the need for involving people who must respond as part of the experiment attest to the desirability of manikins for many acoustic measurements. Our understanding of the role of external ear acoustics will no doubt be enhanced by work with a manikin of this type and, hopefully, can lead to improved sound localization and hearing through better hearing aid design and fitting procedures.

ACKNOWLEDGMENT

The helpful comments and interest of our colleagues is acknowledged. The need for such a manikin was suggested by H. S. Knowles. This paper draws on material included in Industrial Research Products, Inc., Report 20032-1, Nov. 1973 to Knowles Electronics, Inc.

*Present address: Central Institute for the Deaf, St. Louis, Missouri.

Alexander, M., and Laubach, L. L. (1968). "Anthropometry of the Human Ear," AMRL-TR-67-203, Air Force Systems Command, Wright-Patterson AB, OH.

American National Standards Institute (1960). ANSI-S3.3. "Methods for Measurement of Electroacoustic Characteristics of Hearing Aids."

Bauer, B. B., Rosenbeck, A. J., and Abbagnaro, L. A. (1967). "External-Ear replica for Acoustical Testing," J. Acoust. Soc. Am. 42, 204.

Carlisle, R. W., and Mundel, A. B. (1944). "Practical Hearing Aid Measurements," J. Acoust. Soc. Am. 13, 294.

Churchill, E., and Truett, B. (1957). "Metrical Relations Among Dimensions of the Head and Face," WADC Tech. Rep. 56-621, ASTIA Docum. No. AD 110629 (June).

Damaske, P., and Mellert, V. (1969/70). "Ein Verfahren zur richtungstreuen Schallabbildung des oberen Halbraumes uher Zwei Lautsprecher," Acustica 22, 153–162.

Delaney, M. E. (1964). "The Acoustical impedance of human ears," J. Sound Vib. 1, 455–467.

Dreyfuss, H. (1967). *The Measure of Man* (Whitney Library of Design, New York).

Goldman, D., and von Gierke, H. (1961). "Effects of Shock and Vibration on Man," in *Shock and Vibration Handbook*, C. Harris and C. Crede, Eds. (McGraw–Hill, New York), Chap. 44.

Hertzberg, H. T. E. (1972). "Engineering Anthropology," in *Human Engineering Guide to Equipment Design*, H. P. Van Cott and R. G. Kinkade, Eds. (U.S. GPO, Washington, DC), Chap. 11.

Kasten, R., and Lotterman, S. H. (1967). "Azimuth effects with ear level hearing aids," Bull. of Prosth. Res. 10-7, (Spring).

Mellert, V. (1972). "Construction of a dummy head after new measurements of threshold of hearing," J. Acoust. Soc. Am. 51, 1359-1361.

Muldoon, T. L. (1973). "Response Variations of a Microphone Worn on the Human Body," R. I. 7810, U. S. Bureau of Mines, Washington, DC.

Nordlund, N. (1962). "Physical Factors in Angular Localization," Acta Otolaryngol. 54, 75-93.

Shaw, E. A. G. (1966). "Ear Canal Pressure Generated by a Free Sound Field," J. Acoust. Soc. Am. 39, 465-70.

Shaw, E. A. G. (1972). "Acoustic response of external-ear with progressive plane wave source," J. Acoust. Soc. Am. 51, 150 (A).

Shaw, E. A. G. (1974). "Transformation of Sound Pressure from the Free Field to the Eardrum in the Horizontal Plane," J. Acoust. Soc. Am. 56, 1848-61.

Shaw, E. A. G., and Teranichi, R. (1968). "Sound Pressure Generated in an External-Ear Replica and Real Human Ears, by a Nearby Point Source," J. Acoust. Soc. Am. 44, 240-56.

Wilkins, H. (1972). "Kopfbezugliche Stereophonie-ein Hilfsmittle fur Vergleich and beurteilung verschiedner Raumeindrucke (Head related Stereophony-An aid for the comparison and critical examination of different room effects)," Acustica 26, 213–221.

Wiener, F. and Ross, D. (1946). "The Pressure Distribution in the Auditory Canal in a Progressive Sound Field," J. Acoust. Soc. Am. 18, 401–408.

Wonsdronk, C. (1959). "On the influence of the diffraction of sound waves around the human head on the characteristics of hearing aids," J. Acoust. Soc. Am. 31,, 1609–1612.

Yamaguchi, Z., and Sushi, N. (1956). "Real Ear Response of Receivers," J. Acoust. Soc. Jpn. 12, 8–13.

Zwislocki, J. J. (1970). "An Acoustic Coupler for Earphone Calibration," Rep. LSC-S-7, Lab. Sensory Commun., Syracuse U.

Zwislocki, J. J. (1971). "An Ear-like Coupler for Earphone Calibration," Rep. LSC-S-9, Lab. Sensory Commun., Syracuse U.

Chapter 3.

Anthropometric Manikin for Acoustic Research, Supplementary Design Information

M.D. Burkhard
Industrial Research Products, Inc.
Elk Grove Village, IL.

Introduction

Because of lack of space several parameters and acoustical characteristics for the KEMAR manikin were not included in the manuscript reprinted here as Chapter 2. These items are presented in this Chapter in order to provide additional information that might be useful in the application of the manikin.

Concha

Additional parameters not included in the previously published description of the KEMAR manikin concern the shape and volume of the concha region in the ears of the 24 people, who were studied during the process of selection of ear dimensions and appropriate acoustic response. Fig. 3-1 shows the cross-section and volumes for the particular subjects. It can be seen that the shapes are varied, the angle of the opening relative to the head varies remarkably and that the depth of the concha from a reference plane across the pinna is quite different among the subjects. For each person, an ear impression was made and immediately weighed. The volume was computed from the density of the fresh impression material. Then the impression was cut in cross section to obtain the concha shape shown. The cut was made from the front to the back through the greatest distance line that would simultaneously bisect the ear canal entrance portion of the impression.

Fig. 3-2 is a plot of resonance frequency, f_v, (see Fig. 2-4) in the ear response versus concha volume. If this resonance frequency for the ear is correlated highly with the volume of the concha, one would

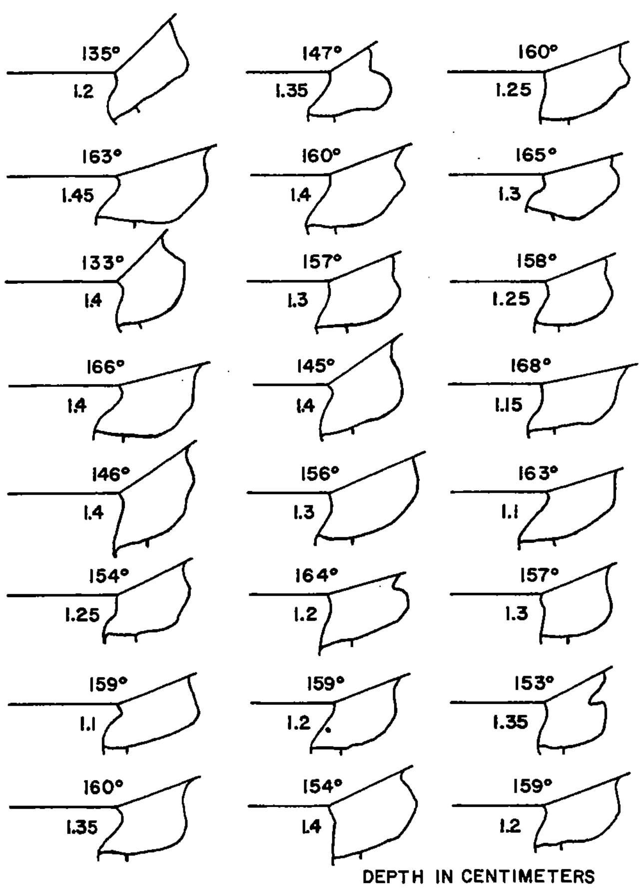


Fig. 3-1. Cross section of the concha of the subjects studied for acoustical characteristics of the external ear. Numbers indicate the angle of the pinna relative to the cheek bone and the depth of the concha to the plane of the ear canal entrance in centimeters.

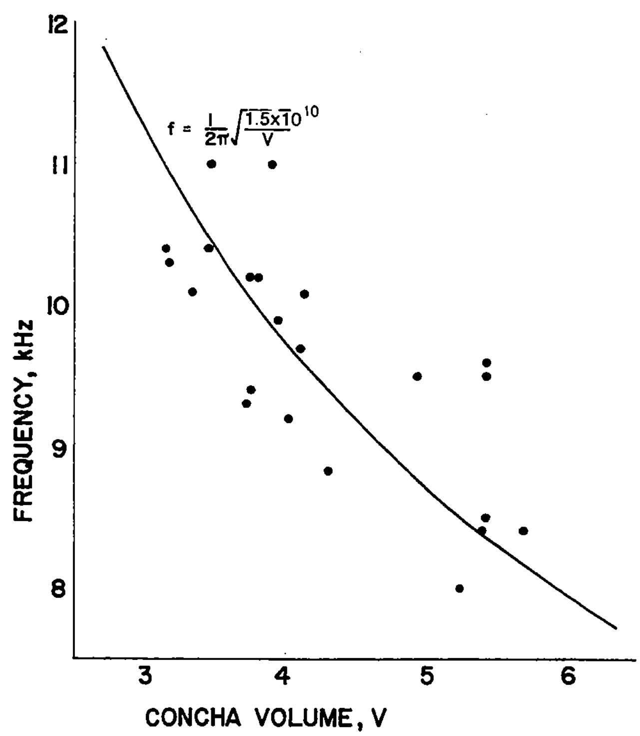


Fig. 3-2. The frequency of resonance, f_V, versus the volume, cm^3, of the concha for subjects used to investigate the acoustics of the external ear.

expect the frequency to be inversely proportional to the square root of the volume. The figure shows that there appears to be a correlation which lends credence to the idea expressed by Shaw that this particular resonance is, indeed, related to the concha portion of the external ear. (cf. discussion in Chapter 2.)

KEMAR Coordinate System

It will be instructive to indicate or describe the orientation and reference conditions for the manikin to facilitate its location in a sound field. Two basic requirements are needed when the coordinate system for the manikin is specified. The first concerns the relationship of the manikin to the external sound sources and sound field. The second concerns the details of alignment of the manikin itself to assure that all of its parts are in proper relationship. Finally, it is helpful to have a system for identifying the location of hearing aid microphones around an ear.

It is our practice to designate zero degree sound source direction as being directly in front of the manikin. As shown in Fig. 3-3, 90° corresponds to the location of the active ear. The active ear is defined as the ear from which an eardrum signal is

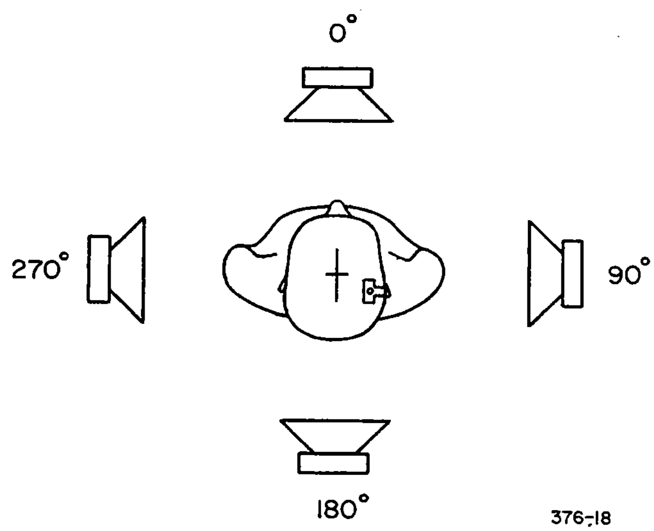


Fig. 3-3. The orientation coordinates for sound sources relative to the KEMAR manikin.

taken. Thus for 90° orientation, the active ear is closest to the sound source. There is no ambiguity when the hearing aid microphone is adjacent to the ear receiving the signal. In the case of a CROS hearing aid, however, a 90° orientation would place the microphone in the head shadow opposite the sound source; the largest output will be observed for a 270° orientation (-90°) when the microphone is closest to the source and the active ear is on the far side of the head (in the shadow). With both ears active, as with binaural hearing aid fitting or binaural recording, the ear closest to the source may be designated as the 90° orientation for anechoic room testing.

Since there is some flexibility built into the manikin, marks can be made to assure various users a common reference condition. It should be repeated that the center of the head is the midpoint on a line through the ear canal openings in the auricles of the manikin. Straight ahead or zero degree orientation, thus, would be on a line perpendicular to the line between the ears. The line between the ears must also be parallel to a corresponding line across the shoulders of the manikin torso. When the head and torso are aligned this way, it is convenient to place marks on the neck to indicate alignment as shown in Fig. 3-4. These alignment coordinates are then extended up to the top of the head, one line for the straight ahead and a perpendicular line that is also parallel to the line between the ears.

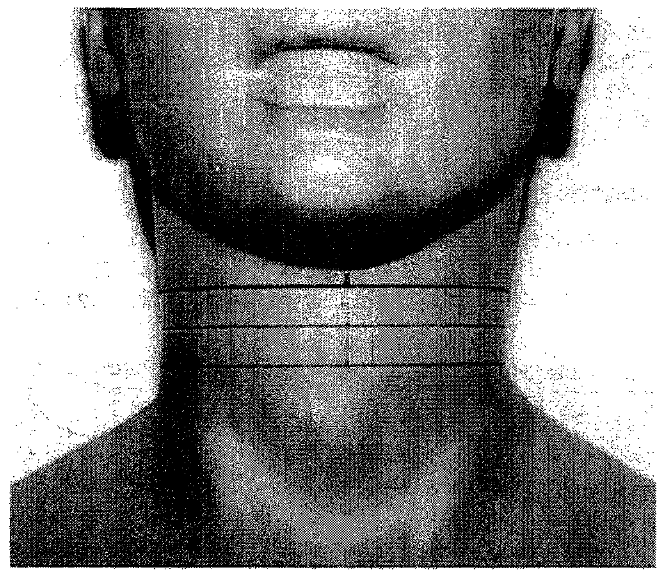


Fig. 3-4. The neck region of the KEMAR manikin showing marks applied for location of the head relative to the torso.

KEMAR Directional Response

Fig. 2-7 in the paper "Anthropometric Manikin for Acoustic Research" shows some KEMAR manikin ear response curves for four different source azimuth angles. Data of similar nature can be shown as polar response plots of the type shown in Fig. 3-5. This plot of pressure at the KEMAR eardrum as a function of rotation angle at three frequencies is very similar to typical data on loudness of sounds as a function of direction such as described by Sivian and White (1933) and Rolls (1973). We will refer to this figure when we discuss measurements of directional hearing aids on KEMAR.

Sound Pressure in the Vicinity of an Ear

When a person is in a sound field he receives sound at his ear from several sources or directions. First there is a main sound wave that comes directly to his ear. Second there will be a sound that propagates around the head to arrive at the ear from the back side. Thirdly there is a sound wave reflected

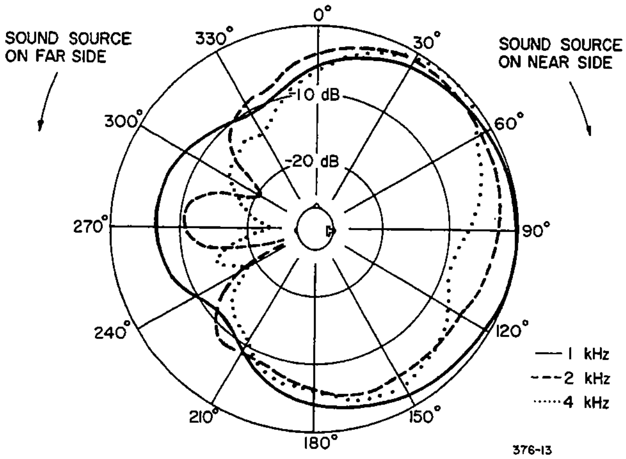


Fig. 3-5. Polar response of the eardrum pressure of the manikin for three frequencies.

from the torso. The drawing in Fig. 3-6 illustrates these three principle paths of sound arriving at an ear, for a single source in front of the manikin in free space such as an anechoic room.

Mr. Madaffari (1974) has measured and reported the pressure around the ear on the KEMAR manikin.

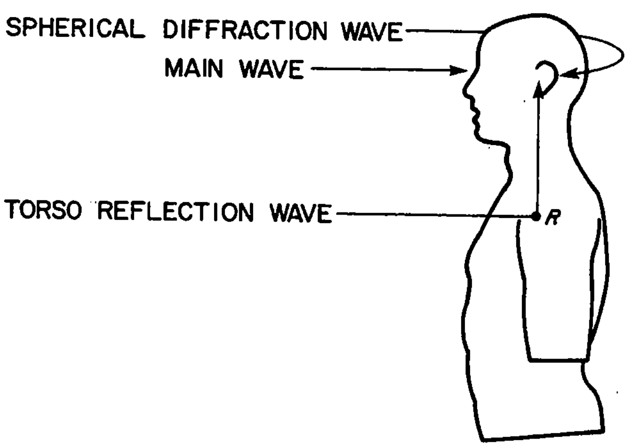


Fig. 3-6. Paths of sound to the ear of a person or the manikin for a sound source directly in front of the manikin or observer.

kin. In the experiment, the manikin was placed in front of an 8" diameter loudspeaker. A grid of measurement locations 2 cm apart was placed on the side of the head as shown in Fig. 3-7. Using the technique of prerecording a drive signal that gives a flat free field at the test position, sans manikin, a miniature microphone with a flat response was located at various points indicated on the grid.

The sound pressure at these locations was then recorded. For the measurements, the sound source is at 0° incidence and the microphone was 5 mm away from the surface of the head. Fig. 3-8 shows four of the twenty-five observations made by Mr. Madaffari. It is evident that the sound pressure at the microphone of a hearing aid varies according to the location of that microphone on the head. This in turn will produce corresponding changes or

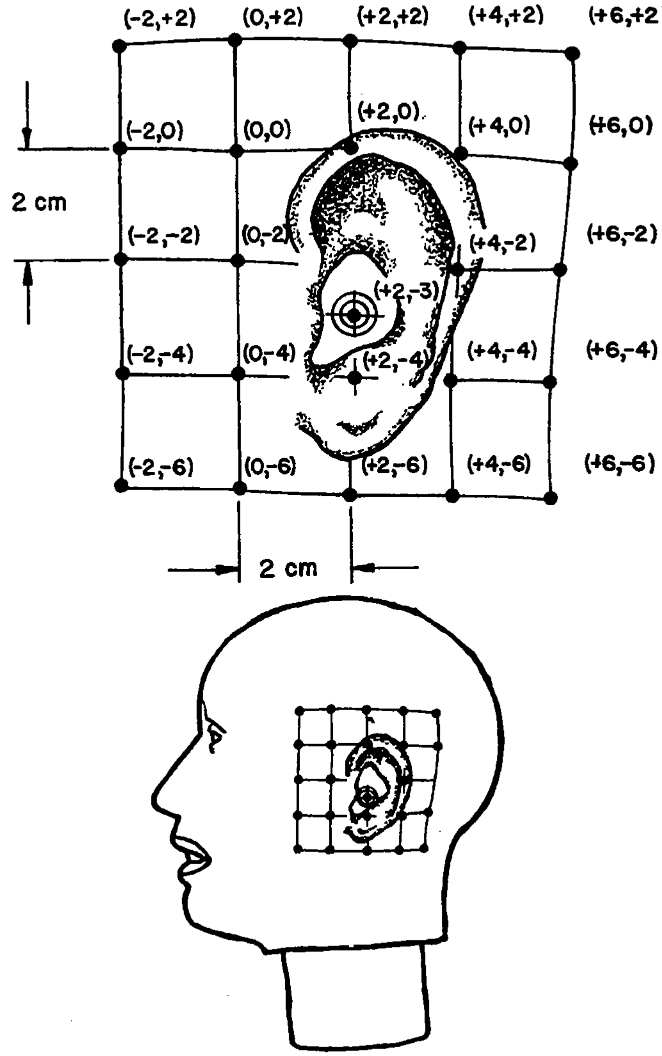


Fig. 3-7. Coordinate system of measurement points around the KEMAR manikin ear.

variations in the input to a head worn hearing aid and the effective or useful gain achieved by a hearing aid wearer as compared to the gain that might be reported for the hearing aid by standard IEC and ANSI hearing aid measurement procedures in a soundbox or a free field.

Using the model for sound around the head, Fig. 3-6, Madaffari derived an empirical formula to describe the sound pressure at various locations on the side of the head, for a frontal incidence source.

$$P = \left[1 + \frac{.08\,(4-X)}{1+(\frac{750}{f})^2}\right] e^{\,ikx\theta} + \left[\frac{8-X}{1+(\frac{750}{f})^2}\right] \frac{e^{\,ik\theta R}}{R}$$

where $R = \sqrt{X^2 + (Y+20)^2}$

$\theta = \frac{2000+f}{4000}$, $f \leq 2000$

$\theta = 1$, $f \geq 2000$

X,Y are horizontal and vertical coordinates of the grid on the side of the manikin head.

Although not exact, the formula has been useful in predicting some of the variations of response, among various hearing aids measured *in situ* on the manikin. The analysis suggested that if a common method for specifying hearing aid input pressures is to be used some form of specification of microphone location on the head is needed.

The sound pressure on the side of the head at two typical hearing aid microphone locations is shown in Figs. 3-9, and 3-10, as a function of sound direction. Pressure on the surface of an earmold in the concha of the KEMAR ear, Fig. 3-9, corresponds to observation location 3 in Fig. 3-8 and has the coordinates of (+2, -3) in the Madaffari measurement. The pressure at the over-the-ear location corresponds approximately to the coordinate (+2, 0) of Fig. 3-7. We see that in addition to the coordinate on the side of the head the direction to the sound source will be an important parameter for general description of an *in situ* hearing aid measurement as well as the general description of the sound field around a persons head.

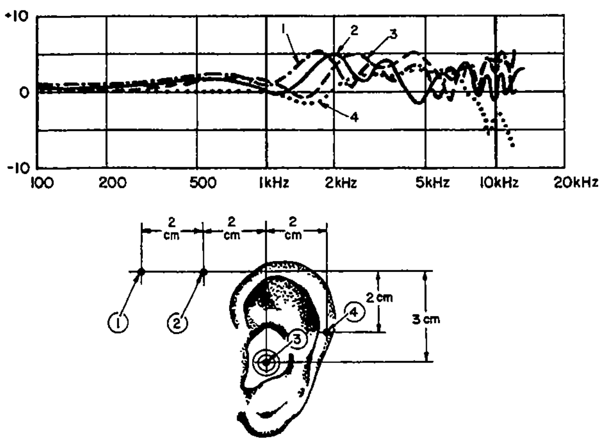


Fig. 3-8. Pressure at various locations around the manikin ear for a constant free field sound pressure for sound source directly in front of the manikin.

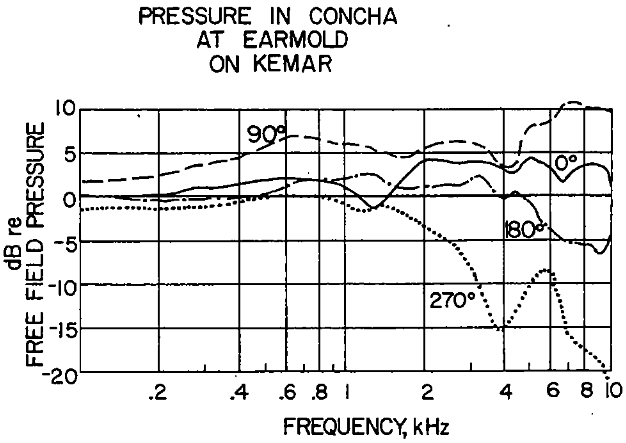


Fig. 3-9. Pressure on the surface of an earmold in the concha of a KEMAR ear when placed in a constant free field sound pressure.

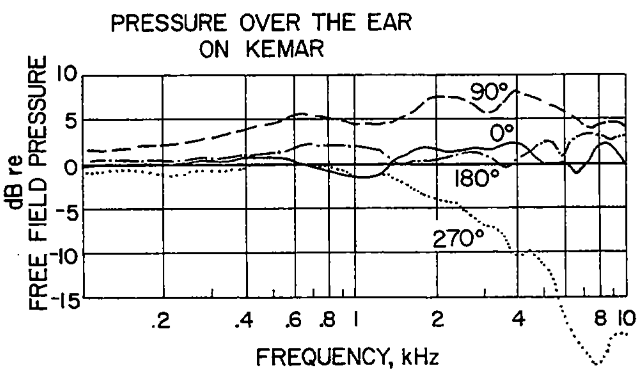


Fig. 3-10. Pressure over the ear of the KEMAR manikin for a constant free field sound pressure.

References:

Madaffari, P.L. (1974) "Pressure Variation about the Ear" Jour. Acoust. Soc. Am. *56* p. S3, abstract B2.

Rolls, E.T. (1973) Polar Frequency Response of the Human Ear, Proc. Physiological Soc. 18P-19P (July)

Sivian, L.J. and White, S.P. (1933) "On Minimum Audible Sound Fields", Journ. Acoust. Soc. Am. *4* pp. 280-321.

Chapter 4.

Gain Terminology

M. D. Burkhard

Industrial Research Products, Inc.

Several in situ measures of hearing aid gain have been discussed or proposed. Many of them have approximately the same meaning although subtle differences have been ascribed to them either by the originators or later users. For a number of years we have suggested the term *Orthotelephonic* response or gain, but the term has not been widely used for hearing aids. (Knowles, 1959; Inglis, 1938)

The term orthotelephonic was devised in telephone communication research to relate the fidelity of sound reproduction by the telephone to face-to-face communication between a normal talker and a normal listener, separated by one meter in an anechoic environment. It is apparent that for telephone communication the listener is remote from the talker and neither one contributes significantly to the perturbations of the sound field in which the telephone listening takes place. In the conventional telephone, the microphone is close to the talkers mouth, and the telephone earphone is against the listener's ear. Because of this direct coupling, any efforts to create equivalence to the face to face condition, must be applied in the electrical and electo-acoustic circuit connections between the two transducers. The system was said to be orthotelephonic if it reproduced the reference face-to-face communication quality.

The same considerations pertain if the naturalness of entertainment or communication earphones is to be evaluated. The free propagation of sound from the environment to the ear drum is replaced by an earphone in contact with the head and ear. There is no diffraction involved and the earphone usually acts as a source with impedance much higher than free space.

By contrast, however, the signals presented to the ear of a hearing aid wearer by his hearing aid are markedly influenced by the perturbations introduced into the sound field by his head and torso. The hearing aid is normally used in face-to-face communication, or in listening in which the user is in the sound field he wishes to hear. A more appropriate term therefore is needed to describe the gain benefit to hearing aid users.

Dalsgaard (1974) used *insertion gain* to characterize the gain provided by a hearing aid *in situ*, the reference condition being the sound in the ear of the wearer sans hearing aid. It is anticipated that the listener with his hearing aid is in the vicinity of or presence of the sound being listened to. Insertion gain (or loss) is a rather commonly used term in communications engineering to describe the effect of introducing an element into an otherwise unchanged system. (IEEE Std 100-1972). The newly introduced element causes a net change in the system as a result of its insertion.

A linquist acquaintance of Killion's when apprised of our interest in a proper term to describe the hearing aid *in situ* gain, suggested a new word—ETYMOTIC. It is derived from two Greek words or word segments: etym, meaning true or real, and otic, meaning ear. Thus one might speak of the etymotic gain or response of the hearing aid where the reference condition would be the unaided sound at the eardrum of an open ear.

A fourth term has been recently used to describe the benefit of wearing a hearing aid; namely, *functional gain* (Pascoe, 1975). Functional gain was used in the Pascoe study to describe the subjectively measured or derived improvement for the hearing aid wearer.

Of the last three terms, insertion gain is a long standing engineering term with rather well defined meaning that can be used unambiguously for describing the change of sound stimulus as measured objectively by probe microphone procedures on persons or by comparison of manikin eardrum sound pressures with and without the hearing aid in place. As used here etymotic would have the same objective measurement meaning. Functional gain would then be reserved for the psychoacoustically derived benefit from the hearing aid as compared to the unaided state. Orthotelephonic would be reserved for the situation in which the listener and his listening aid were not in the presence of the sound source. In all cases the intent of the qualifying term is to indicate the relationship between sounds propagated into an ear naturally and the sound produced at the same place by auxilliary means. In the case of functional gain, we should, perhaps, include stimulation of the nervous system of the inner ear and auditory nerve, so as to include interpretive abilities of the user.

(Editors note: Unfortunately the terms are in a state of evolution so that etymotic, orthotelephonic, insertion, and functional are used for similar measures by the contributors in this proceedings.)

References:

Dalsgaard, S.C. and Jensen, O.D. (1974) Measurement of Insertion Gain of Hearing Aids. Eighth International Congress on Acoustics, London Vol. 1, p 205

IEEE Std. 100-1972. IEEE *Standard Dictionary of Electrical and Electronics Terms.* p. 284. (Wiley Interscience, New York)

Inglis, A.H. (1938) Bell System Tech. Jour. *17,* 373-374.

Knowles, H.S., (1959) HAIC, Allerton House (unpublished)

Pascoe, D. P. (1975) Frequency Responses of Hearing Aids and Their Effects on the Speech Perception of Hearing Impaired Subjects. Ann. Otol. Rhinol, Laryngol. *84* Supplement 23, (Sept., Oct.)

Chapter 5
On Insertion Gain

S.C. Dalsgaard
Research Laboratory for Technical Audiology

Presented in Zurich, March 4, 1976

This is a brief summary of a paper presented at the 8th ICA Congress in London 1974.

The concept of etymotic gain was described by Romanow (1942) and named "insertion gain" by Ayres (1953). Personally, I prefer this name as I feel it more descriptive than etymotic gain.

The insertion gain is more adequate for clinical work than the gain defined by IEC, because it describes the actual amplification the patient is provided with.

Fig. 5-1 illustrates our definition of the insertion gain. We compare the sound pressure level in the

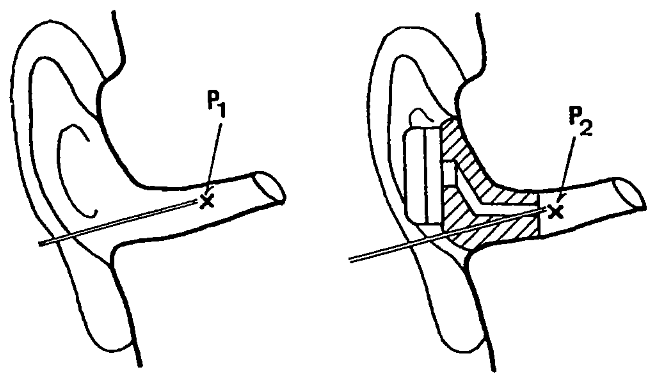


Fig. 5-1. Illustration of the two measurement conditions for determining insertion gain of a hearing aid.

treated ear to the sound pressure level in the untreated ear. We have not chosen the ear drum as a reference, because we want to have a point where we can readily measure the sound pressure. Therefore, we have chosen, as our reference point, a point 5 mm in front of the earmold. Using this definition, we have made measurements of the insertion gain, not on artificial men but on real persons, in order to see how great differences you can expect in real life.

In the experiment we inserted a small probe microphone in the ear canal, as illustrated in Fig. 5-1, and using the recording technique, kept the sound pressure level constant at the reference point. Then we put the probe tube through the earmold, fed the loud-speaker with the recorded signal on the tape and recorded the sound pressure level in the treated ear. In this way we were able to record the insertion gain.

Figs. 5-2 and 5-3 show the measurement results for two types of head worn hearing aids. Fig. 5-2 shows an aid with a microphone pointing downwards and Fig. 5-3 a hearing aid with frontal microphone. The full line is the IEC gain for the hearing

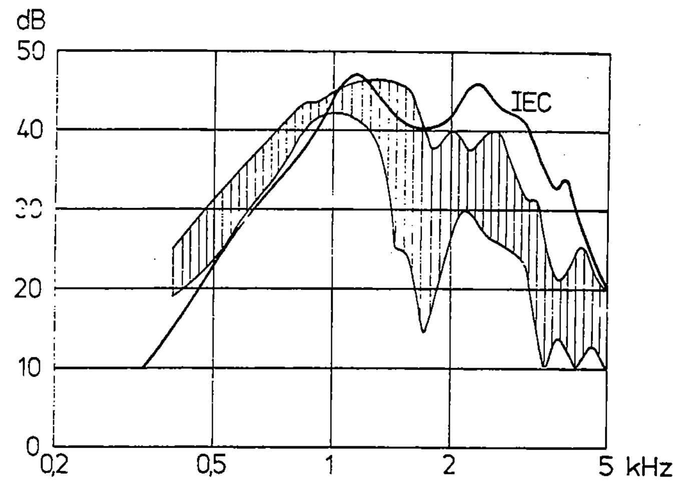


Fig. 5-2. Insertion gain for a behind-the-ear hearing aid, bottom microphone location, on (5) five subjects and the gain measured by IEC methods.

aid and the cross hatched area shows the upper and lower limits for response curves on five different persons. As you will notice there is a very large individual spread in the results. You will also notice the same effect that shows up in Dr. Helle's curves, namely that you get lower insertion gain than the IEC gain at the upper frequencies due to the fact that by putting an earmold into the ear canal you change the natural resonance and hence destroy the natural amplification of the ear canal. Another factor which causes the drop at higher frequencies

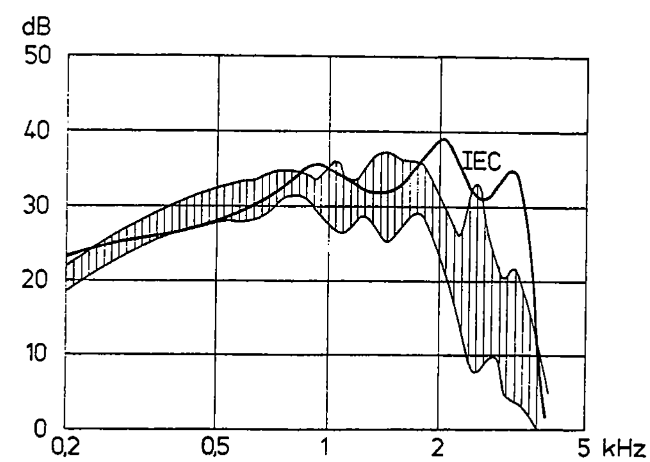


Fig. 5-3. Insertion gain for an over-the-ear hearing aid, frontal microphone location, on (5) five subjects and the gain measured by IEC methods.

is that the canal in the earmold actually used was more narrow than the canal in the 2 cc coupler.

The material has been published later (Dalsgaard & Jensen, 1976).

References:

Ayers, E.W. (1953) A discussion of some problems involved in deriving objective performance criteria for a wearable hearing aid from clinical measurements with laboratory apparatus, Proc. 1st ICA Congress, Delft, 141-143.

Dalsgaard, S.C. & Jensen, O.D. (1974) Measurement of insertion gain of hearing aids, Proc. 8th ICA Congress, London, 205.

Dalsgaard, S.C. & Jensen, O.D. (1976) Measurement of the insertion gain of hearing aids, J.Audiol.Techn. *15*, 170-182.

Romanow, F.F. (1942) Methods for measuring the performance of hearing aids, J.Acoust.Soc.Amer. *13*, 294-304.

Chapter 6.

Frequency Response Of Behind The Ear Hearing Aids Measured On Kemar

Roland Helle

SIEMENS AG,

Medical Engineering Group,

Development Laboratory Electro-Acoustical Sector

Presented in Zurich, March 4, 1976

Introduction

The Recommendation IEC 118 (1959) describes a simple method for evaluating the performance of hearing aids under specified conditions. The properties of a hearing aid are investigated with its microphone being in a free sound field and the output level being measured in the well-known 2 cm^3 coupler according to IEC 126 (1973). This method is well suited for comparing technical data for different types of hearing aids objectively, but as mentioned in the IEC recommendations, it has to be applied carefully as far as actual fittings of hearing aids to hearing impaired persons are concerned.

During the last few years improvements have been made both for the acoustic termination of the hearing aid and for modelling the sound field at the microphone of the hearing aid according to the location of the hearing aid on the body or on the head.

Ear simulators, one of them known as the Zwislocki coupler (1970), imitate the acoustic impedance of the ear canal including the impedance of the eardrum. A life size manikin with the acoustically important dimensions corresponding to the average of adult persons, has been designed that includes a Zwislocki coupler as part of its ear canal. The manikin, named KEMAR, cf. Burkhard and Sachs (1975) has been constructed in a way that hearing aids can be fitted on it as on a living subject. With the KEMAR manikin, the situation of a normal hearing person can be compared to that of a hard of hearing person wearing a hearing aid. Thus, the real gain of a hearing aid, called etymotic gain by Burkhard and insertion gain by Dalsgaard and Jensen (1974), i.e., the difference between the aided and the unaided state, can be determined.

This report starts by describing the standard measurement according to IEC 118, and IEC 126 as a block diagram and then compares it to several methods using the Zwislocki coupler or KEMAR (including a Zwislocki coupler). The output signal to be investigated is the sound pressure level as a function of frequency determined in the coupler. The experiments are carried out in an anechoic chamber by means of a comparison method, the reference signal being the sound pressure level at the location of the KEMAR manikin, sans KEMAR.

Four types of behind-the-ear hearing aids having the microphone port located at different places on the case are examined. Their frequency responses determined according to the standard method (2 cm^3 coupler) are compared to the performance on KEMAR including the calculated values of the etymotic gain.

1. Principle and Measurement Procedure

Whenever the transfer function of an acoustic device has to be investigated the reference signal at the input has to be correctly measured within the sound field. As it is impossible to have at the same time at the same place both the microphone of the device to be examined and the microphone for the reference signal, two different principles have been introduced for the measurement of the reference signal: the substitution method and the comparison method.

When the substitution method is applied, the reference signal is measured before the device under examination is brought into the sound field and auxiliary values, e.g. the voltage at the loudspeaker, are stored and then adjusted at the predetermined value with the device then brought into the sound field. The comparison method takes advantage of the symmetry of the sound field and allows the reference signal to be picked up at the same time at a different place.

All of the measurements described in this report have been carried out by the comparison method or by a modified comparison method.

1.1 Standard Procedure and Measuring Technique with KEMAR.

The different kinds of measuring setup are explained by the block diagrams in the upper and by the schematic transfer functions in the lower part of Fig. 6-1. The block diagrams show the free sound field in the anechoic chamber, the loudspeaker creating this sound field within a certain area, the microphone picking up the reference signal and the equipment controlling its amplitude and frequency, the hearing aid with microphone, amplifier and receiver and finally the coupler with the microphone to detect the output signal L_0 to be investigated and recorded as a transfer function.

The schematic transfer functions in the lower part of Fig. 6-1 are derived from actual investigations of a wide-band hearing aid. They hold for an input level L_I = 60 dB at the location of the microphone port in the undisturbed free sound field. The gain of the aid was adjusted to 40 dB at 1 kHz with L_I = 60 dB under standard conditions corresponding to an output level L_{01} = 100 dB (c f. Fig. 6-1, (a)). Thus the gain control is fixed at the same position for the five different measuring methods.

Fig. 6-1(a), represents the standard procedure according to IEC 118, and in Fig. 6-1(b), the 2 cm³ coupler is replaced by the Zwislocki coupler. Fig. 6-1(c) explains the situation where KEMAR is supplied with a hearing aid, whereas the unaided situation is given by the Fig. 6-1(d). The reference signal (indicated by the abscissa of the diagrams) for the transfer function describing those four cases, is given by the input level L_I = 60 dB in the free sound field.

The Fig. 6-1(e) diagram shows the etymotic gain of the hearing aid. The abscissa of this diagram corresponds to a constant value of the sound pressure level $L_{C,Z}$ within the Zwislocki coupler; nevertheless this diagram too has been calculated with the gain of the aid adjusted as before.

The upper part of Fig. 6-1(e) describes the measurement setup for direct recording of the etymotic gain by using KEMAR with two Zwislocki couplers, one ear (the right) having a hearing aid, the other (the left) being unaided with open ear canal to pick up the reference signal at a place corresponding to the eardrum of a normal hearing subject.

The comparison method for measurement of the reference signal had to be modified when KEMAR was used, as shown in Figs. 6-1(c) and (d). The size of the anechoic chamber available was not big enough, to find a suitable place located symmetrically to KEMAR within the sound field. There-

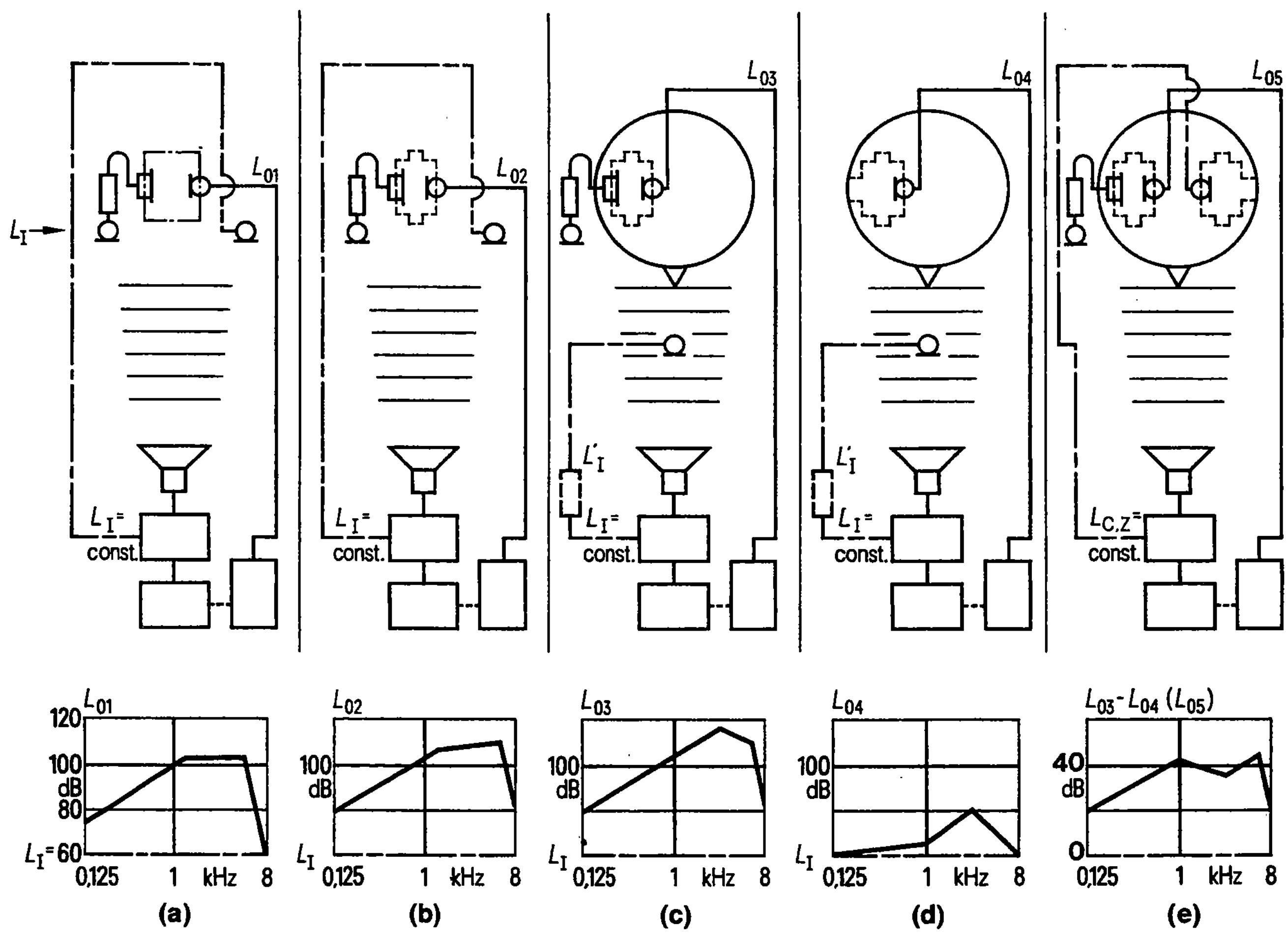


Fig. 6-1. Block diagram for the measuring setup (upper part) and schematic transfer function (lower part). (a) Standard condition according IEC 118, IEC 126. (b) Free sound field and Zwislocki coupler. (c) KEMAR equipped with a hearing aid (aided ear). (d) KEMAR-unaided ear. (e) Determination of the etymotic gain of a hearing aid.

fore the reference microphone was positioned half way between the loudspeaker and KEMAR which were separated by 1.50 m. Thus the comparison method could be applied with acceptable tolerances (c f. section 1.3.).

The schematic diagrams in the lower part of Fig. 6-1 are reproduced from measurements of a wide-band behind-the-ear hearing aid, omitting the peaks and dips of the frequency response in the mid-frequency range.

Replacing the 2 cm^3 coupler by the Zwislocki coupler (Figs. 6-1(a) and 6-1(b)) enhances the output L_{02} compared to L_{01} at low frequencies slightly, as the volume of the Zwislocki coupler is about 1.2 cm^3 for frequencies below 800 Hz and therefore smaller than that of the 2 cm^3 coupler. The further reduction of the effective volume of the Zwislocki coupler at high frequencies and the shape of its volume result in a considerable enhancement of the output level L_{02} at higher frequencies. With the hearing aid fitted to KEMAR, the diffraction of the sound wave around the head becomes effective and therefore the level L_{03} exceeds L_{02} in the mid-freqency range.

L_{03} can be considered as being equivalent to the sound pressure level at the eardrum of a hearing impaired person, equipped with a hearing aid fitted through an ear mold to its closed ear canal. Unfortunately the hearing impaired person does not get the benefit of the full level difference L_{03}- L_I as the sound pressure level L_{04} at the eardrum of a normal hearing subject exceeds the level L_I in the free sound field, too. This transformation is caused by resonances of the open ear canal and by sound diffraction around the head and within the concha.

Thus the real gain, called etymotic gain, of the hearing aid is given by the difference of the sound pressure level at the place of the eardrum in the aided compared to the unaided situation. On account of the resonance peak at about 3 kHz in the unaided ear, the etymotic gain drops especially in the frequency range between 2 kHz and 4 kHz drastically below the level L_{03}.

Within this report measurements are presented for the output level L_{01}according to the standard procedure, for L_{03} the aided and $L_{04}$the unaided situation on KEMAR. The etymotic gain is calculated as level difference L_{03}-L_{04}.

1.2 Measuring setup

The measurements were carried out in an anechoic chamber designed for a low frequency limit of 250 Hz. The size of the chamber was 2.15 x 2.15 x 2.00 m^3. The reference microphone was a B&K Type 4131 1" free field corrected condenser microphone. Transfer functions are reported for the frequency range extending from 300 Hz to 8 kHz. Sound pressure level is stated in dB re. $2x10^{-5}$ N/m^2. KEMAR was used without a jacket, without a wig, and with neck length of 17.6 cm (1 ring). The sound wave was coming from the front (0° azimuth). Measurements were taken for the right ear of KEMAR. The distance from KEMAR to the loudspeaker was 1.5 m, to the reference microphone 0.75 m.

The acoustic tubing following the hook was the same for all the aids investigated: 25 mm flexible tube of 2 mm inner diameter between the tip of the hook and the entrance of the adapter. Both for the 2 cm^3 coupler and for the Zwislocki coupler the length of the adapter (two different makes) connected to the coupler was 18 mm with an inner diameter of 3 mm, according to IEC 126 for the 2 cm^3 coupler. The remaining height of the main volume of the Zwislocki coupler was 12.7 mm with the adapter mounted into it.

The four different types of behind-the-ear hearing aids to be examined were all equipped with electret omnidirectional microphones, but the microphone ports were situated at different locations on the case. Type A had a frontal microphone through the hook, the sound inlet of Type B was at the bottom of the aid, thus looking in a downward direction. The sound inlet of Type C was in the frontal region near the hook but through the upper side of the case. Type D had the microphone port near the hook but through the lower side of the case.

The position of the gain control remained unchanged throughout the complete investigation. The gain was adjusted under standard conditions (2 cm^3 coupler) at 1 kHz at a value of 40 dB as the difference between the output level L_{01} = 100 dB and L_I = 60 dB. None of the aids was driven close to saturation.

1.3 Tolerances

When using KEMAR, the comparison method had to be modified as already described in section 1.1. In order to estimate the error of the measuring setup, two experiments were carried out.

First, with voltage applied to the loudspeaker, the sound pressure level was registered by the reference microphone simultaneously with the sound pressure level where the KEMAR was to be placed (center of its head) but, without KEMAR. This experiment was to find out whether there are deviations from the expected sound pressure level. Then, as a second step, KEMAR was installed at its place and with the same voltage as before the sound pressure level was recorded again by the reference microphone. This experiment was made in order to check the influence of KEMAR on the sound field around the reference microphone.

With those two measurements taken together, it was evident in the frequency range from 0.3 to 8 kHz that the sound pressure level actually presented to KEMAR coincided within ±2.5 dB with that value assumed according to the adjustment of the equipment. The limits were ±2 dB with the exception of the boundaries of the frequency range stated and one additional value around 1700 Hz.

These tolerances are thought to be accurate for testing the performance of hearing aids on KEMAR.

2. Sound Pressure Transformation in the Unaided Ear

The sound pressure level at the eardrum of the open ear canal is different from the level in the free sound field as already mentioned before. Fig. 6-2 shows the output, L_{04} measured by the microphone inside the Zwislocki coupler installed inside the

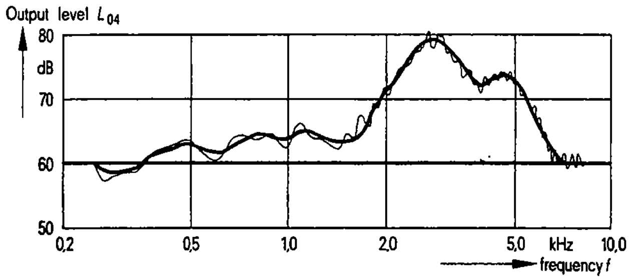


Fig. 6-2. Sound pressure transformation in the unaided ear. Input level L_I = 60 dB SPL. Output level L_{04} measured within the Zwislocki coupler mounted into the head of KEMAR.

head of KEMAR caused by a sound pressure level of L_I = 60 dB in the free sound field. The thin curve represents the actual recording, the thick line is the averaged curve as used for the following calculations of the etymotic gain. The ripple of the actual recording is primarily due to reflections at the grid on the bottom of the anechoic chamber. The smoothing of the average curve at the lower frequencies is introduced for balancing the systematic error described in Section 1.3.

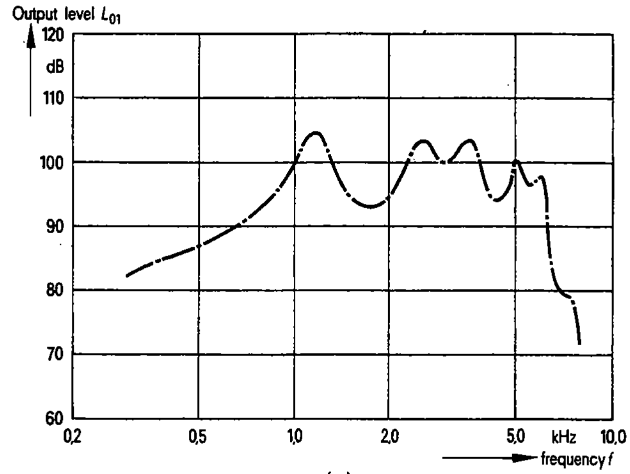
The curve of Fig. 6-2 coincides, within the tolerances of ±2.5 dB (as stated in Section 1.3), with the results given by Burkhard and Sachs (Fig. 2-11). The most prominent deviations are in the 750 Hz to 1500 Hz range where the values of Fig. 6-2 exceed those of Fig. 2-11 by about 2 dB, the same difference occurs again at the maximum near 2.7 kHz and between 4 kHz and 5 kHz.

3. Performance of a Wide-Range Hearing Aid

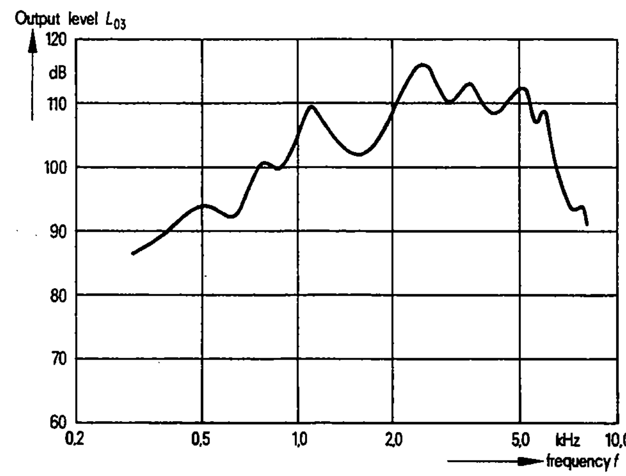
The performance of a hearing aid measured according to the different methods introduced in Fig. 6-1 has been so far described by schematic transfer functions. In this section actual recordings of the frequency response of a wide-range hearing aid, Type C (cf. Section 1.2.), are presented.

Fig. 6-3(a) shows the output level L_{01} measured in the free sound field with the 2 cm³ coupler at an input level L_I = 60 dB. The wide-range character is evident, as the HAIC upper frequency limit extends to 7600 Hz.

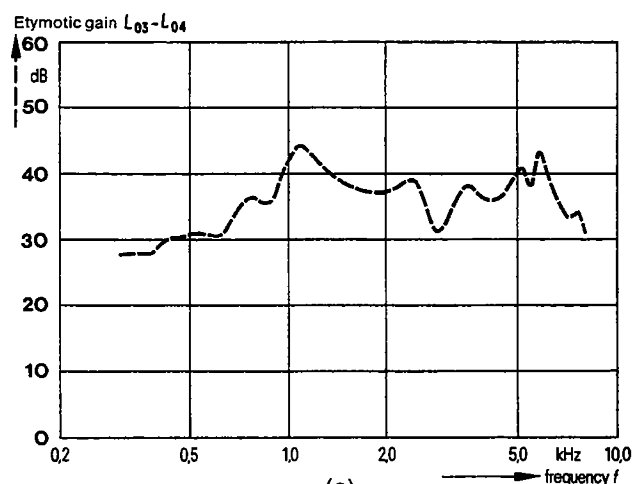
For the same position of the gain control Fig. 6-3(b) shows the level L_{03} recorded with the hearing aid fitted to the right ear of KEMAR. The resonances of the curves shown in Fig. 6-3(a) and (b) differ only slightly, as the acoustic termination of the hearing aid consists of the same tubing plus adapter followed by two different couplers. At low



(a)



(b)



(c)

Fig. 6-3. Frequency response of a wide-range behind-the-ear hearing aid (Type C, cf. Fig. 6-6). Input Level L_I 60 dB SPL, position of gain control to give 40 dB gain at 1 kHz under standard condition. (a) Output level L_{01} measured according to Standard condition IEC 118, IEC 126 (2 cm³ coupler) (b) Output level L_{03} measured with the hearing aid on KEMAR (c) Calculated value L_{03}-L_{04} for etymotic gain.

frequencies up to about 1 kHz L_{03} exceeds L_{01} by about 6 dB, in the mid frequency range L_{03} is up to 10 dB higher than L_{01} whereas this difference amounts to 15 dB at 7 kHz to 8 kHz.

The etymotic gain, $L_{03} - L_{04}$ of the hearing aid is calculated by means of curves in Figs. 6-2 and 6-3(b) and shown in Fig. 6-3(c). Between 650 Hz and 8 kHz the etymotic gain does not fall outside tolerance limits of about ± 6 dB. The etymotic gain curve, for the frequencies examined, never exceeds the curve for L_{03}.

The level L_{01} is significantly higher than the etymotic gain curve only in the frequency range around 3 kHz where the resonance peak of the unaided ear occurs.

4. Comparison of Four Different Types of Behind-The-Ear Hearing Aids.

All aids investigated are of the behind-the-ear type. The position of the microphone port was different, however, for each (cf. Section 1.2.). The results for the frequency response under standard conditions (L_{01}), on KEMAR (L_{03}) and for the etymotic gain (L_{03}-L_{04}) are given in Fig. 6-4 (Type A, frontal microphone through the hook), Fig. 6-5 (Type B, microphone at the bottom), Fig. 6-6 (Type C, microphone in frontal region through upper side of case) and Fig. 6-7 (Type D, microphone near hook through lower side of case). Thus the 3 curves as shown in Fig. 6-3(a), (b), and (c) are now summarized in one diagram.

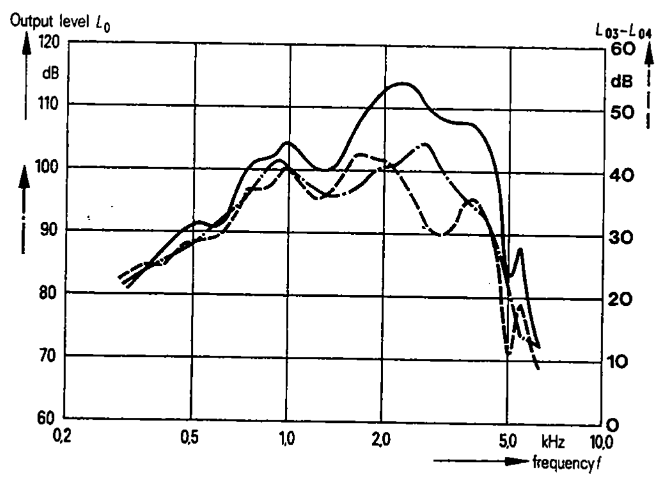


Fig. 6-4. Frequency response of a hearing aid with Type A Microphone Location. (frontal microphone through the hook). The code for the various curves is the same as in Fig. 6-3. Input level, L_I = 60 dB SPL, position of gain control to give 40 dB gain at 1 kHz under standard conditions.

5. Discussion

The discussion is based on the diagrams shown in the Figs. 6-4, 6-5, 6-6 and 6-7. Again, it has to be kept in mind that the reference signal for the transfer function under standard conditions (L_{01}, dash-dotted curve) and on KEMAR (L_{03}, solid

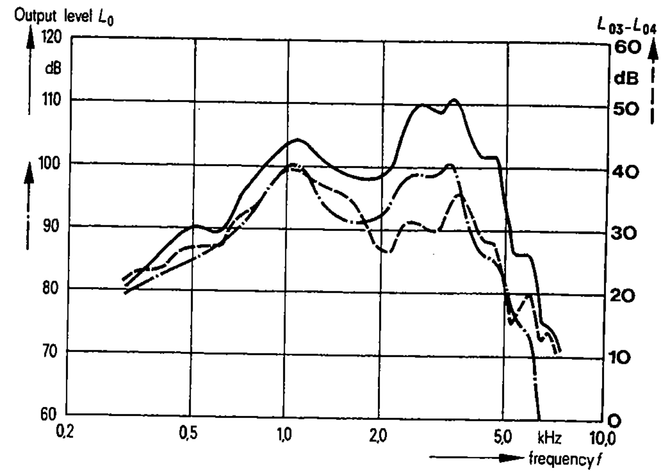


Fig. 6-5. Frequency response of a hearing aid with Type B Microphone Location. (microphone at the bottom of the case). The code for the various curves is the same as in Fig. 6-3. Input level, L_I = 60 dB SPL, position of gain control to give 40 dB gain at 1 kHz under standard conditions.

curve) is the sound pressure level L_I in the free sound field, whereas the etymotic gain (L_{03} - L_{04}, dashed curve) is related to a constant value of the sound pressure level $L_{C,Z}$ at a place corresponding to the eardrum.

Comparison of the 2 cm³ coupler frequency response L_{01} for Type C (Fig. 6-6) and D (Fig. 6-7) reveals that these curves do not deviate by more than ± 2.5 dB below 5.5 kHz. The curves determined on KEMAR for L_{03}, however, differ considerably more if they are compared to each other, especially around 3 kHz and above 5 kHz. Both hearing aids have the microphone port near the hook but at different sides of the case. Thus, these two measurements show that it is not feasible to transfer 2 cm³ coupler curves into KEMAR curves by a simple

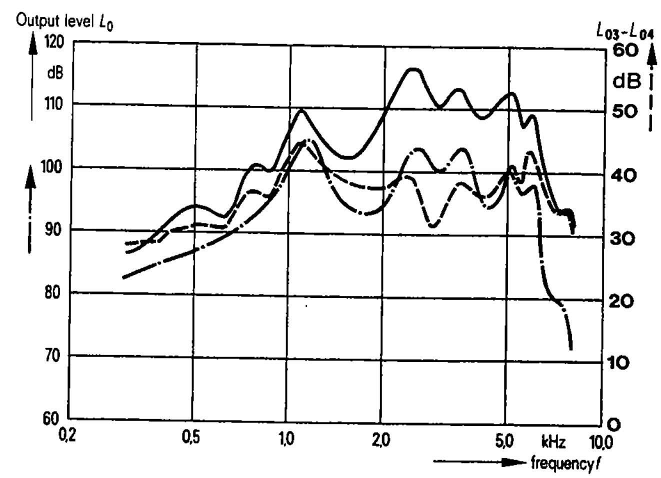


Fig. 6-6. Frequency response of a hearing aid with Type C Microphone Location. (microphone in frontal region through upper side of case). The code for the various curves is the same as in Fig. 6-3. Input level, L_I = 60 dB SPL, position of gain control to give 40 dB gain at 1 kHz under standard conditions.

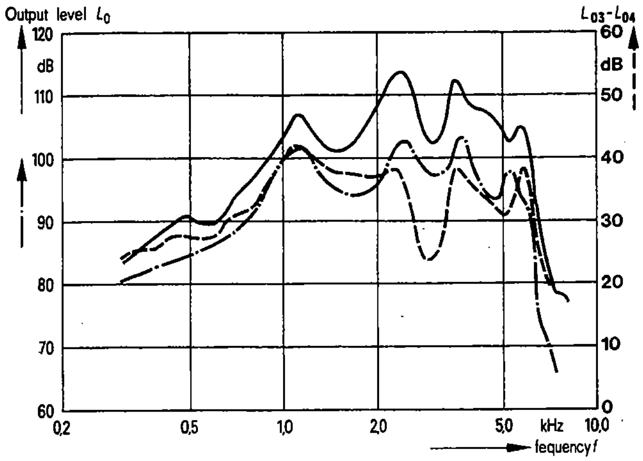


Fig. 6-7. Frequency response of a hearing aid with Type D Microphone Location. (microphone near hook through lower side of case). The code for the various curves is the same as in Fig. 6-3. Input level, L_I = 60 dB SPL, position of gain control to give 40 dB gain at 1 kHz under standard conditions.

transformation rule, not even for rather similar frequency response curves L_{01} and microphone ports which are at adjacent places. Exact transformation rules can only be stated for a specific design of a nearing aid.

The output level L_{03} on KEMAR exceeds within the frequency range examined, the output level L_{01} measured according to the standard conditions on the 2 cm³ coupler. The explanation has been given in Section 1.1. However, it is evident that the difference, L_{03} - L_{01}, in the frequency range of 1500 Hz to 2500 Hz is significantly smaller (up to 8 dB) for the hearing aid Type B than for the other types. The equivalent is true for the etymotic gain. This deviation from the performance of the other types is related to the location of the microphone port, which is at the bottom of the case for Type B. The same finding could be verified for two other types of hearing aids (not reported in this paper) having the microphone port also in the rear part of the case.

Thus the advantage of obtaining higher gain values under standard conditions by separating the microphone from the receiver tube as far as possible to avoid acoustical feedback cannot be realized as a corresponding advantage showing up in the etymotic gain.

The last example has clearly demonstrated that KEMAR allows investigating the influence of different locations of the microphone port on the case of hearing aids.

Comparing the etymotic gain curves, L_{03} - L_{04}, to the 2 cm³ coupler curves for the aids examined, the following summary can be drawn: The etymotic gain exceeds the output level L_{01} of the 2 cm³ coupler at frequencies below 500 Hz by about 3 dB as the volume of the Zwislocki coupler is smaller than 2 cm³. Between 500 Hz and about 1200 Hz both curves have very similar values. In the range of 1200 Hz to about 2000 Hz the etymotic gain exceeds L_{01} up to 4 dB, as the effective volume of the Zwislocki coupler decreases for these frequencies and as the resonance peak of the open ear canal (cf. Fig. 6-2) is not yet reached. Between 2000 Hz and 4000 Hz, the etymotic gain is smaller than the output of the 2 cm³ coupler (8 dB in the average, 13 dB maximum value). For frequencies above 4000 Hz the small volume and the shape of the Zwislocki coupler result in etymotic gain values exceeding those of the 2 cm³ coupler. To investigate this high frequency range properly, wide-range or high-tone hearing aids are required.

7. Concluding Remarks

The introduction of ear simulators, e.g. the Zwislocki coupler, allows the hearing aid to be terminated by the correct acoustic load. The influence of different makes of ear molds can be examined correctly and the performance of hearing aids can be tested up to high frequencies in accordance with the natural situation. The design of a life size mankin offers an opportunity for including sound diffraction of the body and the head in the performance tests of hearing aids. KEMAR, being the combination of both, turns out to be a valuable tool for the development of hearing aids and for testing them under conditions simulating the actual fitting process. Nevertheless, one should keep in mind that KEMAR is an acoustic average, that it is never-the-less satistically accurate, and that hearing aids are fitted on individuals. Thus KEMAR may help to improve the guidelines.

Summary

KEMAR, a life size manikin, has installed an ear simulator (Zwislocki coupler) to imitate the natural ear. According to its design, hearing aids can be fitted in the same way as to living subjects and investigated relative to their real gain, called etymotic gain or insertion gain. The standard procedure according to IEC 118, IEC 126 is compared to several methods using KEMAR, always applying a comparison method for the determination of the reference signal in the sound field. Measurements including calculated values for the etymotic gain are reported and discussed for four different types of behind-the-ear hearing aids. The results clearly demonstrate the influence of different positions of the microphone port.

References

Burkhard, M.D. and Sachs, R.M. Anthropometric Manikin for Acoustic Research. J. Acoust. Soc. Am., Vol. 58 (1975), 214-222 (Reprinted in this proceedings Chap. 2.).

Dalsgaard, S.C. and Jensen, O.D. (1974) Measurement of Insertion Gain of Hearing Aids, Eighth International Congress on Acoustics, London, Vol. 1, 205.

IEC-Publication 118 (1959) Recommended Methods for Measurements of the Electro-Acoustical Characteristics of Hearing Aids. Bureau Centr. de la Comm. Electrotechn. Int., Geneve, Suisse.

IEC-Publication 126 (1973) IEC Reference Coupler for the Measurement of Hearing Aids Using Earphones Coupled to the Ear by Means of Ear Inserts. Bureau Centr. de la Comm. Electrotechn. Int., Geneve, Suisse, second edition.

Zwislocki, J.J. (1970) An Acoustic Coupler for Earphone Calibration Rep. LSC-S-7, Laboratory of Sensory Communication, Syracuse University, Sept.

Chapter 7.

Acoustic Pressure Field Alongside A Manikin's Head With A View Towards *In Situ* Hearing-Aid Tests

G.F. Kuhn and E.D. Burnett
Institute for Basic Standards,
National Bureau of Standards

The following article, reprinted with permission from the Journal of the Acoustical Society of America, formed the basis of a presentation by Dr. Kuhn at the conference April 5, 1976 in Washington, D.C.
Reference to figures in this Chapter, elsewhere in the proceedings, will be designated with the prefix 5—.

The frequency responses of hearing aids measured in a free field differ from those measured on the head of a person or a manikin due to the scattering of the sound by the head and the torso. In order to compare and interpret the response of hearing aids located on the head at various frequencies it is necessary to know precisely the spatial pressure distribution. The amplitude and phase of the acoustic pressure were measured alongside a manikin's head in increments ranging from 2 to 5 mm with frontal sound incidence . The acoustic driver was located in front of the manikin at distances of 1.0 and 3.5 m from the ear-canal axis. The test frequencies were the octave band center frequencies from 0.5 to 4.0 kHz and the third-octave band center frequencies from 4.0 to 8.0 kHz. The sound pressure level varies smoothly, as a function of position, alongside the head for frequencies equal to or less than 2.0 kHz. At frequencies equal to or greater than 4.0 kHz the pressure level changes rapidly with position. Particulary severe pressure minima were found to exist around the pinna at 6.3 and 8.0 kHz. The smoothing effect of test signals using pink noise of 6% and 29% bandwidth on the acoustic pressure variation alongside the head and behind the pinna is also shown.

PACS numbers: 43.66.Ts, 43.66.Yw

INTRODUCTION

In the past, the gain, the frequency response, the saturation level, and the distortion in hearing aids have been measured between 0.2 and 5.0 kHz in an approximate free field using a 2-cm^3 coupler (ANSI, 1976; IEC, 1959). It can be expected that in the near future the useful frequency range of some hearing aids will be extended to approximately 8 kHz and that the tests will be done on a manikin which simulates the actual wearer. It is well known that the head and torso diffract the incident sound causing the sound pressure around the head, where hearing aids are typically placed, to vary considerably with position and with frequency, and to differ from the free-field pressure (see, for example, Wiener, 1947a; Wiener, 1947b; Rzhevkin, 1963; Lybarger and Barron, 1965; Burkhard and Sachs, 1975; and Kuhn, 1976). In order to test and compare the performance of different hearing aids (see, for example, ANSI, 1976 and Veterans Administration, 1976) at various locations alongside the head and around the pinna, it is necessary to know the amplitude and phase of the sound pressure distribution around the wearer's head. Also, knowledge of the pressure distribution is useful (1) to estimate the "gain" of pressure at the hearing aid microphone relative to the free-field pressure, (2) to determine the expected pressure difference due to the uncertainty in the hearing aid placement from one test to the next, (3) to find locations where the acoustic pressure level is smooth or rapidly changing with frequency in order to choose a location where reliable hearing aid response measurements can be made, and (4) to be able to determine and avoid locations where sudden and rapid changes in the phase may occur when hearing aids with two microphone ports are used.

Several past investigations have concentrated on comparing the sound pressure level produced at a few specific locations in the vicinity of the ear for a range of angles of incidence (angle between the sound source and the listener's median plane). Lybarger and Barron (1965) used four microphone locations on four subjects in an anechoic room, sweeping the sound frequency from 0.15 to 7.0 kHz. Temby (1965) used ten subjects, five microphone locations, and white noise with 6% bandwidth centered at 800, 1600, 2400, 3200, and 4000 Hz in a large (nonanechoic) room to measure the sound pressure level over a range of angles of incidence. Olson and Carhart (1975) placed a forward- and backward-facing over-the-ear hearing aid microphone on six subjects and on a manikin's head. They measured the sound pressure as a function of the angle of incidence in a non-anechoic chamber using a 100-Hz-wide random noise, swept from 200 to 5000 Hz. Madaffari (1974) used an anthropomorphic manikin (Burkhard and Sachs, 1975) to measure the acoustic pressure as a function of frequency at 25 locations centered about the ear that were spaced 2 cm apart and 5 mm from the head surface. These measurements were made with frontally incident sound in an anechoic room using a continuous frequency sweep from 0.1 to approximately 12 kHz. The results show that the sound pressure near the head varies smoothly with frequency below approximately 2 kHz. Above 2 kHz, the pressure varies more rapidly with position and frequency as the frequency increases. Theoretical analyses of the pressure distribution around rigid spheres (see, for example, Schwarz, 1943; Wiener, 1947b; and Rzhevkin, 1963) also predict a slowly changing pressure distribution at low frequencies and larger changes in the spatial pressure distribution at high frequencies. Generally, (Schwarz, 1943, Fig. 2 and Table I) the pressure magnitude decreases monotonically from the irradiated pole, $\theta=0°$, to a position about 120° towards the shadowed pole of the sphere. The difference in sound pressure level, at approximately 6.0 kHz, between the $\theta=30°$ and the $\theta=105°$ position is 5.5 dB. No relative pressure minima are predicted between $\theta=0°$ and 120°. (The measurements of this investigation would lie between approximately 30° and 105° on an equivalent sphere whose perimeter is equal to the manikin's head perimeter.)

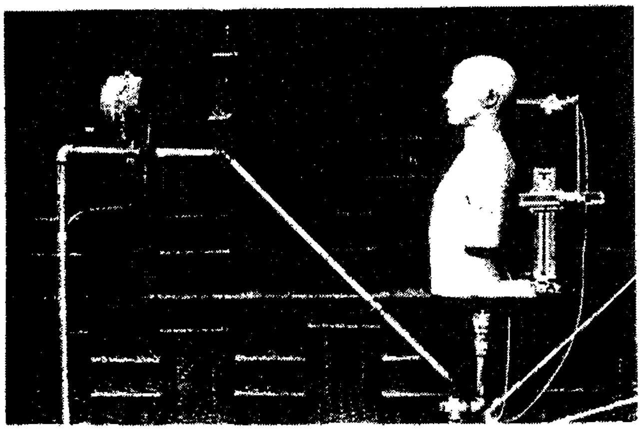


FIG. 1. The experimental configuration in the anechoic room.

Since the head is not perfectly spherical or spheroidal and since it has several protrusions and identations such as the eyes, nose, mouth, and pinnae, and since the torso causes additional scattering, analytical predictions of pressure at the surface of the head can only be approximate. The purpose of this investigation is to provide curves of the sound pressure levels alongside the head relative to the free-field incident pressure. For linear hearing aids, curves of this type may be used directly to convert a free-field hearing aid response to the (manikin-equivalent) response of a head-worn hearing aid. The hearing aid is usually small compared to the acoustic wavelength so that additional diffraction effects due to the hearing aid will be small. In order to have more precise corrections than those already cited, the pressure measurements were made at 348 positions in increments ranging between 2 and 5 mm.

I. THE EXPERIMENT

The measurements of the amplitude and phase of the pressure were made on the manikin (Burkhard and Sachs, 1975) in the upright position in a 425-m³ free-volume anechoic room as shown in Fig. 1. The mouth of the acoustic driver shown to the left on the figure is 49 mm in diameter and was placed either at 1.0 or at 3.5 m from the ear-canal axis.[1] Both the driver and the ear-canal axis lay in the horizontal plane. A "¼-in." microphone placed on the axis of the mouth of the driver was used as a feedback microphone to maintain the same pressure at this location for all frequencies. A "½-in." microphone with a 9.3-cm long, 2-mm internal diameter probe, filled with damping material, was used to measure the pressure alongside the head. Since the pressures around the head were normalized to the free-field incident pressures at each frequency, it was not necessary to calibrate the microphone probe. Initially, some foam was wrapped around the microphone and preamplifier but later removed since it was found to have no effect on the pressure near the head surface. The microphone probe was mounted on x–y–z coordinate mechanical slides which could be adjusted to a resolution of 0.01 mm. The slides themselves were mounted behind the manikin, as shown in Fig. 1, in its acoustic shadow to minimize the effect of the scattered pressures on the measurements.

The incident free-field pressure was measured with the microphone probe at a point vertically above the ear-canal axis on contour 2 (see Fig. 2), with the manikin removed. The measured pressures were normalized to this free-field pressure and converted to pressure levels.

The shape of the head along the measurement contours shown in Fig. 2 (but on the opposite side of the head) was mapped out by noting the appropriate x–y–z coordinate of the tip of the microphone probe when it just came in contact with the head surface. The head surface is

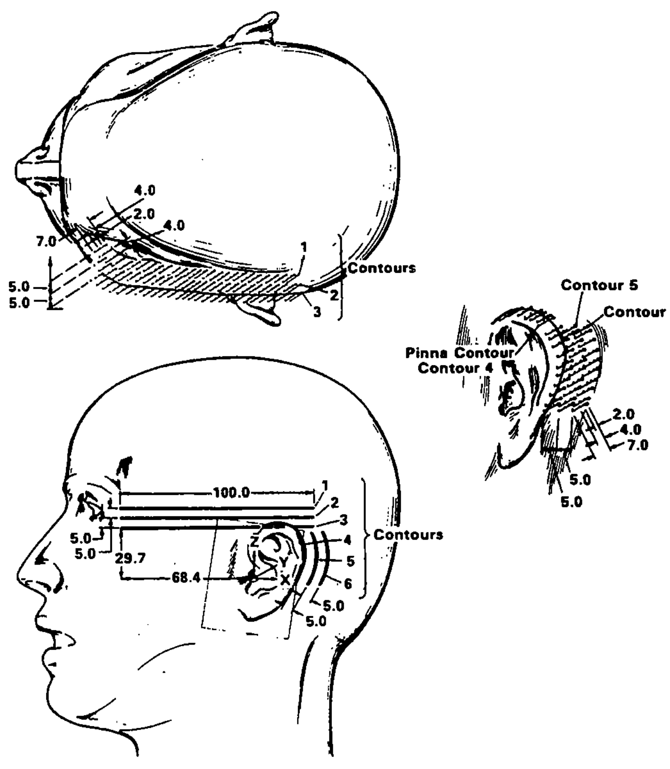


FIG. 2. Measurement contours on the manikin's head (all dimensions are in mm).

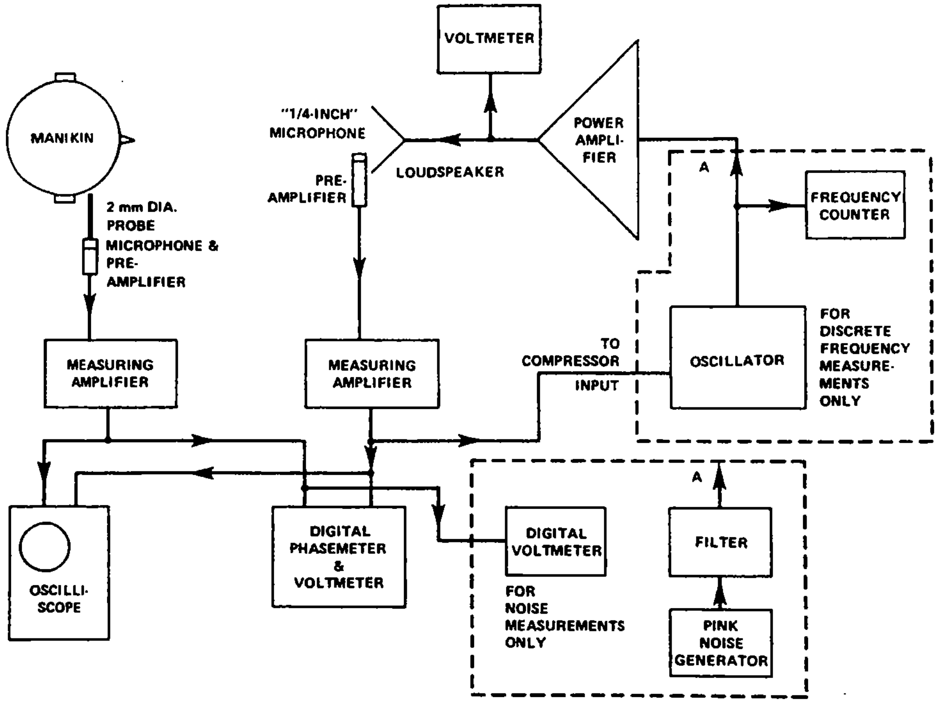


FIG. 3. Block diagram of the instrumentation.

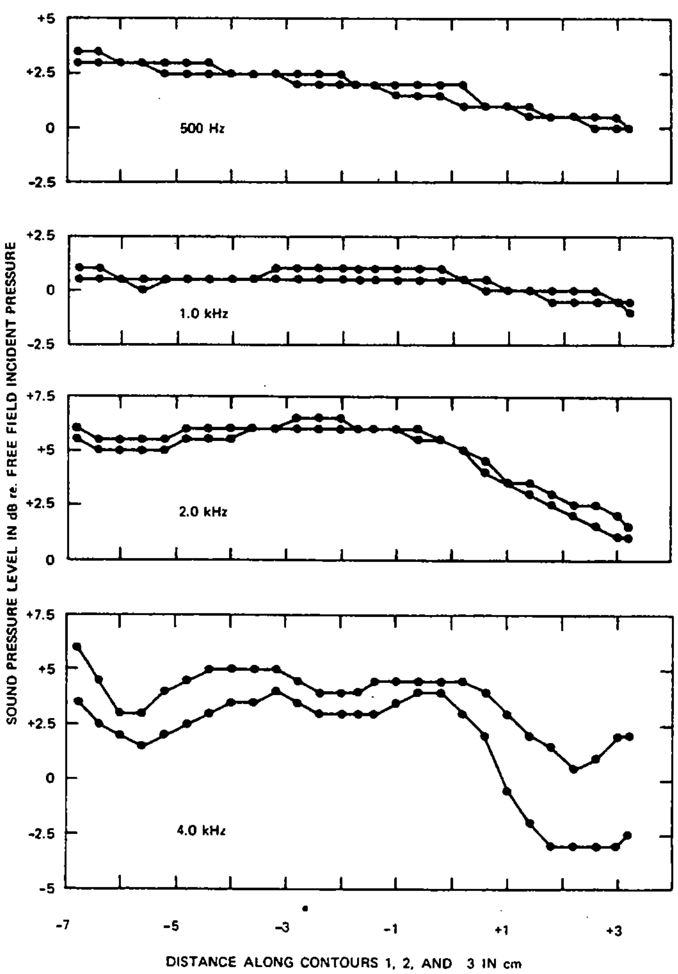


FIG. 4. Range of sound pressure levels for contours 1, 2, and 3 at 2, 4, and 7 mm from the head surface for discrete frequencies. Source-to-ear canal distance is 1.0 m.

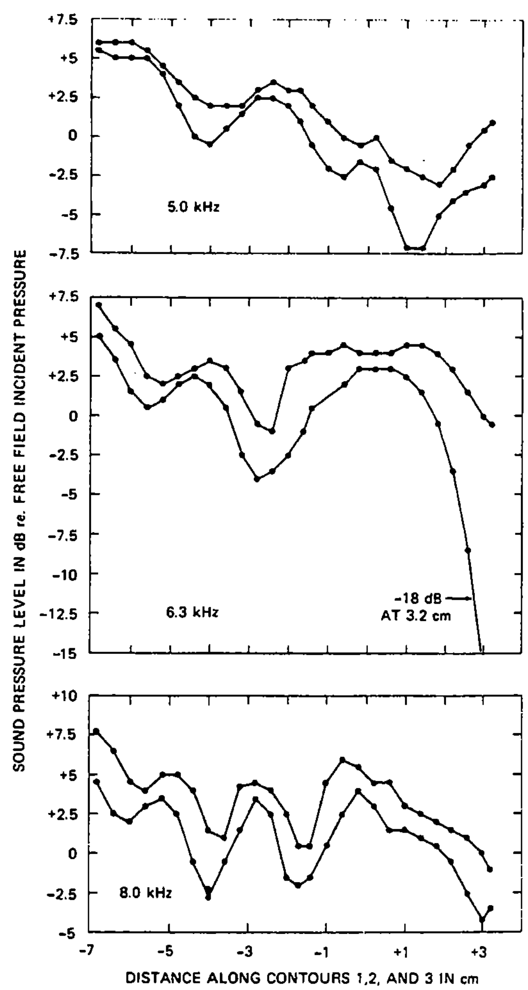


FIG. 5. Range of sound pressure levels for contours 1, 2, and 3 at 2, 4, and 7 mm from the head surface for discrete frequencies. Source-to-ear canal distance is 1.0 m.

defined by the locus of the solid circular dots in Fig. 2. Hearing aid microphones usually lie within about 7 mm of the head surface. Therefore, the positions for the acoustic pressure measurements were chosen to lie 2, 4, and 7 mm, along the y coordinate, away from the head surface. (Note that the y axis coincides with the ear canal axis. The vertical dashes, on contours 1–6 in Fig. 2, indicate the positions at which the acoustic pressures were measured.) Contours 1, 2, and 3 are spaced 5 mm apart in the z direction. Contour 4 was chosen such that the microphone probe just touched the outside perimeter of the pinna. The probe was generally moved in 4-mm increments along the x coordinate for contours 1, 2, and 3 over a total distance of 10.0 cm. Since contours 4, 5, and 6 followed the exact shape of the head and the microphone probe was set in a cartesian coordinate system, it was too complicated to move along any one contour at exactly 4-mm intervals. Therefore, the increments were mostly chosen to be 4 mm along either the x coordinate or the z coordinate depending on whether the contour lay primarily along the x axis or along the z axis, respectively.

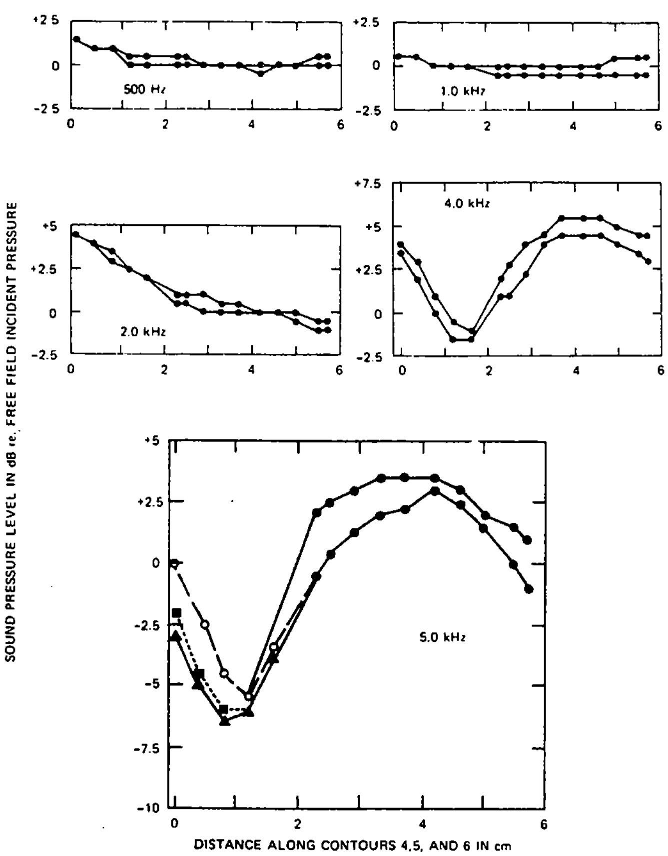


FIG. 6. Range of sound pressure levels for contours 4, 5, and 6 at 2, 4, and 7 mm from the head surface for discrete frequencies. Source-to-ear canal distance is 1.0 m. (At 5.0 kHz along contour 4, the distances from the head surface are: open circles, 2 mm; closed squares, 4 mm; and closed triangles, 7 mm.)

The test frequencies were the octave-band center frequencies from 0.5 to 4.0 kHz and the third octave-band center frequencies from 4.0 to 8.0 kHz. In order to investigate the spatial "smoothing effect" that noise would have on the pressure maxima and minima alongside the head, the measurements were repeated at these test frequencies using 6% (of the center frequency) bandwidth pink noise. Additional measurements were made with 29% (of the center frequency) bandwidth pink noise at 6.3 and 8.0 kHz.

The measurements were made with the equipment shown in Fig. 3 with the exception that the "$\frac{1}{4}$-in." feedback microphone was not used for the measurements using noise.

II. RESULTS

The pressure measurements along contours 1–6 were normalized to the incident free-field pressure in the median plane at the point directly above the ear-canal axis on contour 2. These normalized pressures were then converted to sound-pressure levels and plotted versus the distance along the x coordinate for contours 1–3, or versus the distance

$$[(\Delta x)^2 + (\Delta z)^2 + (\Delta y)^2]^{1/2} \simeq [(\Delta x)^2 + (\Delta z)^2]^{1/2}$$

for contours 4–6; $(\Delta y)^2$ is negligible since $(\Delta y)^2 \ll (\Delta x)^2 + (\Delta z)^2$. Δx, Δy, and Δz are the displacement components between successive measurement points along the x, y, and z coordinates, respectively.

Since the sound pressure measurements were made along six contours at three different distances from the head surface for seven frequencies (with both discrete frequencies and random noise) and for two source-to-ear canal distances, it was necessary to present the data in a condensed form. Therefore only the *range* of the sound pressure levels (at a fixed frequency) for the contour sets 1–3 or 4–6 is presented in Figs. 4–7. Only when there are systematic and large sound pressure level differences from contour to contour, as for example in Fig. 7 at 6.3 kHz, are the specific contours identified.

Figures 4 and 5 show the range of sound pressure levels for contours 1–3 at 2, 4, and 7 mm from the head surface at 7 discrete frequencies. The pressure "builds up" at the front of the head and decreases smoothly to an absolute minimum towards the back of the head between 0.5 and 4 kHz for positions ranging from approximately −6.8 to +0.5 cm relative to the ear-canal axis. The difference between the pressure level at the front to the pressure level at the back increases to 9 dB as the frequency increases from 0.5 to 4 kHz. The pressure field becomes more irregular at higher frequencies with maximum "front-to-back" differences of 13, 25, and 12 dB at 5.0, 6.3, and 8.0 kHz, respectively. For all frequencies forward of the +0.5-cm position, the sound pressure level changes by 4.5 dB or less if the microphone position moves from any one location to another in the y-z plane at any fixed position "x_0". However, as the pinna is approached, severe pressure changes occur as shown, for example, in Fig. 5 at 6.3 kHz. The relative pressure minima between 4.0 and 8.0 kHz are not predicted by the diffraction theory for a rigid sphere.

Figures 6 and 7 show the range of the sound pressure levels for contours 4, 5, and 6 at 2-, 4-, and 7-mm distances from the head surface. The data points between the 0.0- and the 1.6-cm positions are for contour 4 only, since contours 5 and 6 do not extend as far forward as contour 4. The head and/or pinna form a shadow, indicated by the sharp drop in acoustic pressure, when the frequency reaches or exceeds 4.0 kHz. The acoustic pressure levels at any given frequency and at any distance along the contour vary by no more than 3 dB from contour to contour, except at 6.3 and 8.0 kHz where in the vicinity of the acoustic shadow (approximately at the −3-cm position) the variations are much greater than 3 dB. A large systematic difference in sound pressure levels from contour to contour *is* indicated in Fig. 7 at 6.3 kHz around the acoustic shadow. Note that the sharpest pressure minimum does not necessarily lie nearest to the head surface.

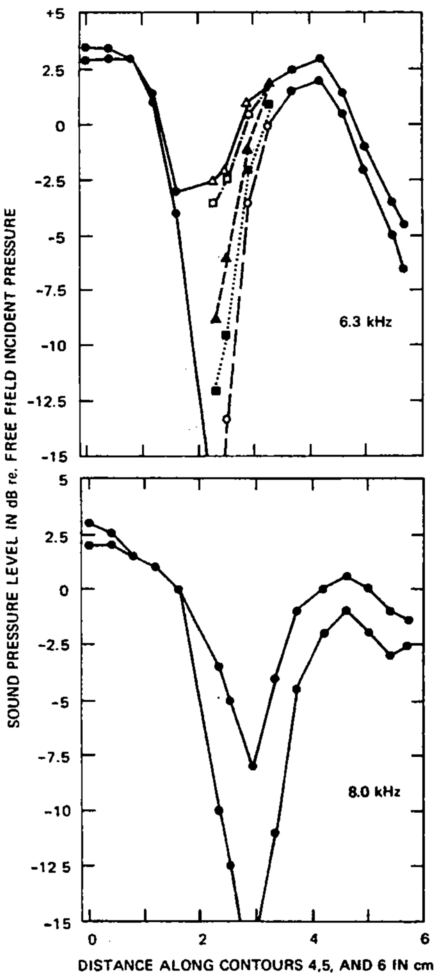


FIG. 7. Range of sound pressure levels for contours 4, 5, and 6 at 2, 4, and 7 mm from the head surface for discrete frequencies. Source-to-ear canal distance is 1.0 m. (At 6.3 kHz: open circles, 7 mm from head surface along contour 4; closed squares, 4 mm from head surface along contour 4; closed triangles, 2 and 4 mm from head surface along contours 4 and 5, respectively; open squares, 7 mm from head surface along contour 6; open triangles, 2 mm from head surface along contour 6.)

The spatial variation of sound pressure around the head can be reduced if random noise rather than a discrete frequency is used for a test signal. Pressure measurements, using filtered pink noise, were made along contour 2 at a distance of 4 mm from the head surface; the results are shown in Figs. 8 and 9. At frequencies equal to or less than 4.0 kHz, the discrete frequency and the filtered pink noise-sound pressure levels differ by less than 1.5 dB. Figure 9 shows the further reduction of the spatial pressure variation when the bandwidth of the noise is increased. For example, at 6.3 kHz and at the −2.4-cm position the pressure minimum is raised by 2.5 and 5 dB relative to the discrete frequency pressure level, respectively, when 6 and 29% bandwidths of pink noise are used. The pressure minima at 8.0 kHz are similarly smoothed.

Figures 10 and 11 show the sound pressure level distribution along contour 4 for discrete frequencies and for filtered pink noise. The pressure minima become more pronounced as the signal frequency is increased. The effect of the (pink noise) bandwidth on the pressure distribution is small at and below 4.0 kHz. Above 4.0 kHz the pressure minimum for the 29% bandwidth noise lies as much as 12 dB above that of the discrete frequency minimum.

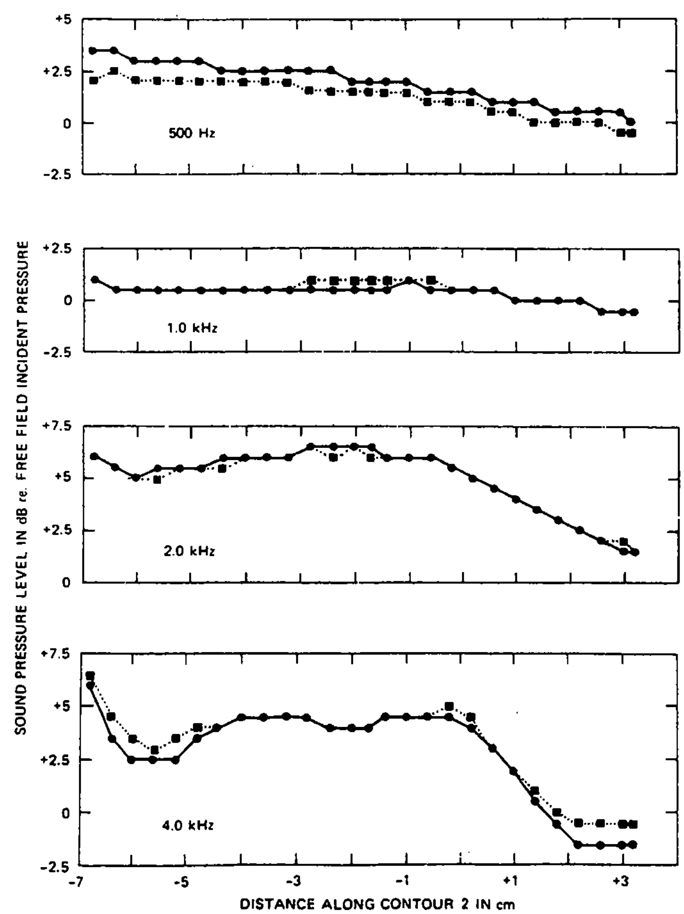


FIG. 8. The effect of random noise on the sound pressure level along contour 2 at 4 mm from the head surface; closed circles, tone; closed squares, 6% bandwidth pink noise. Source-to-ear canal distance is 1.0 m.

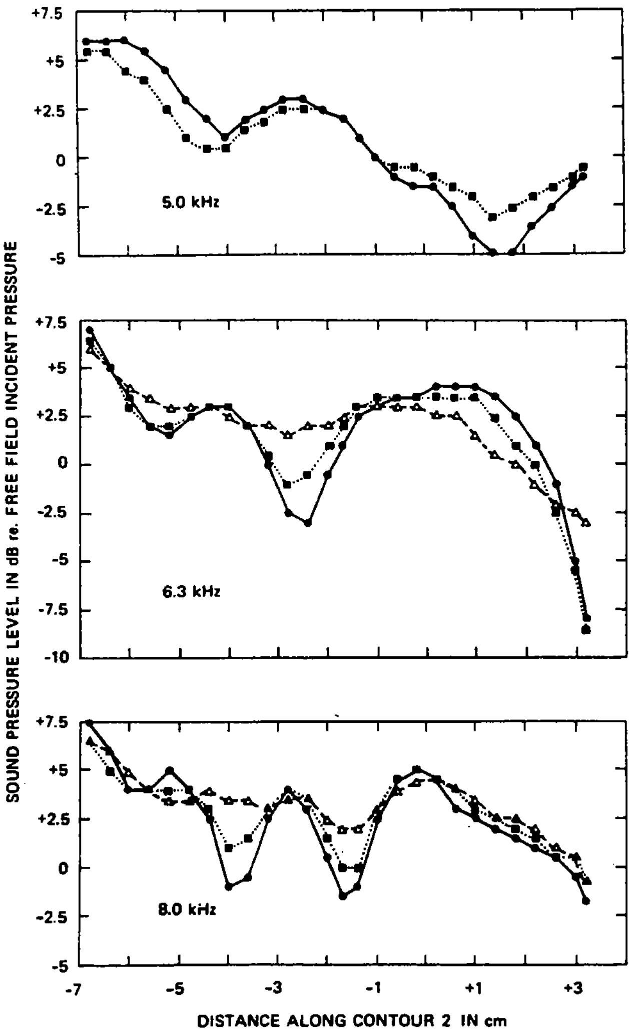


FIG. 9. The effect of random noise on the sound pressure level along contour 2 at 4 mm from the head surface; closed circles, tone; closed squares, 6% bandwidth pink noise; open triangles, 29% bandwidth pink noise. Source-to-ear canal distance is 1.0 m.

Figures 12 and 13 show the effect of the source-to-ear canal distance on the sound pressure distribution. (Although only two distances, 1.0 and 3.5 m, were used here, the results for the 3.5 m distance apply for any source position sufficiently far away to generate an incident plane wave at the body and head.) These results show that the pressure distributions have similar forms regardless of which of the two source-to-ear canal distances is involved. (However, it should not be concluded that the pressure distribution will remain similar for sources closer than 1.0 m, even if they are small compared to a wavelength; nor should this be assumed for distributed sources or sources much larger than a wavelength at any distance from the ear-canal axis.) The pressure levels are within 2 dB for frequencies equal to or less than 4.0 kHz and at 8.0 kHz whether the source is at 1.0 or 3.5 m for the ear-canal axis. (The relatively constant difference at 2.0 kHz could not be traced to a calibration error.) At 5.0 and at 6.3 kHz the shapes of the pressure distribution curves have similar forms but differ primarily around the interference minima. The sensitivity of the pressure to the source location can be expected to be greatest around these minima since a small change in the amplitude(s) and phase(s) of the scattered pressure(s) can have a large effect on the total pressure at the minimum. However, it seems the source distance has little effect on the pressure at 8.0 kHz. For a complex shape such as a torso and a head, generalizations about the scattering from a particular body part and its effect on the pressure around the head are not possible. The total pressure at any one point is built up from the partial contributions of pressures scattered by the entire torso and by the head [Madaffari's (1974) results also illustrate this point], making it difficult to explain why there

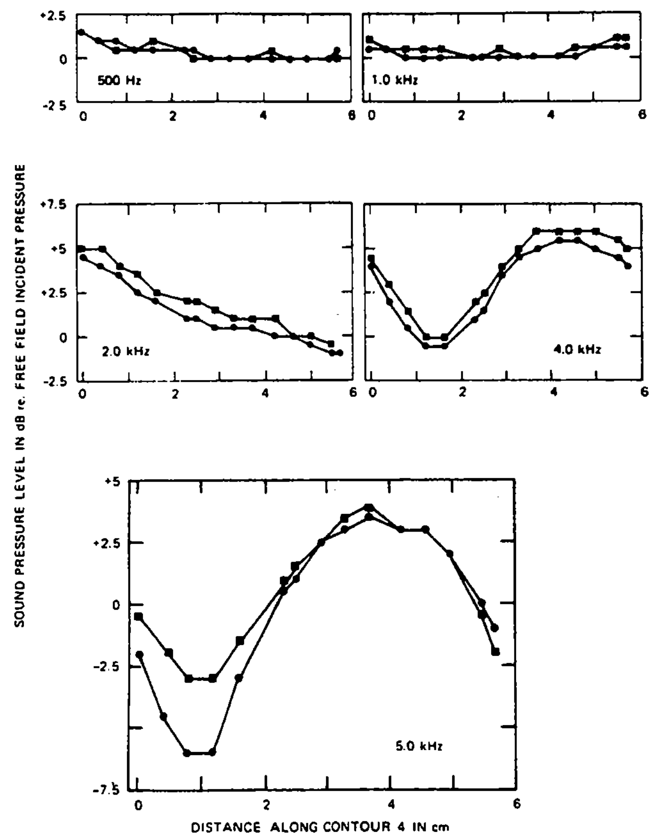


FIG. 10. The effect of random noise on the sound pressure level along contour 4 at 4 mm from the head surface; closed circles, tone; closed squares, 6% bandwidth pink noise. Source-to-ear canal distance is 1.0 m.

is less effect on the pressures at 8.0 kHz than at 5.0 or 6.3 kHz when the source-to-ear canal distance is changed.

Figure 14 shows the sound pressure levels using 6.3 and 8.0 kHz tones and 6 and 29% bandwidth pink noise for plane-wave incidence. Comparing Fig. 14 to Fig. 9 shows that the "smoothing" of the pressure minima using a plane wave is approximately the same as the "smoothing" that occurs for an incident spherical wave, originating approximately 1.0 m in front of the manikin.

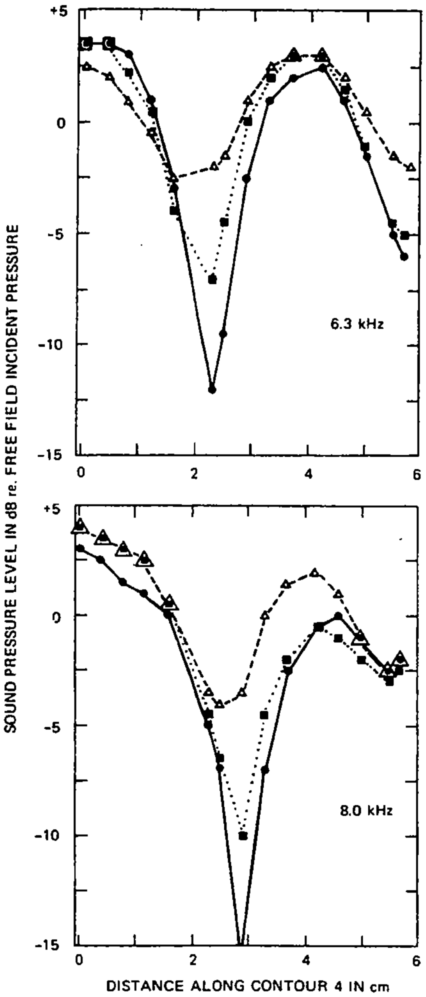


FIG. 11. The effect of random noise on the sound pressure level along contour 4 at 4 mm from the head surface; closed circles, tone; closed squares, 6% bandwidth pink noise; open triangles, 29% bandwidth pink noise. Source-to-ear canal distance is 1.0 m.

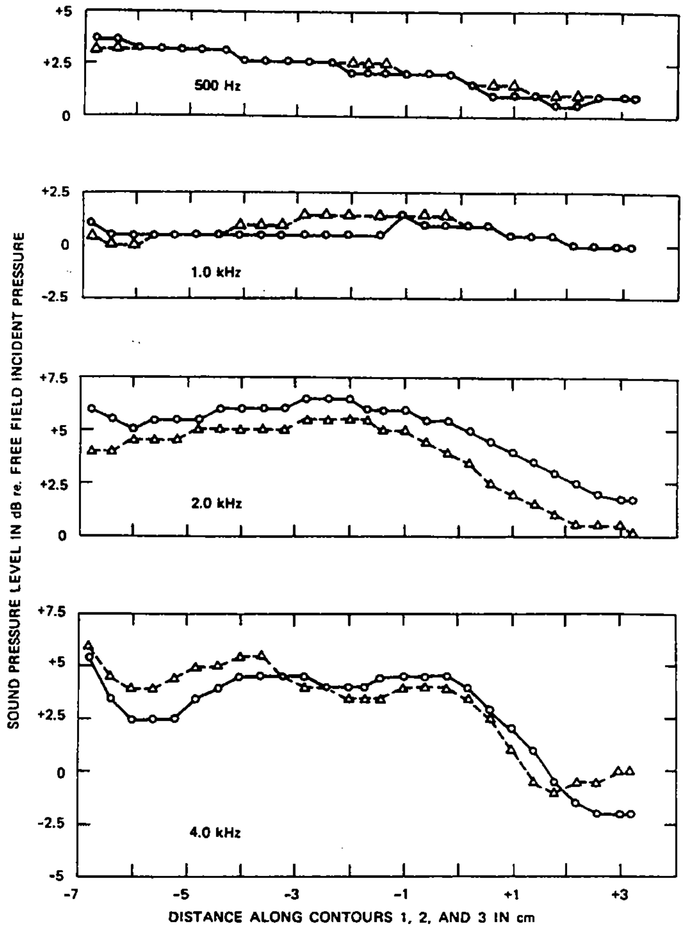


FIG. 12. Comparison of sound pressure level along contour 2 at 4 mm from the head surface for a source placed 1.0 and 3.5 m from the ear canal in front of the manikin; open circles, data for source-to-ear canal distance of 1.0 m; open triangles, data for source-to-ear canal distance of 3.5 m.

III. SUMMARY

The sound pressure levels along contours 1–3 at distances of 2–7 mm from the head are as much as 6.5 dB larger than the incident free-field sound pressure level for frequencies equal to or less than 4.0 kHz and positions more than 1 cm forward of the ear-canal axis. To the rear of the ear-canal axis, the sound pressure level rolls off with distance towards the back of the head. The sound pressure levels along contours 1–3 at frequencies equal to or greater than 5.0 kHz are a strong function of frequency and location, oscillating spatially along any particular contour. An increase in pressure is realized relative to the free-field incident pressure in the first two centimeters of the very forward positions. The above conclusions are only weakly dependent on the source-to-ear canal distance (1.0 and 3.5 m) or on the type of signal (discrete frequency f_0 or pink noise with a 6% or 29% f_0 bandwidth) except in the case of the 29%-bandwidth pink noise excitation where the pressure minima at 6.3 and 8.0 kHz are "smoothed" by as much as 5.0 dB.

The sound pressure levels along contours 4, 5, and 6, around the pinna, vary smoothly with position for frequencies equal to or less than 5.0 kHz. Shadowing effects due to the pinna are negligible for frequencies equal to or less than 2 kHz. However, at 6.3 and 8.0 kHz sharp pressure minima occur behind the pinna which are raised by more than 5.0 and 10 dB if pink noise with 6% and 29% bandwidth, respectively, is used as a test signal.

Figures 4–7 show that the effect of hearing-aid microphone placement accuracy and its effect on the repeatability of the measured sound pressure level becomes more critical towards the back of the head, especially around and behind the pinna. Also, as the frequency increases above 2.0 kHz the reliability of the pressure measurements depends strongly on the microphone positioning accuracy.

The phase of the pressure, although not discussed in this paper, has been measured and is smooth and continuous except at positions centered around the pressure minima on contours 4–6 at 6.3 and 8.0 kHz. The phase changes abruptly by approximately 180° at these frequencies as the pressure minima are traversed. Hearing aids with more than one microphone or microphone port may give unreliable test results at such locations.

Test signals using random noise of some bandwidth may be useful for testing hearing aids, particularly above 5.0 kHz since the pressure maxima and particularly the minima are smoothed out. However, this "smoothing" of the spatial pressure distribution is at the expense of some frequency resolution due to the bandwidth of the noise.

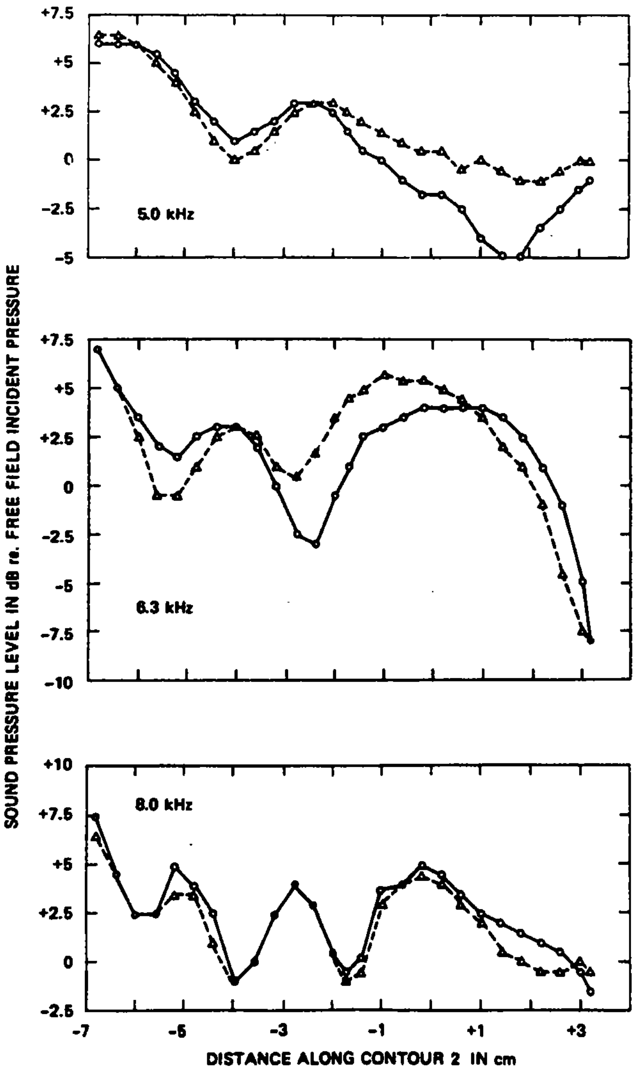


FIG. 13. Comparison of sound pressure level along contour 2 at 4 mm from the head surface for a source placed 1.0 and 3.5 m from the ear canal in front of the manikin; open circles, data for source-to-ear canal distance of 1.0 m; open triangles, data for source-to-ear canal distance of 3.5 m.

[a]Some parts of this paper were initially presented at the 91st Meeting of the Acoustical Society of America, Washington, D.C., 4–9 April 1976 [J. Acoust. Soc. Am. 60, S30(A) (1976)].

[1]The 1.0-m distance is typical of conversational speech and is being considered for hearing-aid testing (Burkhard, 1976). The 3.5-m distance is used to simulate an incident plane-wave condition.

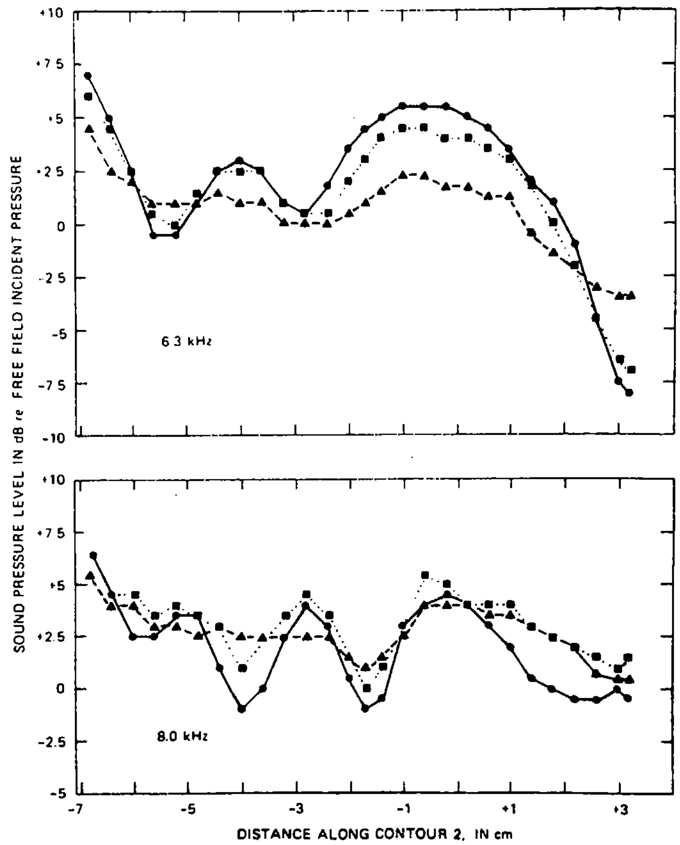


FIG. 14. The effect of random noise on the sound pressure level along contour 2 at 4 mm from the head surface for a source placed 3.5 m from the ear canal in front of the manikin: closed circles, tone; closed squares, 6% bandwidth pink noise; closed triangles, 29% bandwidth pink noise.

ANSI (1976). S3.22. "Specification of Hearing Aid Characteristics," American National Standards Institute, New York.
Burkhard, M. D., and Sachs, R. M. (1975). "Anthropometric manikin for acoustic research," J. Acoust. Soc. Am. 58, 214–222.
Burkhard, M. D. (1976). Personal communication.
IEC (1959). "Recommended methods for measurements of the electro-acoustical characteristics of hearing aids," Publication 118, International Electrotechnical Commission, Geneva, Switzerland.
Kuhn, G. F. (1976). "An objective sound localization model in the azimuthal plane," Seminar at the Natl. Bur. Stand., Washington, D.C., April 29, 1976 (to be published).
Lybarger, S. F., and Barron, F. E. (1965). "Head baffle effect for different hearing aid microphone locations," J. Acoust. Soc. Am. 38, 922 (A).
Madaffari, P. L. (1974). "Pressure variation about the ear," J. Acoust. Soc. Am. 56, S3 (A).
Olson, W., and Carhart, R. (1975). "Head diffraction effects on ear-level hearing aids," Audiology 14, 244–258.
Rzhevkin, S. N. (1963). *The Theory of Sound* (Pergamon, Oxford), pp. 362–363.
Schwarz, L. (1943). "Zur Theorie der Beugung einer ebenen Schallwelle an der Kugel," Akust. Z. 8, 91–117.
Temby, A. C. (1965). "Sound diffraction in the vicinity of the human ear," Acustica 15, 219–222.
Veterans Administration (1976). "Handbook of Hearing Aid Measurements 1976," U.S. GPO, Washington, DC 20402, IB 11–52.
Wiener, F. M. (1947a). "On the diffraction of a progressive sound wave by the human head," J. Acoust. Soc. Am. 19, 143–146.
Wiener, F. M. (1947b). "Sound diffraction by rigid spheres and circular cylinders," J. Acoust. Soc. Am. 19, 444–451.

Chapter 8

A Comparison of Insertion Gain and Substitution Measurement Methods on KEMAR

by

L.B. Beck

Biocommunications Laboratory
University of Maryland
College Park, Maryland

and

G. Donald Causey

Veterans Administration
Washington, D.C.

Presented in Washington, D.C., April 5, 1976

In the audiologist's task of restoring communicative ability to the hearing-impaired individual, the selection of the appropriate amplifying device is a prime consideration. Hearing aids are described in terms of their electroacoustic characteristics, and the audiologist applies instrumental criteria in his ultimate determination of a "good" hearing aid. One of the problems in this chain of events has been the lack of agreement between electroacoustic and behavioral measures. Among the many contributing factors which must be considered are the effects of the person, including head diffraction and/or body baffle effects and ear canal resonance, interacting with the hearing aid's physical characteristics to influence the hearing aid's performance on the person.

The *K*nowles *E*lectronics *M*anikin for *A*coustic *R*esearch, known as KEMAR, is an anthropometric manikin which provides us with the means to evaluate the many factors that contribute to hearing aid performance on the person. Burkhard and Sachs (1975) demonstrated that KEMAR (which is used with the Zwislocki coupler) simulated the acoustic response of a human in the free field.

Presently, there are two techniques for measurement of hearing aids on KEMAR. One technique is the insertion gain method, also called the orthotelephonic response, whereby gain is defined as the difference between the sound pressure level at the eardrum with the hearing aid in place and the sound pressure level at the eardrum with no hearing aid in place. The technique represents a distinct departure from the conventional definition of gain. The orthotelephonic response, as explained by Beranek (1949), examined the transmission system (the hearing aid) without the contribution of the basic reference system (the ear).

Another technique for measurement of aids on KEMAR is the substitution method whereby gain is defined as the difference in sound pressure level at a test point and sound pressure level at the coupler when KEMAR, with the hearing aid in place, has been located at the test point. The term "substitution" is used herein to describe that measurement technique whereby a constant sound pressure level (SPL) is established at the test point with a reference microphone. The reference microphone is then removed and KEMAR, with a hearing aid in place, is located at the test point and exposed to the sound field signal previously established.

The purpose of this paper is to describe the instrumentation and procedures necessary to measure hearing aids using the insertion gain and substitution methods and to provide a comparative analysis of the resulting frequency responses.

Equipment

A block diagram of the instrumentation used for measurement of hearing aids on KEMAR is shown in Fig. 8-1. Measurements were made with a Bruel & Kjaer hearing aid test system and an anechoic chamber having an internal dimension of 343 cubic feet. A beat frequency oscillator generated the signal which was led to a JBL LE8T speaker, located in the anechoic chamber, and placed directly in front of KEMAR in a 0° azimuth relationship. The

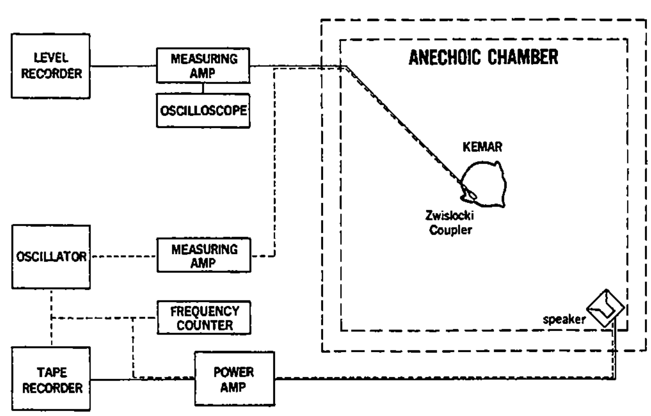


Fig. 8-1. Block diagram of the instrumentation used for measurement of hearing aids on KEMAR.

center of the loudspeaker cone was one meter from the midpoint of a line between KEMAR's two ears. A Scully ½" tape recorder was used in the system to record and playback the test signals. The Zwislocki coupler assembly (coupler and plate) was attached to a ½" condenser microphone and mounted on KEMAR. The microphone output was then led to a measuring amplifier for determination of sound pressure level at the test point (KEMAR's ear). The output of the measuring amplifier was delivered to a graphic level recorder to record frequency response. A compression or regulating circuit was connected to the test microphone to maintain constant sound pressure level when necessary.

This equipment array could be manipulated for measurement of hearing aids using the insertion gain and substitution measurement methods.

Prior to initiation of the procedures for measurement of hearing aids on KEMAR, the effects of the manikin on the constant free sound field signal at 0° azimuth were examined. A constant sound pressure level sweep frequency signal was presented to the manikin and the effects of the manikin on this signal were recorded. Fig. 8-2 shows these frequency-dependent effects for two different test sessions eight days apart.

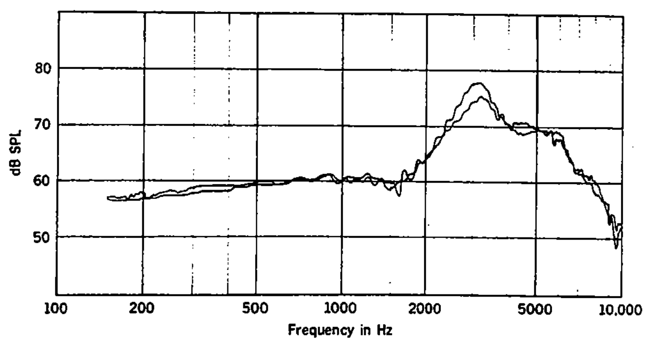


Fig. 8-2. The effects of KEMAR on a free sound field measured at the eardrum with source location at 0° azimuth for two test sessions eight days apart.

Procedures

To measure the insertion gain of hearing aids, KEMAR was located in the chamber at a distance of one meter from the speaker. The ½" condenser microphone attached to the Zwislocki coupler in KEMAR's ear (the eardrum microphone) was made the regulating microphone by activating the compression circuit. The oscillator output, controlled by the compression circuit to produce 60 dB SPL throughout the frequency range, was recorded on magnetic tape. The eardrum (coupler) microphone was converted to the measuring microphone (by disconnecting the compression circuit) and the tape-recorded signal was played back through the speaker to KEMAR's eardrum microphone. The graphic level recorder, connected to the output of the measuring amplifier, provided a frequency response tracing of the signal occurring at the eardrum microphone during the recording and playback portion of test signal preparation. The insertion gain measurement technique compensated for the effects of the loudspeaker and the effects of KEMAR (including head diffraction and ear canal resonance) on the resulting response.

An example of the steps involved in generating the insertion gain test tape signal, the voltage required to produce constant SPL at the eardrum microphone across frequency, is shown in Fig. 8-3. The response labeled A is KEMAR's response

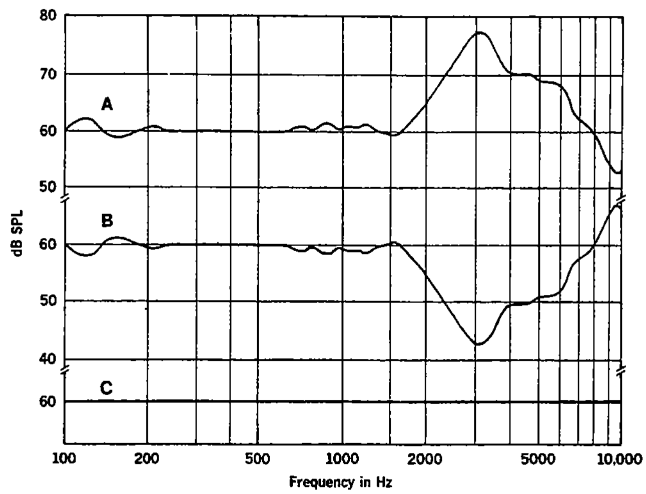


Fig. 8-3. Generation of the insertion gain test tape signal. Response A is KEMAR's response measured at the eardrum microphone to a constant input SPL signal at 0° azimuth. Response B is the sweep frequency test signal stored on magnetic tape. Response C is the sweep frequency signal measured at the eardrum microphone when response B, the test signal, is presented to KEMAR, response A.

measured at the eardrum microphone to a constant input SPL. The test signal stored on magnetic tape is represented by the response labeled B. The sweep frequency test signal stored on tape and the frequency response of KEMAR to a constant SPL input were reciprocal. When the signal stored on tape, response B, was presented to KEMAR, the resulting response was constant across frequency as shown by the response curve labeled C. The signal on tape, which becomes the test signal for insertion gain hearing-aid measurement, was depressed in the frequency range where the effects of KEMAR were present. Consequently, the effects of the manikin were subtracted from the insertion gain frequency response.

Another technique which can be used to measure the insertion gain of hearing aids on KEMAR is a procedure which requires that Zwislocki coupler assemblies be mounted on each of KEMAR's ears, connected to ½" condenser microphones, and subsequently, to measuring amplifiers. A compression circuit is then connected to one side and that microphone becomes the regulating or compressor microphone, while the other side is the test

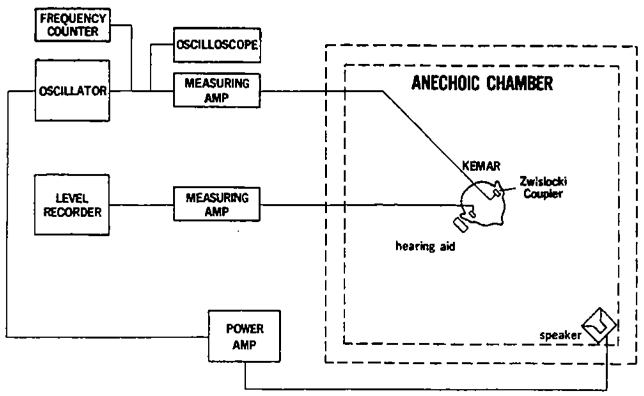


Fig. 8-4. Block diagram of instrumentation used for the two Zwislocki coupler measurements procedure on KEMAR.

microphone, to which the hearing aid is connected. The resulting instrumentation arrangement, diagrammed in Fig. 8-4, was analogous to the one employed with the 2cc coupler whereby a constant sound pressure level was maintained at the test point by a regulating microphone. The compression microphone on one side then, compensated for the effects of the manikin and the loudspeaker. If KEMAR were symmetrical, and all components equal in performance, these effects should equal the ones present on the test side and therefore also be compensated. Preliminary attempts at using this procedure were not successful in our laboratory; however, at least one other laboratory has reported eventual success with the method, and we intend to give it further consideration.

To measure hearing-aid gain using the substitution method, a ½" condenser microphone was connected to a compression circuit. With KEMAR out of the chamber, the compression microphone was placed at the test point, a position which was the same height as KEMAR's ear canal at a distance of one meter from the loudspeaker. A constant 60 dB

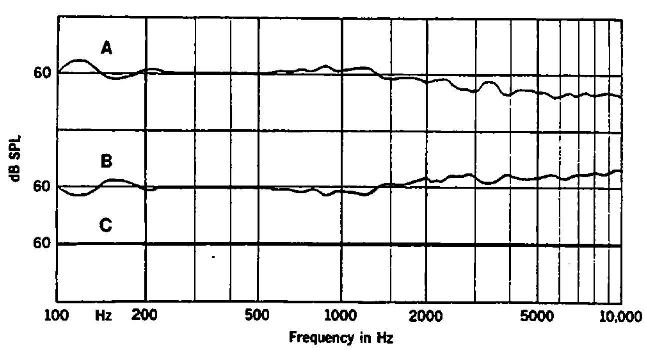


Fig. 8-5. Generation of the substitution taped signal. Response A represents the loudspeaker's response. Response B represents the sweep frequency test signal stored on magnetic tape. Response C is the sweep frequency signal measured at the test point when response B, the test signal, is presented through the loudspeaker, response A.

SPL signal was maintained across the frequency range from 100 to 10,000 Hz, and the necessary levels to achieve this signal were recorded on tape. The compression circuit was disconnected and the signal on the tape was played back through the loudspeaker to the ½" condenser microphone. The resulting response, as evidenced by a flat frequency tracing on the level recorder, was a constant input SPL at the test point. The signal on tape represented the levels require to achieve constant SPL across frequency at the test point.

Fig. 8-5 presents an example of the steps involved in generating the taped signal for the substitution method. The frequency response labeled A represents the signal presented to the microphone through the loudspeaker resulting from constant voltage input; the frequency response labeled B represents the test signal stored on magnetic tape. When frequency response B, the signal on tape, was presented through the loudspeaker to the microphone located at the test point, the resulting response was constant across frequency, as indicated by the tracing labeled C. The use of this technique compensated for the presence of any loudspeaker effects. KEMAR (with previously-described coupler assembly) was then placed in the anechoic chamber at the test point. This tape-recorded signal was used as the test stimulus for measurement of hearing aids using the substitution method.

Prior to recording the sweep frequency signal for both measurement methods, a 1000 Hz signal was also recorded for use as a reference signal. Additionally, it was necessary to generate a new tape recording of the sweep frequency signal periodically since it tended to depart from flatness over time.

In measuring the performance of hearing aids the 0° relationship between KEMAR and the sound source (loudspeaker) was changed slightly during the routine procedure of repeatedly placing earmolds and hearing aids on KEMAR. Since it did not seem possible to avoid these disturbances, differential effects of slight azimuth changes on the test signal were examined. The insertion gain test signal at 0° azimuth was generated at KEMAR's eardrum microphone. KEMAR's turntable was rotated 2.5° and 5° left and right from 0° and a response was obtained using the 0° test signal. The effects of the these angle rotations on the test signal were shown as deviations from the flat response and can be seen in Fig. 8-6. The response marked 0° was the response measured at 0° azimuth after the manikin had been rotated to the previously-stated azimuth angles and then returned to 0°. Differences of less than ±2 dB were seen across the frequency range of interest. At 2.5° there was less than 1 dB change. When the manikin was returned to 0° azimuth position following azimuth angle rotation, the resulting response was still within acceptable tolerance limits. Slight angle variations, such as those resulting from routine test

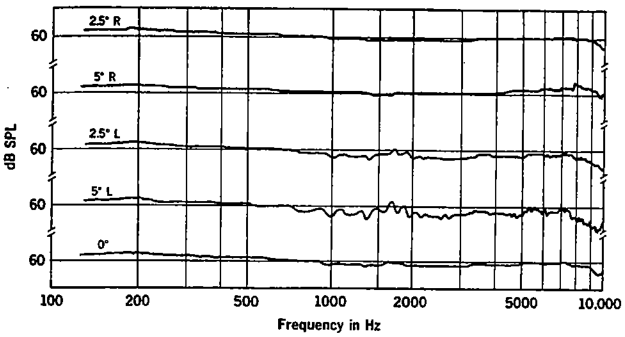


Fig. 8-6. Effects of slight azimuth angle changes on the insertion gain signal measured at KEMAR's eardrum microphone. After the initial insertion gain signal was generated, the manikin was rotated 2.5° and 5° left and right from 0° azimuth and then returned to the original 0° azimuth position.

procedures, would not be expected to affect test results significantly. However, such angle shifts might contribute to variability in repeated measures.

KEMAR was outfitted with a tee shirt and wig for testing since it is recommended that clothing, such as tee shirt, lab coat, and wig be used with KEMAR. Two post-auricular hearing aids at full-on volume control settings were evaluated on KEMAR with both measurement techniques. An occluding earmold with cemented tubing of 2 mm inside diameter and 4 cm in length, was used for testing the two aids on KEMAR. The insertion depth of the earmold canal was such that it was 2 mm short of being flush with the Zwislocki coupler joint.

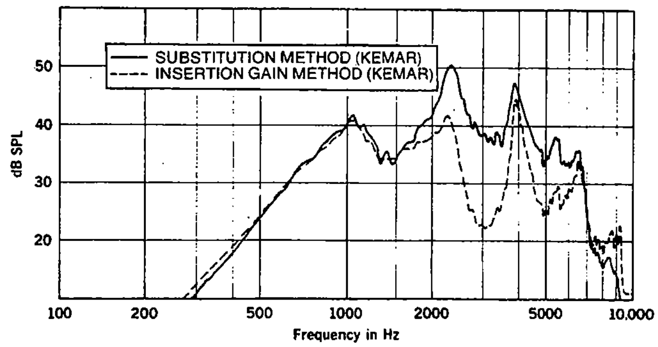
Results

The performance of the two hearing aids on KEMAR as a function of measurement method are shown in Fig. 8-7 (a) and (b). The solid line is the substitution method response and the dotted line is the insertion gain method response. Performance differences between the two methods for the two hearing aids are seen in the frequency region above 1500 Hz.

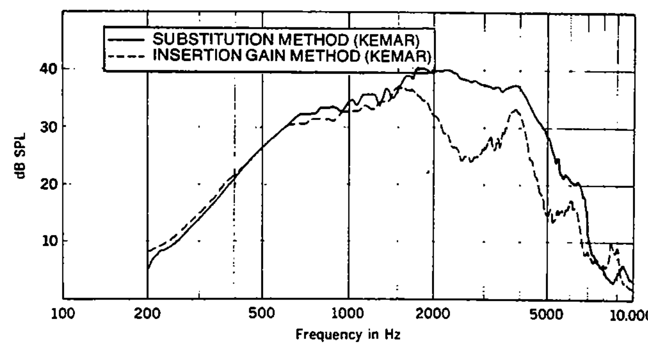
Table 8-I displays the mean amount of difference in dB at discrete frequency points between the insertion gain and substitution methods for each hearing aid along with the range of difference for the two hearing aids. These data indicated that the difference between methods for both hearing aids was observed principally in the frequency region above 1500 Hz with the substitution method measuring more gain. The greatest difference was observed in the region around 3 kHz. The differences may not be uniform for all aids because of microphone placement, relative to location on the manikin, or mechanical problems in signal presentation and recording.

Table 8-I. Amount of difference in dB and the range of difference between the substitution and insertion gain measurement methods for two hearing aids.

	Frequency in Hertz						
	500	700	1000	1500	2000	3000	4000
HA1	0	1	2	2	7	15	5
HA2	1	.5	0	-2	5	10.5	3
Range	1	.5	2	4	2	4.5	2



(a)



(b)

Fig. 8-7. Frequency responses of two hearing aids with the insertion gain and substitution measurement methods. The solid line is the substitution response and the dotted line is the insertion gain response.

Discussion

The difference between the insertion gain and substitution methods for measurement of hearing aids on KEMAR can be explained by examining the procedures inherent in preparation of the test signals. For the insertion gain method you will recall that the signal stored on tape and presented to the hearing aid under test was a sweep frequency signal which provided a constant SPL at the eardrum microphone. In order to be flat at the eardrum microphone, the compressor circuit adjusted the gain across the frequency range as necessary to compensate for KEMAR's head diffraction, ear canal resonance, and any effects of loudspeaker response.

For the substitution method, the signal stored on tape represents the voltage across frequency which provides a constant SPL at the test point. In order to be flat at the test point (with KEMAR absent), the compressor microphone circuit adds and subtracts gain as necessary to compensate the loudspeaker effects.

Fig. 8-8 shows the actual frequency response recorded on tape for each method. The difference between the taped signals for the two methods was observed in the range above 1500 Hz where the insertion gain signal provided the additional de-emphasis necessary to achieve a flat signal at the eardrum microphone. The de-emphasis in this frequency range is seen as the gain difference above 1500 Hz between the two methods for each hearing aid. The insertion gain method subtracted the effects of the manikin from the hearing-aid frequency response, while the substitution method incorporated these effects into the hearing-aid frequency response, thereby showing more gain for the same hearing aid in the frequency region above 1500 Hz.

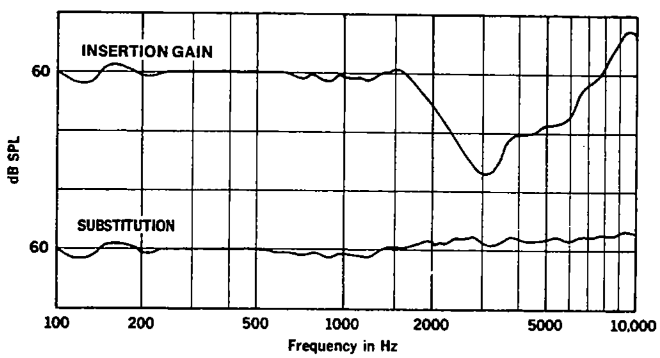


Fig. 8-8. Sweep frequency test signals for the insertion gain and substitution measurement methods.

The task remains to devise audiologic procedures that relate hearing sensitivity to hearing-aid performance on KEMAR. Cole (1975) suggested that the difference between aided and unaided pure-tone thresholds was the behavioral counterpart to the insertion gain of the hearing aid.

The development of KEMAR permits the evaluation of hearing aids in a fashion that can relate directly to wearer performance. We have many issues to contemplate, including repeatability of measures at different laboratories, adoption of standard methods, and consideration of techniques that relate measures on KEMAR to behavioral measures.

BIBLIOGRAPHY

Beranek, L.L. *Acoustic Measurement;* New York, John Wiley and Sons, 1949.

Burkhard, M.D. and Sachs, R.M. (1975). Anthropometric manikin for acoustic research. *J. Acous. Soc. Am.* 58, 214-22.

Cole, W. "Hearing Aid Gain—A Functional Approach" *Hearing Instruments, October, 1975.*

Shaw, E.A.G. (1974). "Tranformation of Sound Pressure for the Free Field to the Eardrum in the Horizontal Plane." *J. Acous. Soc. Am.* 56, 1848-61.

Chapter 9.

Some Considerations In Using KEMAR To Measure Hearing Aid Performance

D.A. Preves

Starkey Laboratories, Inc.

Presented in Washington, D.C., April 5, 1976

Introduction

There are many pitfalls that may trap acousticians who use a manikin to measure performance of hearing aids. Unless proper procedures are used, incorrect results, which are difficult to detect as being invalid, may be obtained and incorrectly applied. This paper will present some of the pitfalls that may occur and methods of avoiding them.

Results obtained with KEMAR must be carefully evaluated to determine what they actually represent. One of the purposes of using a manikin is to include the effect of head and torso diffraction on incident sound in hearing aid performance. Therefore, in order to obtain only the gain of the hearing aid itself, diffraction of the manikin must be differentiated in some manner in the data obtained. The advantages and disadvantages of different methods for obtaining and expressing hearing aid performance will be discussed.

Requirements for the Anechoic Chamber

In order to effectively use KEMAR, an anechoic chamber with certain minimal characteristics must be used. One of the first questions that arises is what minimum size chamber is required to allow diffraction effects caused by KEMAR to occur without modification. Designers of anechoic chambers have commonly placed the test point at least ¼ wave length away from the nearest wedge tip to obtain accurate measurements at the lowest frequency of interest. If performance is to be measured down to 150 Hz, the minimum distance from the wedge tips to any surface on KEMAR must be at least 22 inches. Taking KEMAR dimensions into consideration, the minimum inside dimensions for the anechoic chamber would be about a six foot cube. Of course, this volume is considered to be minimum, and with a larger size, more accurate results may be obtained.

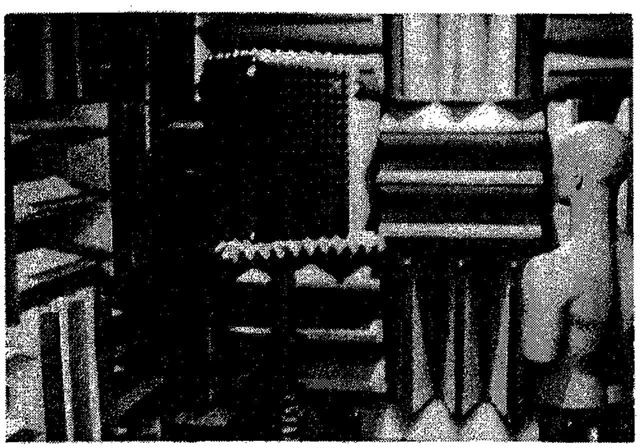


Fig. 9-1. Picture of KEMAR in anechoic chamber used for this study.

Solid surfaces such as speaker enclosures and metal objects inside the chamber should be covered with a highly absorbent material to prevent reflections. Fig. 9-1 shows a picture of the KEMAR manikin in the anechoic chamber used for this study. The inside dimensions for this chamber are 8'x8'x6'9".

Inverse square law measurements may be used to determine the suitability of the chamber for accurately obtaining directivity data. Sound pressure should theoretically decrease by 6 dB as the distance from the speaker along its axis is doubled.

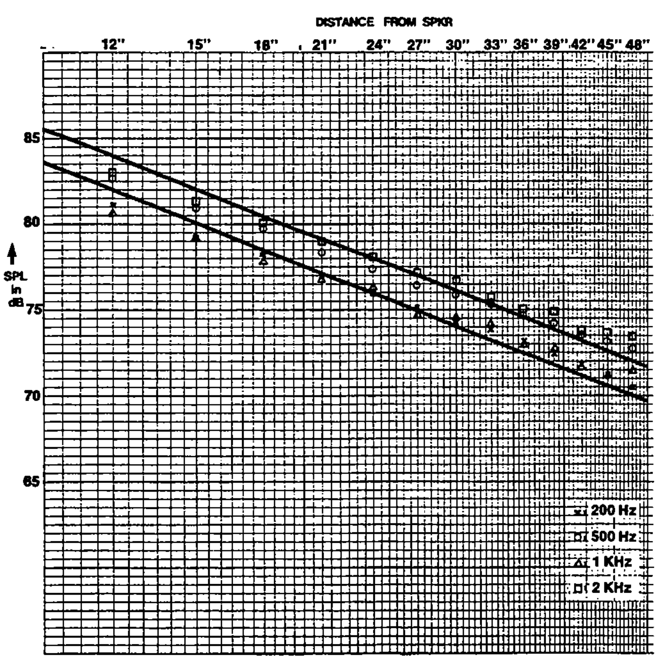


Fig. 9-2. Inverse square law measurements with 3 way speaker on 3 separate axes.

Figure 9-2 shows the results of a pitfall which occurred in the initial stages of our measurement set-up. The speaker used for these inverse square law measurements was a JBL Century L-100, 3-way speaker system, which is a high quality system having three separate speakers on different axes in the enclosure. The use of such a speaker system violates the theory that sound should emanate from a point source in the anechoic chamber. More than one speaker on separate axes will result in the test point receiving 3 separate wave fronts from the speakers with possibly different phase angles and directivity patterns. Figure 9-2 thus shows relatively large deviations of 1 to 2 dB from ideal inverse square law variations of 6 dB per doubling of distance which is represented by solid lines.

Figure 9-3 shows the improved behavior of the sound field resulting from changing the 3-way speaker system to a single speaker. Here, deviation from ideal inverse square law variation is only about ½ dB maximum at distances greater than 2 feet from the speaker.

The inverse square law measurements were taken on the speaker axis to obtain the data shown in Figs. 9-2 and 9-3. In order to determine how sound pressure varies in the volume occupied by KEMAR, which is centered on the speaker axis, measurements of sound pressure deviation at different angles off the speaker axis on several radial arcs were taken. Figure 9-4 shows the results of these measurements.

The maximum deviation from axis sound pressure is 2 dB at 1000 Hz at a 20° angle off the axis for a 48 inch radial arc. Table 9-I has a complete listing of the data plotted in Fig. 9-4.

Measurements on KEMAR with Speaker Response Equalized (Substitution method)

If a control microphone/compressor amplifier system is used in real time to control the sound pressure incident to hearing aids on KEMAR, diffraction from KEMAR will effect the control microphone output. Therefore, some of the diffraction pattern will be disturbed by the action of the control microphone/compressor amplifier system. In order to prevent this, some means other than a real

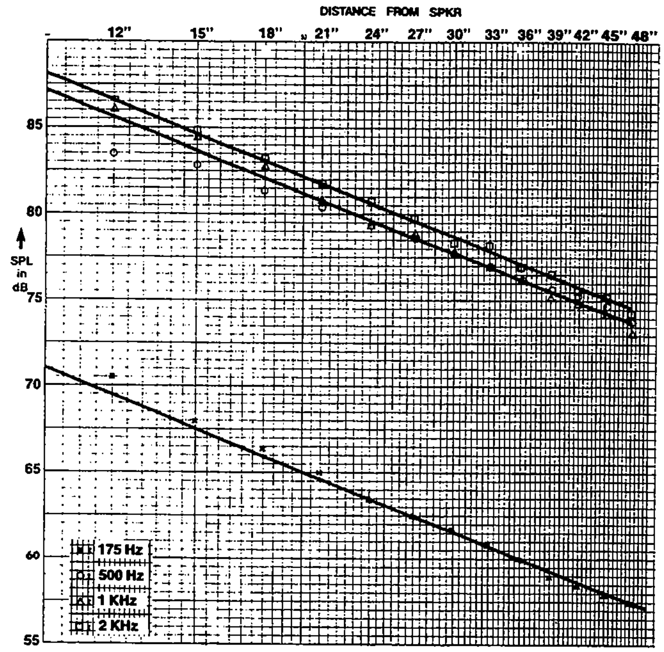


Fig. 9-3. Inverse square law measurements with single speaker.

time compressor must be used to equalize the speaker response to provide a flat sound pressure level over the frequency range.

Two possible approaches are:

1. Drive the speaker through a staggered-tuned filter set whose response compensates for variations in speaker sensitivity over the frequency band
2. Tape record the signal required from the amplifier to provide a flat response at the test point using the compressor amplifier system without KEMAR present in the room. When this tape is played back directly to the speaker with KEMAR in the room, all of the diffraction effects caused by KEMAR will occur.

Figure 9-5 shows the system used for tape recording equalization. One disadvantage of this system is that it is difficult to manually synchronize the beginning of the sweep tone on the tape recording to the XY recorder. This problem may be

Table 9-I. Sound pressure measurements on radial arcs at discrete angles from speaker axis at 500 Hz, 1 KHz, 2 KHz (in decibels).

	Hz	0°	+5°	+10°	+15°	+20°	-5°	-10°	-15°	-20°
24"	500	81.2	81.2	81.2	81.1	80.6	81.2	81.0	81.0	80.8
	1000	80.0	80.0	79.5	79.4	78.6	80.0	79.5	79.3	79.1
	2000	78.6	78.6	78.4	78.4	77.6	78.6	78.2	78.1	77.7
36"	500	78.1	78.2	78.0	77.9	77.6	78.1	78.0	77.9	77.6
	1000	76.2	76.2	75.5	75.3	75.0	76.3	75.9	75.3	75.3
	2000	75.5	76.2	76.2	75.8	75.5	75.8	75.4	75.0	75.0
48"	500	75.5	75.7	75.8	75.5	74.9	75.7	75.6	75.7	75.3
	1000	74.2	73.9	73.5	72.8	72.2	74.1	73.8	73.5	73.5
	2000	73.3	73.0	73.5	73.0	72.7	73.1	72.7	73.4	72.5

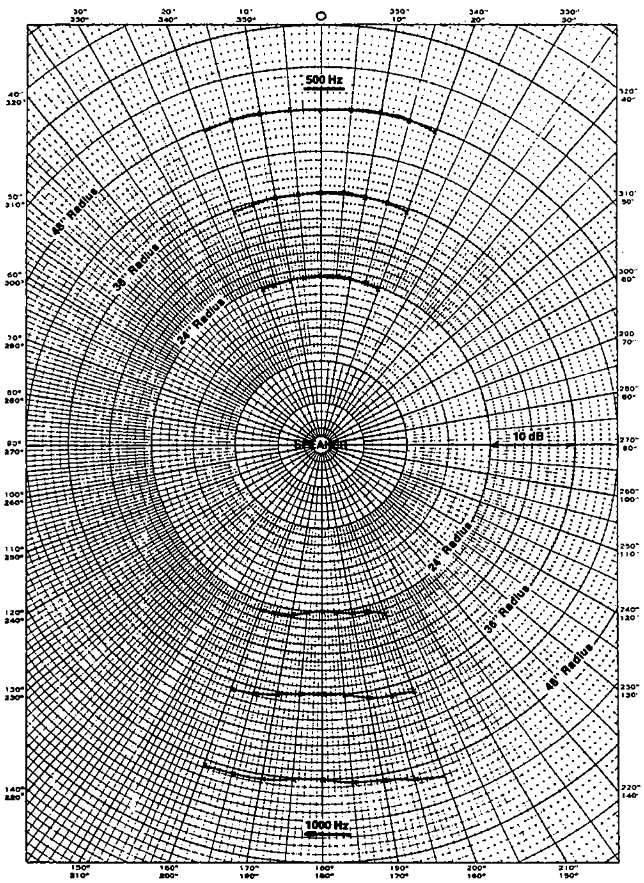


Fig. 9-4. Deviation from constant SPL, off speaker axis, on 3 radial lines in anechoic chamber at 500 Hz, 1 kHz.

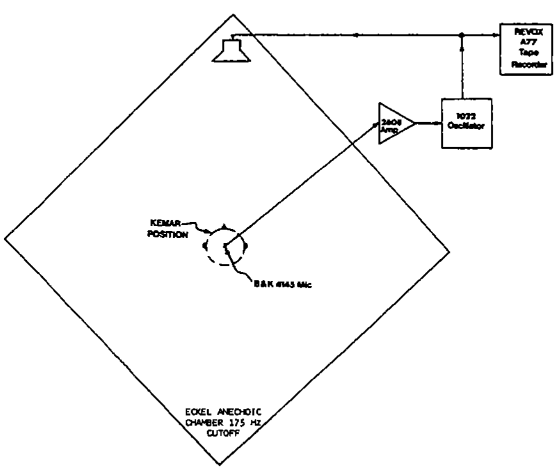


Fig. 9-5. Instrumentation to provide flat SPL at center position of KEMAR head without KEMAR present.

solved by triggering the XY recorder with an electronic sensor on the tape at the beginning of the sweep.

Figure 9-6 shows the amplifier output recorded to provide a flat sound field over the frequency range with KEMAR not present. This is essentially the inverse of the speaker response at the test point.

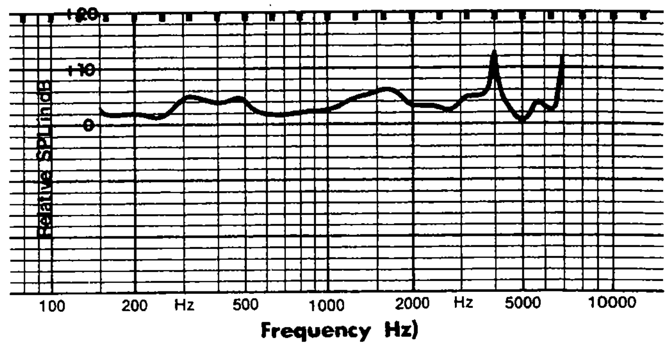


Fig. 9-6. Input to speaker to provide a flat SPL at position of center of KEMAR head without KEMAR present.

Measurements made with this drive signal may be called the *response of the hearing aid on KEMAR* rather than the *gain of the hearing aid* since diffraction of KEMAR is included in the data.

Measurements on KEMAR with Both Speaker Response and KEMAR Diffraction Equalized (Etymotic Response)

In order to determine the absolute gain of the hearing aid itself, both speaker response variations and diffraction from KEMAR over the frequency range must be taken into account. By tape recording the signal from the compressor amplifier required to produce a flat sound pressure over the frequency range at the KEMAR eardrum, variations caused by speaker response and diffraction are equalized.

Figure 9-7 shows the system configuration for this method and Fig. 9-8 shows the tape recorded amplifier output to the speaker required to provide a flat sound pressure at the KEMAR eardrum. The signal is essentially the inverse of the speaker response plus KEMAR diffraction and ear canal resonance.

Hopefully, the results obtained of gain versus frequency of a hearing aid on KEMAR using this drive signal should correlate to unaided minus aided thresholds of a hearing impaired subject wearing the hearing aid.

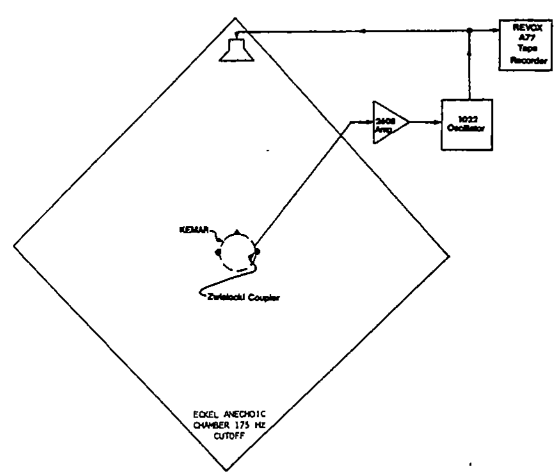


Fig. 9-7. Instrumentation to provide flat SPL at KEMAR eardrum.

Comparison of Etymotic and Substitution Methods

Both methods have several disadvantages for use in obtaining other measurements on hearing aids: the amount of harmonic distortion produced by a hearing aid may be related to the input sound pressure level incident to the hearing aid microphone. The same is true for saturation sound pressure level measurements. Thus, harmonic distortion and saturation sound pressure level measurements should be performed with a constant input to the hearing aid to obtain data for the aid

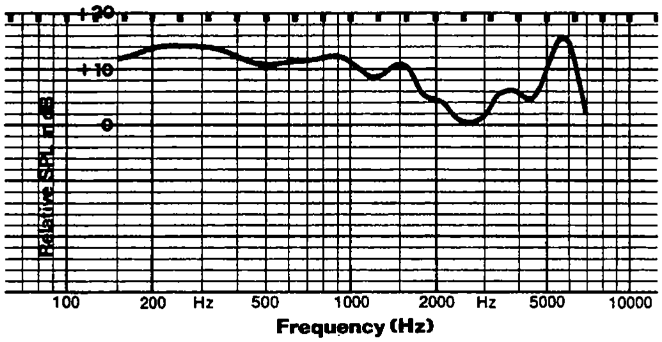


Fig. 9-8. Input to speaker to provide a flat response at KEMAR eardrum – 0° incidence.

itself. Obviously, the signal at the hearing aid microphone will not be constant over the frequency range for either the etymotic or substitution methods. Therefore, neither method may be suitable for SSPL or distortion measurements. However, in the actual use of the aid, head diffraction does take place. Therefore, the substitution method seems to be the logical choice for showing how the aid will perform *as worn,* even though the data obtained will not be for the aid itself.

Figure 9-9 shows SSPL 90 curves obtained for a hearing aid with the substitution method and the etymotic method. Note the curve labeled KEMAR correction is considerably below the curve labeled speaker correction in the high frequencies. The speaker correction curve is presumably the true SSPL 90 curve for the aid *as worn,* but not for the aid itself.

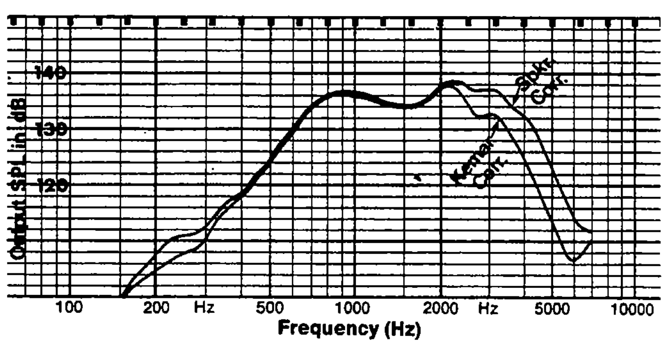


Fig. 9-9. SSPL 90 with substitution method (speaker correction) and etymotic method (KEMAR correction) inputs.

Another problem that arises with the etymotic response is that many master hearing aids utilize supra or circumaural cushions which do not permit all of the head diffraction to occur. Results obtained from these master hearing aids are modified substantially by head diffraction when the hearing aid is placed on the head. Consequently, it would be desirable to include the gain of the head in the manufacturer's specification sheet for the hearing aid so that dispensers may see how the aid will actually perform as worn.

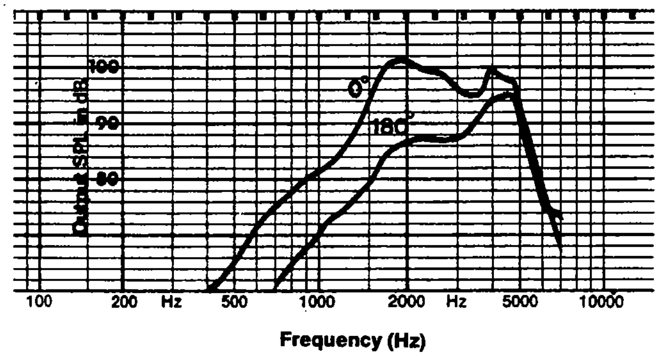


Fig. 9-10 (a). Directional ITE aid response in KEMAR using control microphone positioned 2" above center of KEMAR head.

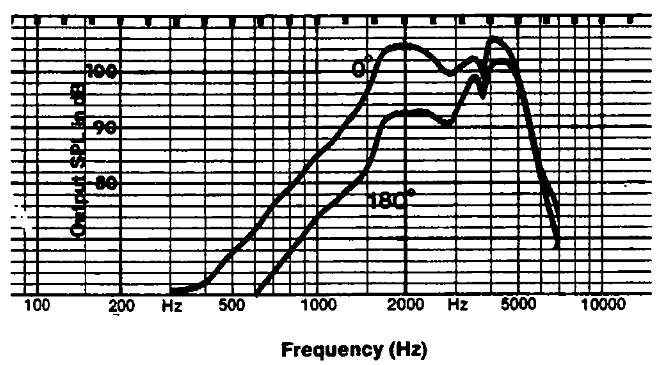


Fig. 9-10 (b). Directional ITE aid response on KEMAR using control microphone positioned on axis ½ way between KEMAR and speaker – input SPL 6 dB higher.

Comparison of Methods for Obtaining Data for Hearing Aids with Directional Microphone

The effect of KEMAR diffraction is particularly significant for measurements of hearing aids using a directional microphone. As mentioned previously, the use of a microphone and compressor amplifier to control the sound pressure at the test point in real time partially negates diffraction effects.

Response curves of an in-the-ear aid using a directional microphone were obtained using a control microphone in different locations feeding a compressor amplifier as well as with substitution and etymotic methods. Results are shown in Fig. 9-10 and 9-11. Figures 9-10 (a) and (b) show the 0°

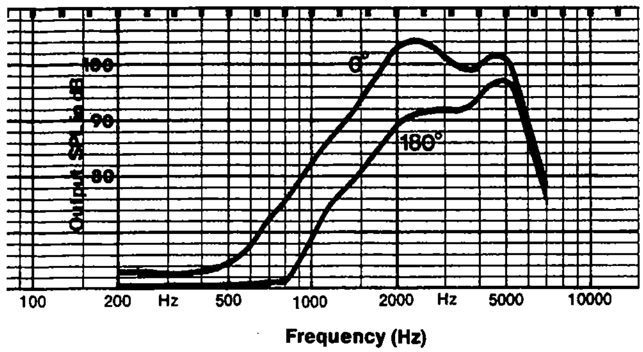


Fig. 9-11 (a). Directional ITE aid response on KEMAR using tape recorder to provide flat SPL at position of center of KEMAR head without KEMAR present.

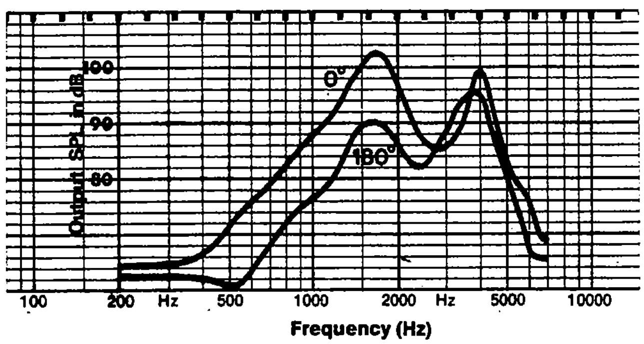


Fig. 9-11 (b). Directional ITE aid response on KEMAR using tape recorder to provide flat SPL at KEMAR eardrum with no aid present. Separate correction curves for 0°, 180° incidence.

and 180° incident responses with real time compression and the control microphone positioned 2 inches above the center of KEMAR head and one-half way between KEMAR and the speaker, respectively. For the measurements in Fig. 9-10 (b), the control microphone/compressor amplifier held the input SPL constant at 66 dB because, in an anechoic chamber, the sound pressure should decrease by 6 dB as distance is doubled. Thus, the sound pressure at the test point should be 60 dB, ignoring the diffraction from KEMAR.

The curves in Fig. 9-11 (a) and (b) were generated using the tape recorder to drive the speaker to produce substitution method and etymotic responses respectively. Note that the curves in Fig. 9-10 (a) closely approximate the curves in Fig. 9-11 (a), indicating that the much simpler method of utilizing a real time compressor with control microphone directly above KEMAR's head may produce results that are very similar to the substitution method curves.

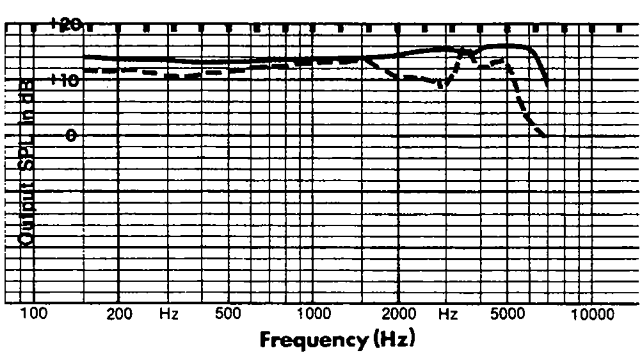


Fig. 9-12. Response at KEMAR eardrum with KEMAR at 180° incidence using tape recordings for flat SPL at eardrum at 0° incidence (-----) and at 180° incidence (———).

The 180° incident curve in Fig. 9-11 (b) was made using a separate tape recording from that used for the 0° incident curve in the figure. This procedure is required because the diffraction caused by KEMAR is changed as the angle of incidence is varied. If separate tapes are not used, a sound field similar to the dotted curve in Fig. 9-12 will be obtained. With KEMAR at 180° incidence with respect to the speaker, the solid line in Fig. 9-12 represents KEMAR eardrum pressure using the correct tape recording to equalize KEMAR diffraction at 180° and the dotted line shows KEMAR'S eardrum pressure using the tape recording that equalizes for 0° incidence. The difference between the two curves is essentially the inverse of the difference between KEMAR's diffraction patterns at 0° and 180° incidence.

Effect of Clothing Manikin

There has been considerable discussion about whether clothing on KEMAR effects the data obtained. Fig. 9-13 shows that there is really almost no difference in the sound field at KEMAR'S eardrum with or without a wig. However, if a hearing aid case comes in contact with the wig, damping

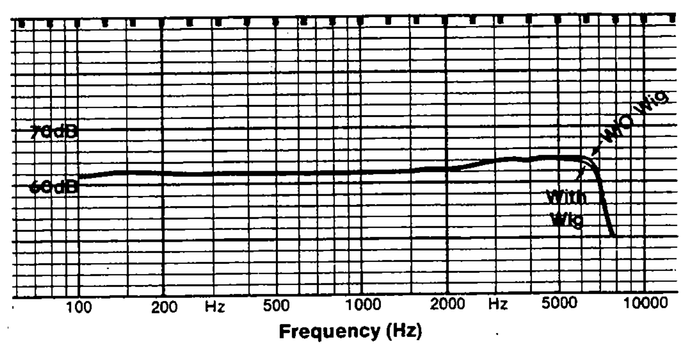


Fig. 9-13. Sound field at KEMAR eardrum with and without wig on KEMAR using tape recording for etymotic response made without wig.

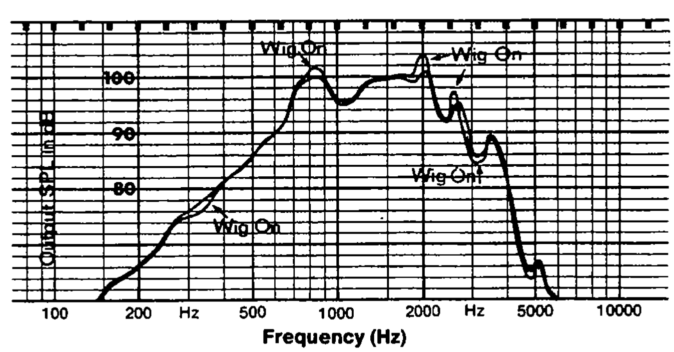


Fig. 9-14. Response at 0° incidence of a directional aid on KEMAR with and without wig.

provided by the soft wig material may change the response curve due to altering resonance in the hearing aid response as shown in Fig. 9-14. Response curves of hearing aids obtained with or without a sweater on KEMAR were found to be the same, so the effect of clothing the torso does not seem significant.

Effect of Earmold Data

There is a great need for standardization of earmolds used on KEMAR. For the present, physical characteristics of earmolds used should be specified with hearing aid performance data published using KEMAR. Fig. 9-15 shows an SSPL 90 measurement on a hearing aid using two different earmolds designed for the KEMAR ear. The curve generated with the long, small o.d. canal shows greater SSPL at mid to high frequencies than with the short, large o.d. canal, using the same canal bore diameter. Thus, the earmold characteristics can have a significant effect on the data obtained.

Repeatability of Measurements

If tape recording equalization is used, it is of great value to re-check the flatness of the sound pressure level over frequency periodically to insure that the flat sound field provided has not degraded. If using the control microphone/compressor amplifier method in real time, it is not as necessary to do this.

Figure 9-16 is an example of what may happen if the X-Y recorder is not perfectly synchronized with the tape recording. Each curve was run twice. One of the 180° incident curves shows a significant shift when exact synchronization was not achieved. Subtle differences such as these may cause wide deviations in results obtained and degrade repeatability.

Summary

This paper has documented some of the problems encountered in making measurements with KEMAR as well as discussing some of the trade-offs in methods of obtaining and expressing the data. Particular care must be taken in providing an anechoic chamber with certain minimal characteristics. The test setup for sound field, clothing for KEMAR, and earmolds used must be carefully selected to achieve the desired measurement conditions.

Although both methods give valid data, the substitution response is generally preferred over the etymotic response for obtaining hearing aid performance data since it represents how the hearing aid will operate as worn for the performance parameters generally measured.

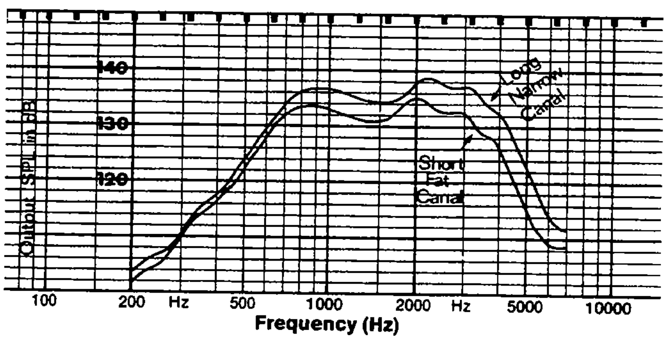


Fig. 9-15. Effect on SSPL90 of earmold canal length and O.D. using same bore diameter.

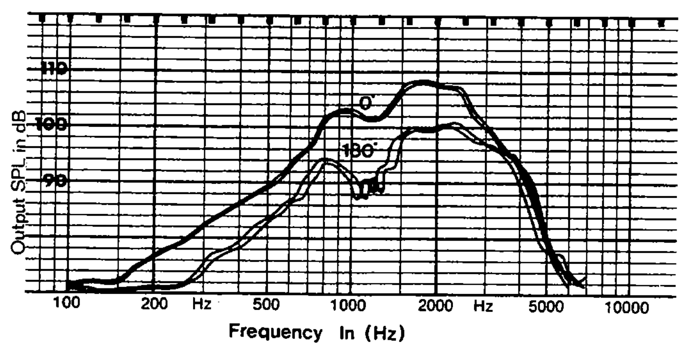


Fig. 9-16. Curve shift from imperfectly synchronizing tape recording with X-Y recorder.

Chapter 10.

Measurement Procedures

M. D. Burkhard
Industrial Research Products, Inc.

Controlling the Sound Field for Manikin Measurements

It has been common practice to use a "servo" controller with a standard microphone to control the sound pressure at a constant value in sound boxes and anechoic rooms. The output of the hearing aid placed in this constant sound pressure field is plotted to give the gain and response as a function of frequency. Unfortunately, such a constant control method cannot be used with a hearing aid on a manikin because the servo control system will work to suppress the sound field perturbation deliberately introduced by the manikin. The closer the control or reference microphone is to the manikin and hearing aid the more the variations of the manikin sound field are removed. A way to circumvent this dilemma, is to store a signal that gives a stimulus sound pressure that is constant at the measurement site before the test is made and to then use this stored signal as the test signal. Depending on the measurement philosophy or objective, the stored signal would likely be one that produces a constant sound pressure at the point in front of a loudspeaker where the center of the manikin head is to be placed or the signal that produces a constant sound pressure at the eardrum of the manikin. A quality tape recorder or a computer memory are convenient signal storage media. The discussion that follows goes into various aspects of the measurements and the variables that should be considered when a magnetic tape stored signal is used.

The tape recorder used in our work is an Ampex Model AG-500-2P. It is a two channel reel-to-reel recorder which we operate at 15" per second to ensure maximum band width and signal to noise ratio. The two channels can be recorded simultaneously or independently and one can be used as a signal source while a recording is made on the other one. These features have proved useful as will become apparent in other discussions in these proceedings.

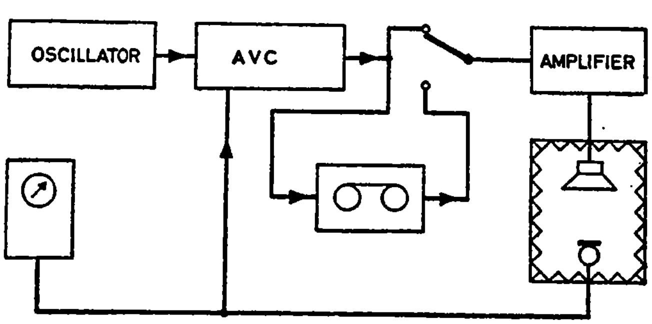


Fig. 10-1. A block diagram of the system for generating a tape recorded control signal for a constant free field.

The anechoic room has fiberglass wedges, 2 feet long. Inside dimensions are 8' x 8' x 8'. The source for most of the measurements was a 8" diameter speaker cone mounted in a .014 m^3 box. The center of the manikin head was placed 1 meter from the center of the speaker on its axis.

Wansdronk (1959) described the basic recorded signal method, Fig. 10-1. This would apply specifically to the case when the manikin is located in a free sound field. First with the switch in the upper position, the voltage that creates a constant pressure at the microphone is recorded as the oscillator is swept through the frequency range. The microphone is then removed, and the device or system in question is placed at that location. The response of the device or system is recorded on a chart recorder that runs in synchronism with the tape when it is played back with the switch in the lower position. A typical set of data in our room is shown in Fig. 10-2.

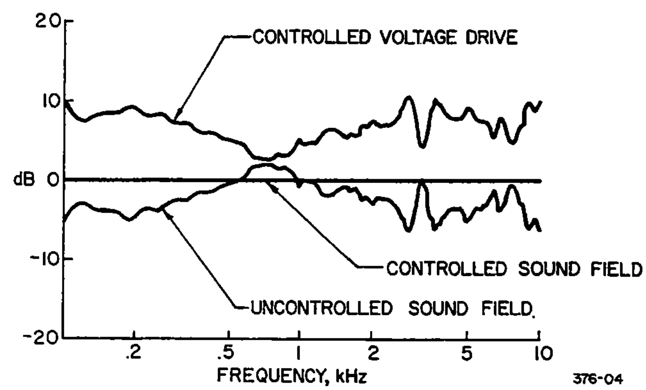


Fig. 10-2. The response curves showing servo controller operation to produce a constant sound field at the test point in a free field.

The lower curve is the sound field at the test location with a constant voltage from the oscillator signal source. The upper curve shows the signal that was recorded when the control circuit was active to produce a constant sound field pressure. The center line at 0 dB is the resulting sound pressure at the test location when the controlled voltage drive signal is played back. Care must be taken in this type of measurement to synchronize the frequency scale of the chart recorder with the tape recorded signal.

The next few figures illustrate the control that typically exists for this type of measurement scheme. In Fig. 10-3, the change of free field sound pressure with repeated replay of a tape segment may be seen. Slight imperfections are produced by lack of flatness of the tape recorder response as shown by the "new tape" plot.

There is of course a limit to the number of times a prerecorded magnetic tape can be played back before the signal quality and dynamic range become unacceptable. After twenty replays of the tape, the quality deteriorates to a marginally acceptable level, as seen in the "worn tape" response curve. This degradation observed with an "analog" recording would be almost totally eliminated with a

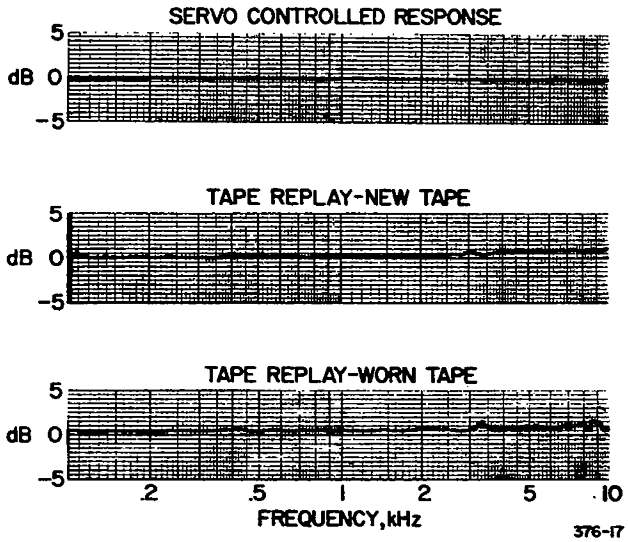


Fig. 10-3. Uniformity of first and twentieth tape play back sound field level compared to the level when the servo control is working directly.

"digital" or an "FM" type of recording. Digital solid state memory storage of the signal would be the most resistant to deterioration with use.

The uniformity of field in the region where the manikin is placed is illustrated in Fig. 10-4. A circle

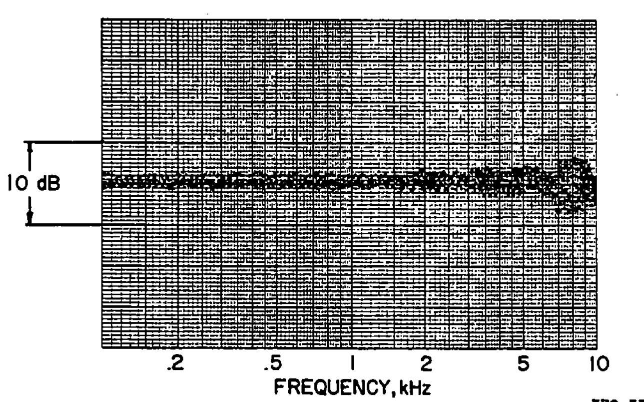


Fig. 10-4. A plot of the uniformity of the sound field on a 15.2 centimeter diameter circle, centered 93 centimeters from the source, for a constant free field pressure level at the center point of the circle.

of 15.2 cm diameter is approximately the size of the KEMAR manikin head. The results are somewhat tied to the speaker used. All of the manikin data reported by Burkhard and Sachs and by Burkhard in this proceedings, have this field uniformity variable, unless otherwise noted. A control or reference microphone was placed at the test location, i.e., where the center of the manikin head was to be placed, and the response curves were plotted for the other locations relative to the "head center". Two of the locations were along the axis of the sound propagation from the loudspeaker, thus giving rise to around 2.5 dB of level variation from maximum to minimum.

In addition to the above variations that enter into a hearing aid measurement on a manikin, the way its position in the sound field is controlled and the details of the manikin itself affect the precision of results. There is a great temptation to forego the complication of a prerecorded signal source and place a reference or control microphone in the field simultaneously with the manikin to control the sound pressure. An indication of the confusion that might be created by such a procedure is shown in Fig. 10-5. The solid line is the free-field open ear response of a KEMAR manikin. A prerecorded sig-

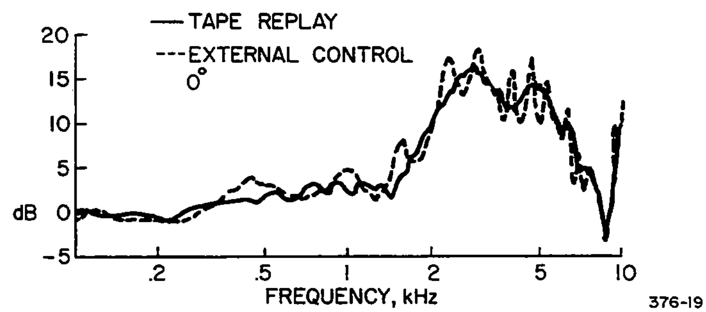


Fig. 10-5. The comparison of the open ear response of the KEMAR manikin for a tape replay free field and for control by an external microphone placed between the loudspeaker and the manikin, two feet from the loudspeaker.

nal that produced a constant sound pressure with frequency at the test location was used. By contrast an eardrum pressure response with oscillations, the dashed curve, is produced if a standard control microphone is placed between the loudspeaker and the manikin, two feet from the speaker. (The manikin is one meter away.) As the control system works to achieve a constant pressure at the control microphone, by eliminating reflections between the manikin and the speaker, unacceptable variations are created in other parts of the field, and at the manikin eardrum in particular. Experimental results have been reported with a manikin between the source and sound field control microphone, but we were not able to produce a smooth free field equivalent with the technique in our small anechoic room.

The manikin position in the sound field and its orientation should be controlled accurately. Fig. 10-6, shows how pointing the manikin active ear 5° closer to or 5° away from the source can introduce up to 2 dB difference in the eardrum sound pressure at some frequencies. These changes in ear sound levels arise primarily from changes in the head shadow and torso sound reflection.

Just as there are asymmetries in persons, the ears of the KEMAR manikin show some acoustic

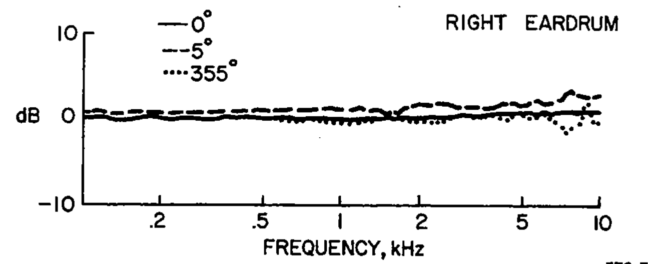


Fig. 10-6. The change in sound pressure at the right eardrum of the KEMAR manikin for small misalignments in the orientation of the manikin in a free sound field. Five degrees would cause the ear to be closer to the loudspeaker. 355° would cause the ear to be away from the loudspeaker in the shadow of the head.

asymmetry. The Burkhard and Sachs paper, Fig. 2-9, showed the differences between the sound pressures at the entrance to the two KEMAR ears. These differences carry through to the two eardrums as shown in Fig. 10-7. The stimulus or reference signal condition is the free sound field, sans manikin.

In general these differences are quite small except around 8 kHz, but they may lead to some frustration when, as some investigators have suggested, one ear of the KEMAR manikin is used to control the sound pressure at an eardrum, while a hearing aid is evaluated on the other ear. In addition to depending on small imperceptible differences in the ears of the KEMAR, these differences between the ears are very sensitive to the symmetry of the sound field, the orientation of the manikin in

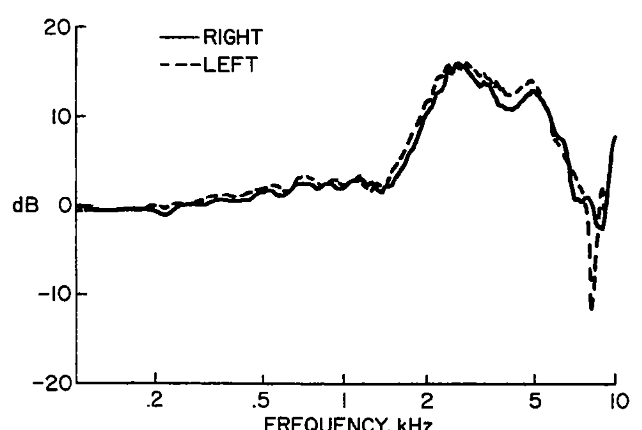


Fig. 10-7. Sound pressure at the eardrum in the KEMAR left and right ears for a free field sound pressure.

the sound field, the direction of the axis of the loudspeaker, and the elevation angle of the source. The reason for considering such an experimental set-up will be discussed later.

The general dependence of response of the manikin ears to elevation angle of the source is indicated by Fig. 10-8, which is reproduced from some work by Shaw. This figure indicates that the dependence on elevation at small angles may in fact be greater for the vertical than the azimuthal angle, especially at frequencies above 4 kHz.

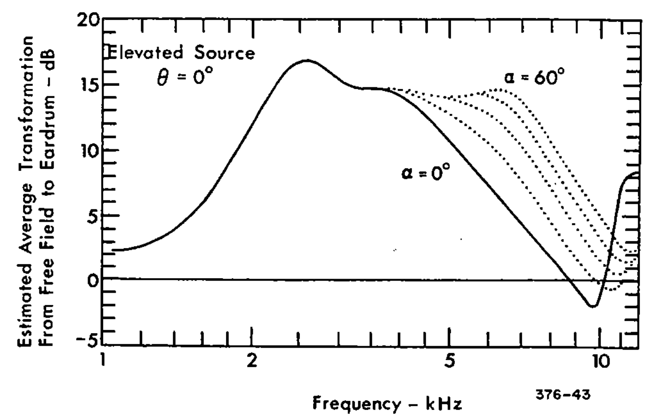


Fig. 10-8. The sound pressure at the eardrum for an average person with the elevation angle of the sound source as a variable. (Shaw 1974).

Source Size and Distance

To achieve consistency of hearing aid measurements on manikins, among laboratories, restrictions on loudspeaker sources may be necessary that were heretofore irrelevant. In particular the size of the loudspeaker and the distance from it cause measurable effects because the test object now includes the whole head and torso. These subtend a relatively large solid angle and the diffraction effects at the ear therefore depend on the uniformity of field over a larger space. Symmetry of

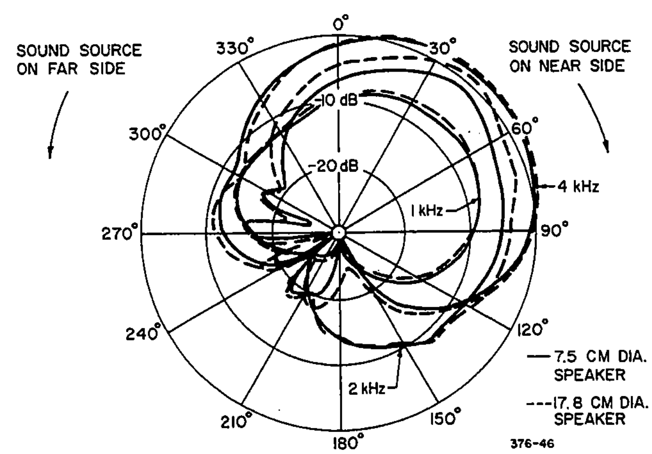


Fig. 10-9. Polar response of a directional hearing aid on the KEMAR manikin for three frequencies and for two different sized loudspeakers.

sound field about the axis of radiation of a loudspeaker will be limited by the cone driver system itself if care is taken to provide symmetrical mounting in a baffle or box.

A single driver loudspeaker system should be used for manikin measurements, as with listening tests with persons. A vented or base reflex type of speaker box may be used if the vent is concentric with the cone driver system. The important consideration is that the apparent source location should be independent of frequency. This will not be true for most modern home "hi-fi" speaker systems, because of the use of multiple drivers. The magnitude of the speaker size problem for a directional hearing aid is illustrated in Fig. 10-9.

Relative polar responses of a directional hearing aid on the KEMAR manikin are shown for three frequencies and two sizes of loudspeaker. The diameter refers to the mounting hole in the box containing each of the cone type loudspeakers. The 8" unit was mounted in a .014 cubic meter box; the 3" unit was in a .0115 cubic meter box. Only the 8" unit was covered with sound absorber to reduce reflections and standing waves between the box and the manikin. The distance between the loudspeaker and the manikin test point (head center) was one meter.

If one postulates that the test conditions should approach face to face speech communication, then a small size sound source is desired, but adequate sound levels for most tests cannot be obtained at low frequencies. A further compromise between speaker size and test signal levels results from the distance that can or should be used for the tests. Greater sound field symmetry, a better approximation to a plane wave at the test location, exists at longer distances between the source and test location. Again a larger speaker must be used for greater distance. These conflicting requirements will have to be resolved and agreed to in measurement standards if uniformity of data is to be assured.

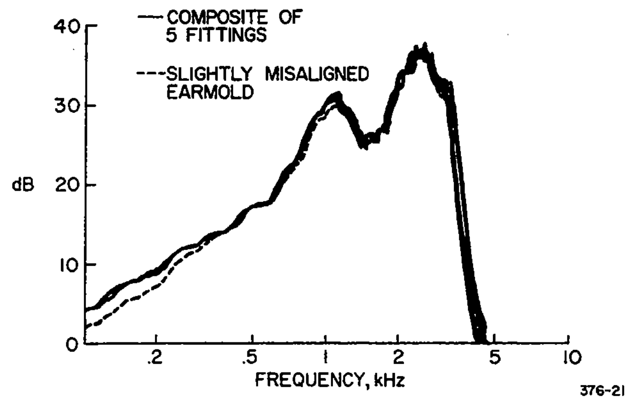


Fig. 10-10. Reproducibility of a hearing aid response curve measured on the KEMAR manikin.

Reproducibility of Hearing Aid Gain Measurements

A variety of questions can be asked concerning the reproducibility of hearing aid measurements on a KEMAR manikin. One example is given in Fig. 10-10. The hearing aid was removed and refitted onto the manikin a total of six times. In one of these the earmold was deliberately misaligned to cause, as it turned out, some low frequency leak. The earmold was an acrylic mold made from an impression of the KEMAR ear. There is evident in the figure some variation in the frequency synchronization of the chart and the oscillator. But in general the differences among the gain curves are not more than 1 decibel over most of the frequency range.

References:

Wansdronck, C. (1959) On the Influence of the Diffraction of Sound Waves Around the Human Head on the Characteristics of Hearing Aids. J. Acoust. Soc. Am. *31*, 1609-1612.

Shaw, E.A.G. (1974) The External Ear, a chapter in *Handbook of Sensory Physiology* Volume V/I edited by W.D. Keidel and W.D. Neff. (Springer-Verlag)

Chapter 11

Typical Hearing Aid Measurements on KEMAR

M.D. Burkhard
Industrial Research Products, Inc.

The KEMAR manikin was developed, as indicated before, to facilitate *in situ* measurements of hearing aid characteristics. Participants in these conferences and others have shown examples of hearing aid response curves taken in various ways on the manikin. Some examples are given here to permit some comparisons of the agreement among laboratories.

Because of the strong interaction between the hearing aid and the wearer, it becomes important to include the diffraction of the head in any directional hearing aid measurement. The nature of the open ear eardrum sound pressure as a function of direction to the sound source, was shown in Figs. 3-5 and 2-7. The presence of the head shadow, which has been discussed in many contexts in the literature is readily apparent. When a conventional pressure response hearing aid with its omnidirectional microphone over the ear is mounted on the manikin, much the same polar response patterns are observed, in that the head shadows of ear sound pressures are preserved. Fig. 11-1 shows, at 2000 Hz, how a hearing of this type has no directional response sensitivity when measured by itself in an anechoic room, and that the directional contribution of the wearer is introduced when in use. By contrast the polar plot of response for a "directional" hearing aid, is shown in Fig. 11-2. Again, the influence of the wearer on the polar response at 2000 Hz is evident. The simple cardioid forward pattern found in a free field has been shifted by the presence of the manikin so that the direction of maximum response is around 45° to the side. To obtain a realistic measure of the effectiveness of the directional hearing aid, the manikin can become an important and necessary element in the measurement and specification.

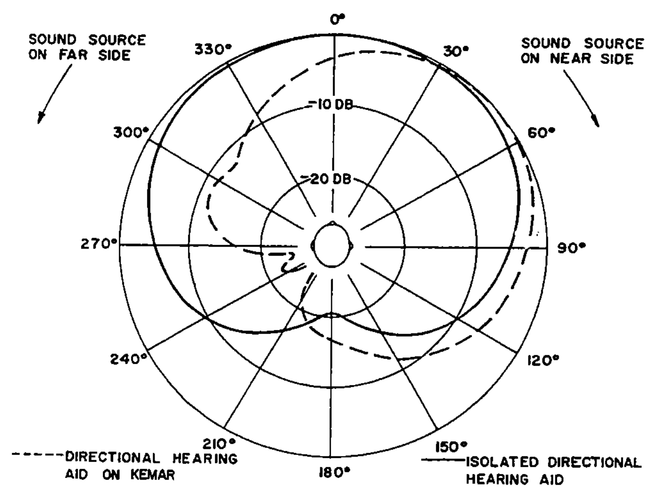


Fig. 11-2. The polar response at 2000 Hz for a directional hearing aid in a free field and mounted on the KEMAR manikin.

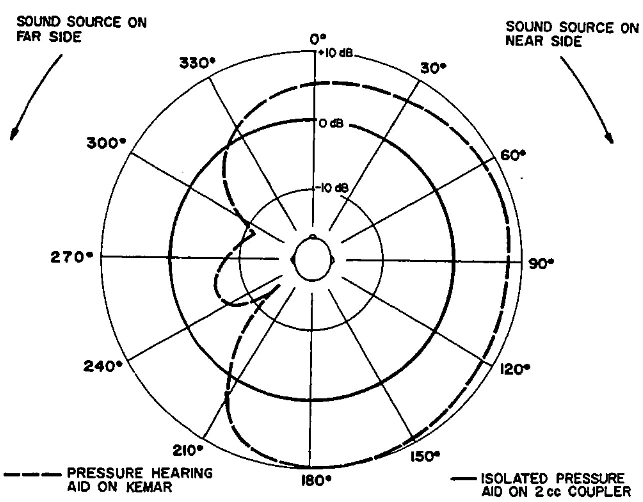


Fig. 11-1. The polar response at 2000 Hz for a non-directional type hearing aid in a free field and mounted on the right ear of the manikin. The microphone is at the bottom of a behind-the-ear hearing aid.

Another example of an *in situ* measurement in which the wearer's influence on the sound field is important is the various open ear hearing aid fittings. When gain is measured according to ANSI or IEC methods, the hearing aid being in a free sound field and with the output sound pressure measured in a 2cm³ coupler, the dash-dot curve in Fig. 11-3, is recorded. When the same hearing aid is applied as a CROS fitting on a KEMAR manikin, the gain relative to free field, i.e., the sound pressure at the manikin eardrum for a free sound field at the point where the center of the manikin head is placed, plotted as the dashed and solid line curves are obtained. The pressure at the manikin eardrum without the hearing aid turned on is shown by the dotted curve. The two CROS fitting curves illustrate the change of gain that can be introduced by changing the depth of insertion of the tube in the ear canal, 4 mm and 13 mm, respectively. A quantitative indication of the CROS fitting gain is possible because the manikin ear accurately reproduces the standing-wave patterns of a typical ear. The eardrum impedance at the end of the ear canal produces both a sound absorption and a sound reflection like a real ear and the sound is radiated out of the open ear canal in a manner similar to a real ear. As a result, the high frequency emphasis of gain known to exist in open ear hearing aid fittings is depicted. Finally, the etymotic or insertion gain of the hearing aid for this fitting may be obtained by subtracting the open ear unaided pressure, dotted line, from the aided ear sound pressure, dash or solid line.

Having introduced the insertion gain, this is an appropriate place to discuss possible methods for measuring it directly.

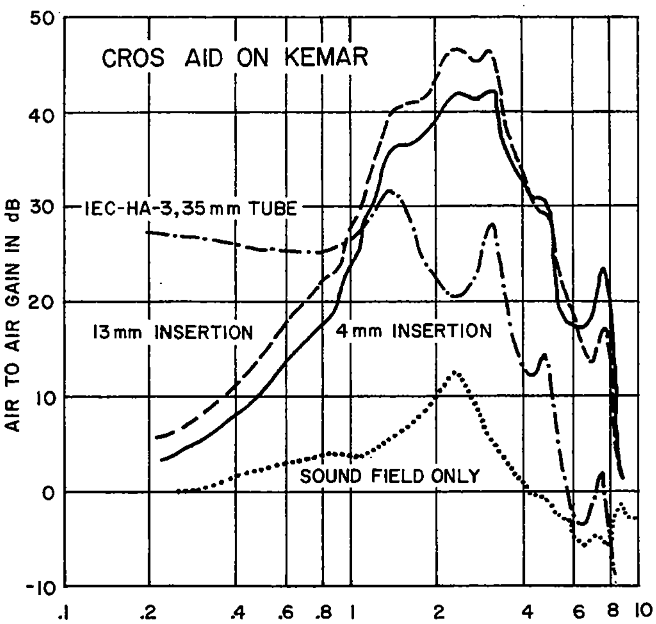


Fig. 11-3. Various gain curves for a CROS hearing aid on the KEMAR manikin and the same hearing aid measured on an HA-3, 2 cm^3 coupler system.

Measuring Insertion Gain

Insertion gain, etymotic gain and functional gain are considered to be the quantities really needed when predicting, by physical measurement or subjective listening tests, the benefit offered to the user by the hearing aid. A user receives sound at his ear drum by natural airborn sound propagated into and through his ear canal in the absence of a prosthesis. The prosthesis, i.e., the hearing aid, replaces or supplements that natural pathway. The important question to answer by the test or measurement, is: what is the change in sound pressure level at the hearing aid users eardrum, due to the aid. Some choices for this measurement with a manikin will be described briefly.

It should be noted that insertion gain is a function of location of the sound source relative to the active ear. Therefore, it will be necessary to specify sound source orientation. Sensitivity to orientation will make some measurement procedures more attractive than others.

Prerecorded Signal Method

In this method the system shown and discussed in connection with Fig. 10-1, is used, except that the microphone is now the eardrum microphone of the manikin ear to which the hearing aid earphone will deliver amplified sound. The signal level controlling system maintains a constant sound pressure at the manikin eardrum and the corresponding voltage at the amplifier is recorded when the switch is in the upper position. When the switch is in the lower position and the recording is played, a sound pressure, independent of frequency is produced at the manikin eardrum. When this same signal is played and the hearing aid is on the ear, and the conditions of measurement are unchanged except for application of the hearing aid, any changes from a constant sound pressure must be due to insertion of the hearing aid into the transmission

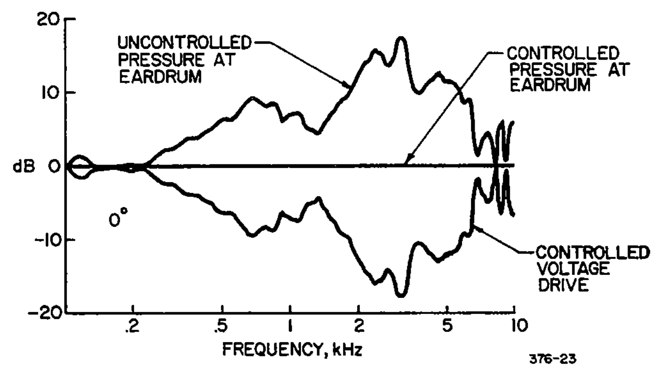


Fig. 11-4. The signals involved in producing a constant sound pressure as a function of frequency at the eardrum of the KEMAR manikin in an anechoic room. The manikin is 1 meter distant from an 8" loudspeaker.

path. Thus the method gives directly the insertion gain (or loss) of the device applied to the ear. Fig. 11-4 shows the signal at the eardrum of the KEMAR manikin as driven by a constant voltage to the power amplifier, the recorded signal that was created by the level control circuit and the resulting flat spectrum response at the eardrum. The prerecorded signal simultaneously compensates for the irregular response with frequency of the sound source and the manikin. It is our experience that this method is less sensitive to manikin or speaker orientation errors and speaker sound field asymmetry. The method may be used for any sound source direction from the manikin, if the particular controlled drive signal has first been recorded.

Other Ear Control Method

If it is assumed that the manikin and the sound field from the loudspeaker are both symmetrical about the axis of the speaker and the center of the head when the manikin faces the source, the signal from one eardrum microphone may be used in a servo-control system to maintain a constant eardrum sound pressure while the output of the hearing aid is recorded from the other ear. Both ears will of course have complete ear simulators, including "identical" microphones. The conditions for symmetry, in addition to those mentioned above, are achieved only for the manikin facing the sound source. It cannot be used for insertion gain measurements in any other orientation, because symmetry will not be maintained. Other ear control may not be desired for testing of CROS type fittings, because the presence of the hearing aid body has the potential for perturbing the sound field around the control ear excessively. This would have to be investigated for the particular hearing aid. Asymmetry due to the sound source can be minimized by increasing the distance to the manikin. The principal advantage of the other ear control method is avoidance of a prerecorded signal drive system, but an equalizer filter may be required in one of the microphone channels to compensate small differences of asymmetry. The effects on responses of errors in pointing of the manikin toward the loudspeaker or placing it on the axis of the loudspeaker tend to be doubled when other ear control is used.

Etymotic or Insertion Gain Measurement: Critique of Methods

Since, in my opinion, the effectiveness of a hearing aid fitting should be judged by its etymotic or insertion parameters, it is important that the differences between basic methods of measurement that have been discussed be emphasized because the published properties of gain, saturation output, etc. may be affected. This is done at the risk of some redundancy. To recapitulate, there are three methods that have been discussed which can be enumerated briefly as follows:

Method 1. Refer all measurements to a constant free field sound pressure sans manikin. (A prerecorded test signal or an equalized sound source system is used in an anechoic room.)

- Record free field open ear eardrum sound pressure level
- Record eardrum sound pressure level with hearing aid in place
- Calculate or otherwise produce the difference of these two response levels. (See the example and discussion in connection with Fig. 11-3.)

Method 2. Control eardrum sound pressure in the test ear. (This will be done with a recorded test signal or equalized sound source system that gives a constant sound pressure to the eardrum of the manikin sans hearing aid.)

- Record eardrum sound pressure level with the hearing aid on the test ear.

Method 3. Control eardrum sound pressure in the other ear. (This uses a sound level servo-control system or an equalized sound source system)

An analysis of the three methods will readily show that the pressure at the hearing aid microphone and also the output pressure level in the ear is not the same in the three methods. To illustrate the difference, we define the following terms with the aid of Fig. 11-5.

d_{mc}	$= p_{mc}/p_{ff}$, diffraction factor converting p_{ff} to p_{mc},
d_o	$= p_{eo}/p_{ff}$, diffraction factor for converting p_{ff} to p_{eo}.
g	$= p_{ec}/p_m$, pressure gain for the hearing aid (see Chapter 12)
K_e	= The calibration constant for the loudspeaker (including power amplifier) to the eardrum of the manikin when it is mounted in the field.
K_f	= Calibration constant for the loudspeaker (including power amplifier) to the free field test location.
p_{eo}	= open ear sound pressure at the eardrum
p_{ff}	= free field sound pressure at the test location
p_{mc}	= pressure at the hearing aid microphone, ear closed.

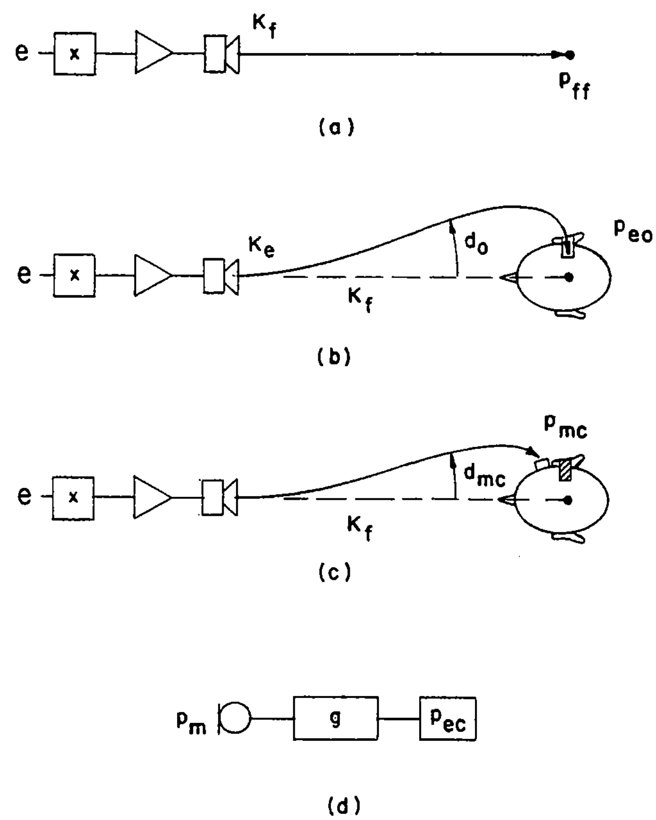


Figure 11-5. Representation of sound from a voltage source through a signal conditioner, a power amplifier and a loudspeaker to the measurement point (a), the eardrum of the open ear of the manikin or a person (b), or the pressure to the microphone of a hearing aid (c). A hearing aid with microphone gain g and output pressure p_{ec} into a closed ear is given by (d).

$T_e = p_{eo}/p_h$ pressure transfer from the ear entrance to the eardrum.

x = optional operator for keeping a constant sound field.

e = system signal voltage, constant.

For a servo-control system or an equalizing filter producing a constant free-field, for example, $x = 1/K_f$.

Method 1. The pressure level at the microphone contains only the deviation from a constant free sound field pressure that would be due to the diffraction of the head and/or torso. Pressure level in the manikin ear, i.e., the hearing aid output pressure in the ear, deviates from constant to the extent that the pressure at the microphone and the inherent pressure gain of the hearing aid are not constant.

Thus from the above definitions, the pressure at the hearing aid microphone is, for Method 1,

$$p_{mc1} = d_{mc}\, p_{ff}$$

and

$$p_{mc1} = d_{mc} e, \text{ when } x = 1/K_f$$

Similarly, the closed ear pressure due to the hearing aid gain is, for Method 1.

$$p_{ec1} = g\, d_{mc}\, e$$

Methods 2 and 3. The pressure at the hearing aid microphone location differs from the pressure produced in Method 1. The deviation is due to diffraction as indicated in Method 1 plus deviations due to the transfer from the ear entrance to the eardrum.

Using the above definitions, the pressure at the hearing aid microphone is for Methods 2 and 3,

$$p_{mc2} = g\, d_{mc} e\, K_f / K_e,$$

In this case

$$x = 1/K_e$$

Similarly, the closed ear pressure due to the hearing aid gain is, for method 2,

$$p_{ec2} = g\, d_{mc}\, e \frac{K_f}{K_e}$$

The ratios of the two microphone pressures and the two closed ear canal pressures are the same,

$$\frac{p_{mc2}}{p_{mc1}} = \frac{p_{ec}}{p_{ec1}} = \frac{1}{K_e}$$

The curve labeled "uncontrolled pressure at eardrum" in Fig. 11-4 is a plot of K_e for our anechoic room and 8 inch diameter loudspeaker and is just the ratio (difference in dB) of the pressures incident on the microphone of the hearing aid for the two methods, as well as the eardrum location. Note that the results for Method 3 will be the same as for Method 2, to the extent that symmetry is maintained.

These results may be restated by recalling that the servo-control and spectrum equalizer methods for maintaining constant sound pressure will have the following effects on the pressure at the hearing aid microphone location for the three methods of measurement:

Method 1. Pressure at the hearing aid microphone increases relative to the free field sound pressure if a diffraction of the head and torso causes an increase and *vice versa,* if the sound pressure at the hearing aid microphone location goes down relative to the free field as a result of diffraction, the pressure at the hearing aid microphone will decrease.

Methods 2 and 3. The pressure at the hearing aid microphone decreases if the diffraction and ear canal resonance causes increased eardrum sound pressure in the open ear canal and, vice versa, the pressure at the hearing aid microphone will increase if diffraction and canal resonance causes decreased eardrum pressure in the open canal.

The differences are unimportant if gain of the hearing aid is the only parameter of interest and if the hearing aid is operating without overload or distortion and hence in a linear transfer region in each test condition.

Using the above definitions it is easily shown that the insertion gain, by either method is,

$$\frac{p_{ec}}{p_{eo}} = g\frac{d_{mc}}{d_o}$$

The ratio d_{mc}/d_o is the difference, in dB, between the pressure at the open ear eardrum and the pressure that will be incident on the hearing aid microphone as worn.

Substantially different results may be created among the methods, for example, for measurements of

a. Equivalent noise input.
b. Saturation level, maximum output pressure, maximum input pressure, etc.
c. Distortion when referenced to either the input or the output.

It is important to repeat that contrary to previous practices for hearing aid evaluation, *in situ* and in particular insertion measurements or etymotic measurements are produced with neither a constant input pressure at the hearing aid microphone nor a constant output pressure into the coupler or ear simulator. The deviations from constant input or constant output pressures will be different for method 1 than for methods 2 and 3. It is my recommendation that the preferred measurement for the hearing aid characteristics *in situ,* other than gain, should be made or obtained with a condition of constant free field sound pressure and that they should be made, however, on the basis of insertion, i.e., the comparison of the aided to the unaided ear conditions.

Chapter 12.

Estimating *In Situ* Gain Without A Manikin

M.D. Burkhard
Industrial Research Products, Inc.

It is not always convenient or economically feasible to measure *in situ* gain and response of hearing aids with a manikin. Some of the *in situ* data, primarily the basic insertion gain or response information, may be obtained in a sound box, with less accuracy. Here, we present a scheme for measurements and/or corrections that can be made in a sound box if basic data about the sound pressure on the head of the manikin are available for the particular hearing aid design.

The usual practice for hearing aid measurements in sound boxes, or anechoic spaces, has been to impose the "free field" conditions assumption. The hearing aid is assumed to be a calibratable entity by itself; the input sound pressure for the gain measurement is that present sans hearing aid. For the substitution method, the sound pressure is first measured at the location to be occupied by the hearing aid and then the aid is placed there and its output noted. For the comparison method, sound field symmetry is assumed so that the sound pressure at a standard microphone location is held constant while the hearing aid is placed at a symmetrical location in the sound field. The two, microphone and hearing aid, are separated from each other sufficiently to avoid the diffraction field of one affecting the diffraction field of the other. In both methods the diffraction of the hearing aid becomes part of its gain calibration. Even in the case of small head worn hearing aids the contribution of diffraction to net gain can be observed. The measured gain may be further affected by the necessary proximity of the output coupler, or ear simulator, and its associated amplifiers, if they are extra-ordinarily large or if the hearing aid is very small.

Fortunately, the diffraction of most hearing aids by themselves is relatively unimportant compared to the perturbing diffractions of the body and head of the wearer. But it is still preferable to have a measurement scheme that gives only the pressure gain of the hearing aid and adds the contribution of the head and body diffraction as a correction based on the pressure at the place where the microphone will be placed. It is important when this is done, though, to separate the diffraction contribution at the *input* to the hearing aid from the ear canal loading or impedance effects on the *output* of the hearing aid. Referring to the discussion of insertion gain, it is implicit that the signal being amplified by the hearing aid is the pressure at the location of the hearing aid microphone on the head. This input pressure varies with the location, frequency and sound direction. The data of Madaffari and of Burnett and Kuhn (Chapters 3 & 5) show this. For example, Fig. 3-9 shows the pressure on the surface of an occluding ear mold in a KEMAR manikin ear as a function of frequency for several sound directions, with a constant free field sound pressure. These curves represent the pressure as a function of frequency and sound source location, that exists at the microphone of a typical in-the-ear hearing aid when the KEMAR manikin is placed in a constant free field sound. Figure 3-10, shows similar data for an over the ear hearing aid microphone location, also with the ear canal and concha of the KEMAR manikin occluded.

Pressure Gain of a Hearing Aid

Basically, then, a pressure gain for the hearing aid rather than a free field gain is needed and is most useful for making the necessary corrections to predict *in situ* insertion gains from sound box measurements. To measure pressure gain of a hearing aid, the pressure at the hearing aid microphone is measured or controlled with a pressure type microphone, according to the scheme shown

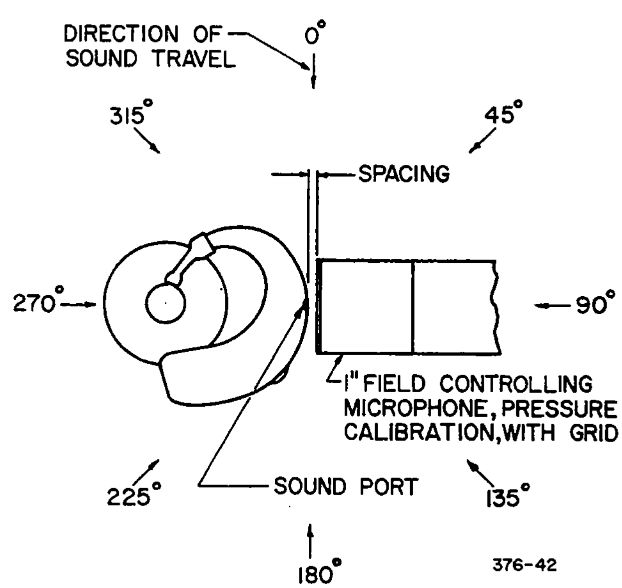


Fig. 12-1. Experimental arrangement for measuring the pressure gain of a hearing aid.

in Fig. 12-1. Even though there is sound diffraction in the region, the reference microphone senses only the sound pressure at its diaphragm. As shown in Figs. 12-2, 12-3 and 12-4, the output of a hearing aid microphone, and hence the hearing aid gain,

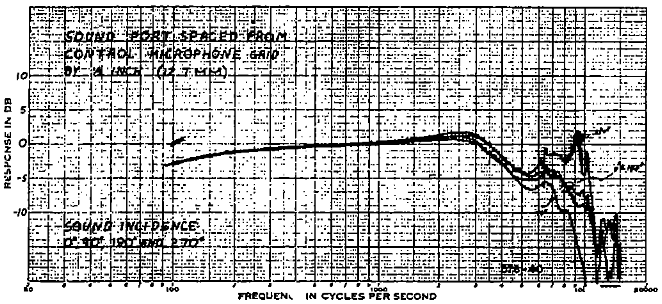


Fig. 12-2. Output of a hearing aid microphone with a 1" condenser microphone controlling the sound field 1/2" from the port of the hearing aid, for various sound incidence angles.

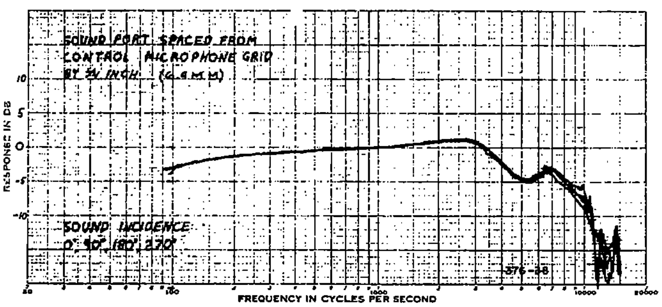


Fig. 12-3. Output of a hearing aid microphone with a 1" condenser microphone controlling the sound field 1/4" from the port of the hearing aid for various sound incidence angles.

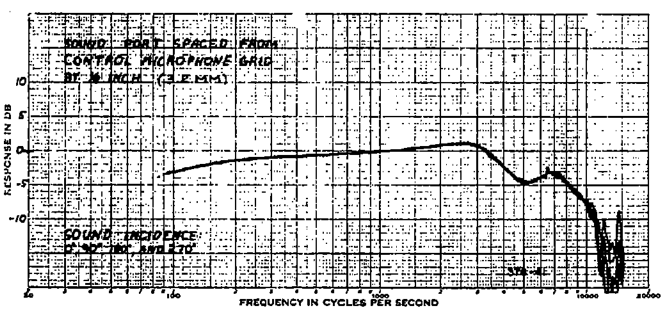


Fig. 12-4. Output of a hearing aid microphone with a 1" condenser microphone controlling the sound field 1/8" from the port of the hearing aid, for various sound incidence angles.

becomes independent of sound direction and hence hearing aid diffraction as the control microphone is placed closer to the hearing aid. As the hearing aid is placed in various orientations in the sound field the microphone sound port is shadowed by various amounts. The output of the reference *pressure* microphone, never-the-less, is proportional only to the pressure at its diaphragm, as just mentioned. When the hearing aid and the microphone are separated by 12.7 mm, 1/2", there is sensitivity to angle of incidence of the sound because of the shadow of the hearing aid. But the hearing aid is small enough that the shadow is unimportant for frequencies below 2000 Hz and less than 1 dB variation is evident for frequencies below 5000 Hz. As the microphone to hearing aid separation decreases, the frequency range for good control of the sound pressure at the hearing aid microphone increases. With a 3.2 mm, 1/8", separation the output of the hearing aid microphone is independent of the direction of sound incidence for frequencies up to 10,000 Hz. From this observation, one must conclude that the sound pressure is being controlled to a constant level by the reference microphone. E.V. Carlson took this data with a 1" condenser microphone. Equally good consistency can be observed with a 1/2" reference microphone, although microphone size is not an important parameter if it has less than 1/2 wave length diameter at the highest frequency of interest. Since the pressure at the hearing aid is being controlled by this system the possibility of arbitrary input sound pressures as a function of frequency exists.

There may be a philosophical problem of where in the measurement sequence to apply the pressure variations due to the head and body diffraction. The idea of a constant free field input pressure for all hearing aid measurements has a long history of use. But Figs. 3-9 and 3-10, show that in general, a head worn hearing aid receives a sound pressure that varies with frequency. It seems reasonable therefore that the stimulus pressure for a hearing aid measurement should carry the variations as created when the aid is worn, and that all the pertinent parameters be determined not with a constant input, but with the variable (with frequency) input. Thus, for an in-the-ear hearing aid with the sound source in front of the person, a pressure versus frequency of the form shown in Fig. 3-9, for 0° would be applied to the hearing aid microphone. This might be done with filters on the electrical signal that drives the loudspeaker or it may be included in servo level control circuits associated with a standard reference microphone.

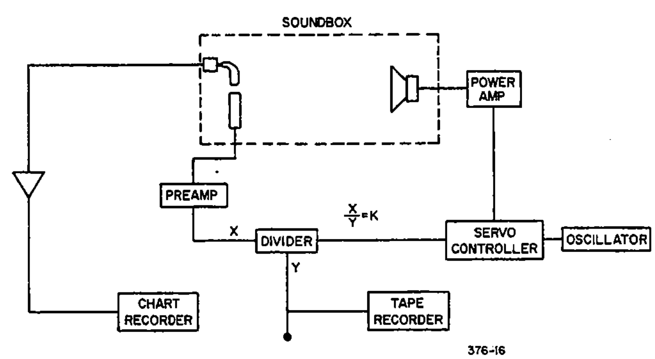


Fig. 12-5. Measurement and sound level control system for simulating the measurement of a hearing aid *in situ* gain without a manikin.

If the variable pressure function desired at the hearing aid is incorporated into a servo control system, a filter with the inverse of the desired function is inserted between the microphone and the rectifier of the controlling modulator or multi-

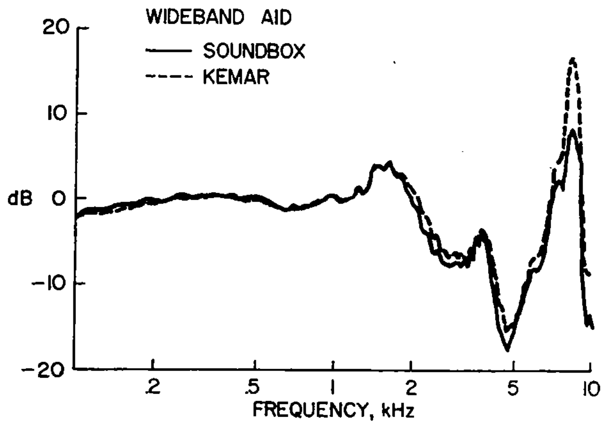


Fig. 12-6. Two response or gain curves showing the hearing aid gain measured on the KEMAR manikin and a hearing aid gain measured by the corrected pressure gain sound box method.

plier. The system then acts to control the pressure at the microphone in exact opposition. When the filter causes an increase in the control signal from the microphone, the controller causes a decrease in the signal driving the loudspeaker, and vice versa. This results in a sound pressure at the hearing aid corresponding to the pressure at the side of the head of a wearer. Under some circumstances, the function may be simple enough that filters are practical, especially if a given microphone position and source direction are to be used repeatedly. In other cases it may be worth the somewhat more complicated, but more flexible, procedure of storing the input pressure function and using it to modify the control circuits. A system that will do this is shown in Fig. 12-5. Its presentation here is primarily for illustration of the principle. Modern digital data storage and manipulation methods can undoubtedly result in simplifications.

In Situ: Free Field Reference

The first step in using the stored signal control method in a sound box, or sans manikin situation, is to obtain a stored signal itself, i.e., the pressure on the side of the head where the hearing aid microphone is to be placed. It seems quite reasonable that a tape recorded signal might be supplied for this purpose. The signal would be generated with a two channel tape recorder on which the channels can be recorded independently. The signal that produces a constant sound pressure at the test location in an anechoic room is first recorded on one channel. Next the manikin is placed in the sound field at the test location with the ear occluded appropriately or partially blocked, according to the type of hearing aid being simulated. A miniature microphone or a probe microphone is then placed on the manikin head at the location for the hearing aid microphone of the particular hearing aid design. It is recommended that a complete hearing aid case be used. The microphone may be in the hearing aid or may be adjacent to the hearing aid port, the latter being preferred when standard microphones are used. In fact if the first recording is made with the hearing aid microphone that will be in the hearing aid when the second recording is made with all other parts of the system unchanged, the "standard" microphone need not be used; the shortcomings of microphone response cancel and the recorded signal is the same as if two ideal flat response microphones were used. The recorded signal that produces constant sound pressure at the test location is then played back and the signal from the microphone just mounted on the manikin with the hearing aid is recorded on the second channel. This is in fact exactly the procedure used for recording the response curves in Fig. 3-9 and 3-10 except that the signals from the head mounted microphones were recorded on a chart recorder. The signal that is recorded on the second channel is the signal that is desired at the input of a hearing aid in the sound box, or in any hearing aid measurement sans manikin. It is the signal to be played from the tape recorder in Fig. 12-5.

If the sweep of the oscillator and the recorder are at the same rate, the combination of the tape play back signal and the divider, acting on the input of the reference microphone, is equivalent to inserting a complimentary filter in that circuit. It is of course preferable that the output of the hearing aid be into an ear simulator rather than a $2cm^3$ coupler. If a $2cm^3$ coupler is used, then additional corrections to the gain would be required to account for the differences between the coupler and the ear simulator that are now well known, and shown in Fig. 15-3.

In Situ: Eardrum Reference

As has been discussed, the insertion gain or etymotic gain is the most useful and perhaps the most important hearing aid fitting parameter. A good sound box estimate of etymotic gain can be achieved with an approach that is similar to the one just described. The first step, as before, is to obtain a correct stored signal for replay or for creating a filter network. The signal to be played from the tape recorder in Fig. 12-5 in this case, is generated in the following way: The signal that produces a constant sound pressure at the manikin ear drum is recorded on one channel, for the etymotic or insertion gain measurement. A level controlling system is used with the manikin in an anechoic room to create a sound pressure that is constant with frequency at the eardrum of the manikin. Next, the ear canal and concha of the manikin are occluded in the way that duplicates the way the hearing aid is to be worn. A microphone is placed at the location to be occupied by the hearing aid microphone, also. It may be desirable to also include a hearing aid shell to complete the fine details of the diffraction pattern. Finally, the just recorded signal is played back and the signal from the external microphone is recorded on the second channel of the tape recorder. This is the signal that is desired at the input of a hearing aid in a sound box, or in any measurement sans manikin, to record directly the insertion gain of a hearing aid. It is the signal to be played from the tape recorder in Fig. 12-5. As before, if the sweep of the oscillator and the recorder are at the same rate and synchronized, the combination of the tape play back signal and the divider, acting on the output of the reference microphone, is equivalent to inserting a complimentary filter in that circuit. It will be noted that this set-up is essentially the same as shown in Fig. 10-1, which is the conventional servo control level system, with the addition of the signal conditioner function K.

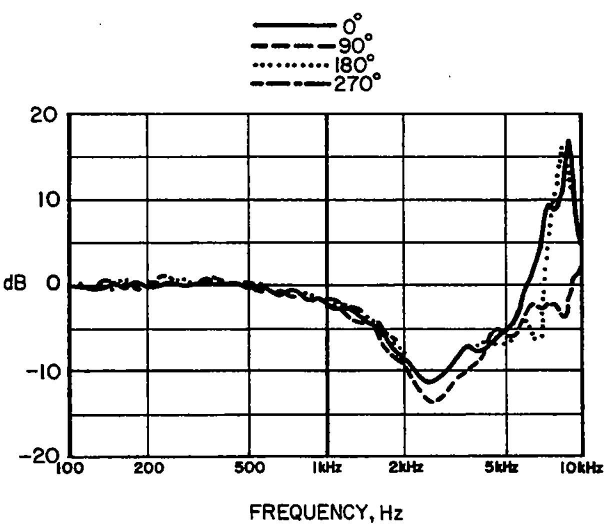


Fig. 12-7. Function to convert a pressure gain measured in a sound box with a Zwislocki type ear simulator into an etymotic or insertion gain. This is for an in-the-ear hearing aid with an occluded ear fitting.

As before, the output of the hearing aid would be fed into an ear simulator to produce the best approximation to insertion gain. If a 2cm^{3}ANSI or IEC coupler were used, additional corrections to the gain would be required. The agreement between an insertion gain measurement simulated by controlled sound field method just described, and a measurement on the manikin in an anechoic room is shown in Fig. 12-6. The hearing aid is a laboratory combination of microphone, amplifier, and experimental earphone, which accounts for the unconventional response shape. It does show that the extra step of correcting for the hearing aid entrance pressure electronically in the control circuits produces results that agree quite well with the gain measured directly on the manikin. The scale of gain on the ordinate of the figure is arbitrarily normalized to "0" dB at 1 kHz.

The pressure versus frequency for the hearing aid microphone input for the insertion gain meas-

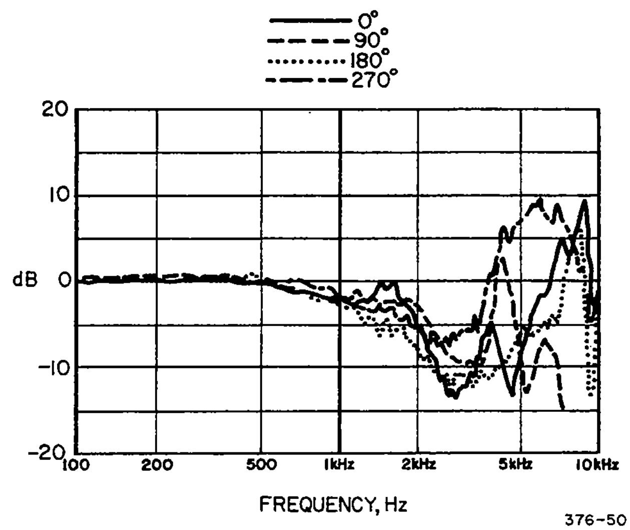


Fig. 12-8. Function to convert a pressure gain response of a hearing aid measured in a sound box with a Zwislocki coupler into an etymotic or insertion gain. The hearing aid is an over-the-ear, top microphone type.

urement are illustrated in Fig. 12-7 for a typical in-the-ear hearing aid, in Fig. 12-8 for an over-the-ear microphone location, and in Fig. 12-9 for a behind-the-ear lower position microphone. In each case the pressure is shown for several source positions. For an in-the-ear hearing aid for which this method is particularly attractive, it appears that a single function could be used for frequencies up to 5,000 Hz, that would apply to all sound source orientations. Unfortunately the situation is not quite so nice for the over-the-ear and the behind-the-ear microphone locations. This is apparently due to the greater distance from the ear opening in these cases. It has been pointed out by Shaw and others that the sound pressure at the entrance to the ear canal is, on the average, directly related to the pressure at the eardrum and

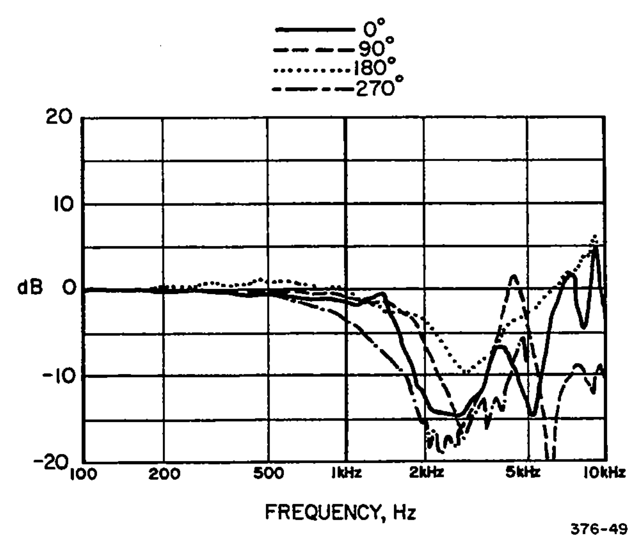


Fig. 12-9. Function to convert a pressure gain of a hearing aid measured in a sound box with a Zwislocki ear simulator into an etymotic or insertion gain. This is a behind-the-ear hearing aid with bottom microphone.

independent of direction.

Several points about the method should be noted. The recorded signal is obtained with the ear canal closed in the way the hearing aid will be used. This means total occlusion and filled concha, especially for in-the-ear type hearing aids. Resonances in the open ear extend sufficiently far from the ear canal to affect the pressure at likely head worn hearing aid microphone locations, and the resonances of course are not there when the ear is closed. The control signal is unique to the sound source direction. This is relatively unimportant for in-the-ear hearing aids. But for others, a different recorded signal is required for each direction to be investigated. In other words, the control signal is dependent on the location of the microphone, so that different hearing aid designs require different stored signals. The method produces a sound pressure at the hearing aid microphone and hearing aid output that varies with frequency as discussed in Chapter 11. This method has not been investigated for open ear type hearing aid fittings. There is reason to believe that it may not be practical and further investigation for these applications is warranted.

Chapter 13.

Non-Hearing Aid Uses Of The Kemar Manikin

M.D. Burkhard
Industrial Research Products, Inc.

It should be clear throughout the material presented in these proceedings, that the primary concern, when the KEMAR manikin was designed, was to provide a tool for predicting *in situ* performance of various hearing aids and hearing aid fittings. Questions were immediately asked about the possible use for other types of measurements; e.g., earphones, ear defender and bone conduction transducer calibration.

Earphones on the KEMAR Manikin

Two examples of responses of earphones intended for over-the-ear use are given first, one supra aural and one circumaural. In Fig. 13-1 the response of a TDH-39 earphone in an MX41/AR cushion is shown with an NBS 9-A coupler, a complete Zwislocki type ear simulator, and on the KE-

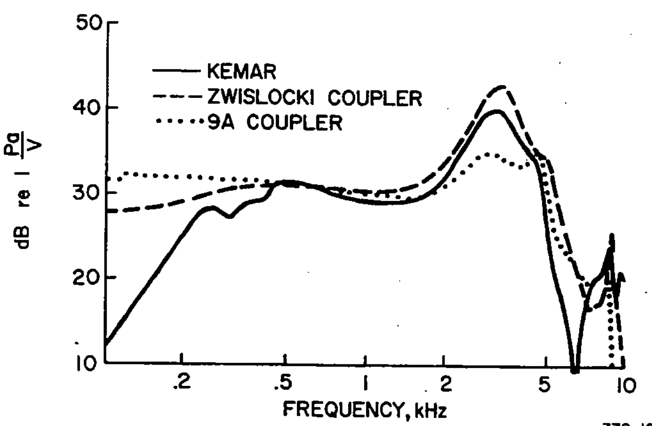


Fig. 13-1. Response of a TDH-39 earphone on KEMAR, on a Zwislocki type ear simulator and on a 9A coupler.

MAR manikin. Since the 9-A coupler seals the earphone to the coupling volume, there are no low frequency acoustic leaks, and the contributions of the ear canal geometry to frequency response at the microphone or ear drum location are missing. Because the complete Zwislocki coupler structure, as used by us, also seals to the earphone cushion, and the low frequency leak is omitted, there is no low frequency loss of response in this coupler either. The Zwislocki simulator does include a network in its concha region that is intended to duplicate the compliance and losses associated with bulk compression of the pinna. This gives rise to the lowered response at frequencies below .5 kHz. The pressure increase at the microphone or eardrum location due to the ear canal standing wave resonance is evident around 3 kHz. The response on the KEMAR manikin shows most of the features included in the Zwislocki simulator. In addition, a low frequency acoustic leak is evident as a loss of low frequency response. There is a deep anti-resonance or response minimum at 6.5 kHz that may be attributed to acoustic modes in the pinna and concha coupled with the ear canal, that are not present in the simple cylindrical concha simulation of the Zwislocki device.

To answer the question of how realistically the acoustic leak is reproduced, this data may be com-

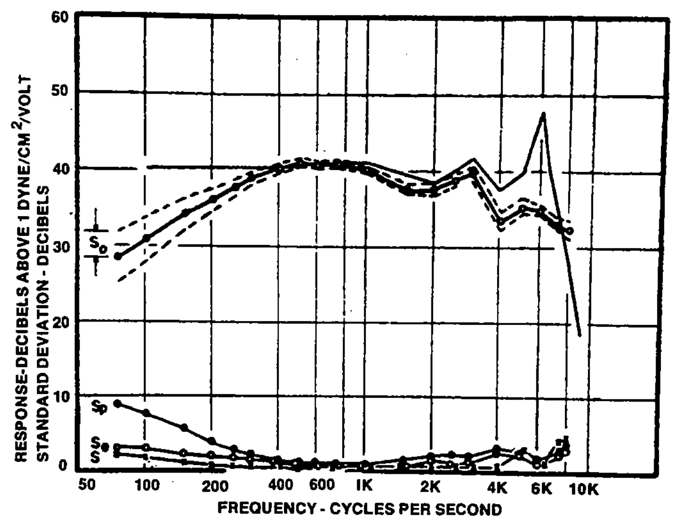


Fig. 13-2. Response of a Permoflux type PDR-8 dynamic earphone in an MX41/AR cushion measured by probe tube on persons. (Burkhard and Corliss (1954)). The average response for the subjects as shown with the root variances of the mean S_o, of measurements among people S_p, between the two ears of a person S_e, and for measurement error S.

pared with published earphone responses on a group of subjects, Fig. 13-2 (Burkhard & Corliss 1954). The PDR-8 earphone shown, also in an MX41/AR cushion, is quite similar to the TDH-39 over most of the low and middle frequency range where a comparison would be valid. This data on people was obtained with a calibrated probe microphone at the entrance to the ear, whereas the responses in Fig. 13-1, were obtained with the microphone at an eardrum like location. While the leak on the manikin is quite large, it appears to be consistent with the range of leaks found on individuals. The leak occurs because the pinna of the KEMAR was made somewhat thicker than a real ear to ensure adequate robustness in the normal handling during hearing aid measurements. As a result, the rubber pinna isn't as compliant and cannot be compressed as much as a typical human pinna, with typical headband pressures.

An example of a circumaural earphone calibration on the KEMAR manikin, is shown in Fig. 13-3, together with a calibration on the complete Zwislocki ear simulator system. The earphone is a Sharpe type HA-10A. In general the results by the two methods are similar. The apparent shift of frequency for maximum pressure response and the difference of response below .5 kHz, is consistent with an explanation that the earphone to microphone distance is slightly longer on the finished manikin ear than on the Zwislocki type simulator. The volume attributable to the added length would produce a lower pressure level at low frequencies on the manikin.

At this time we cannot say whether responses represent a true measure of the pressures in average real ears and whether they may be converted directly into corresponding threshold sound pressures for these earphones when used in audiometry. Loudness balance, threshold transfer or other appropriate psychoacoustic experiments will have to be done to answer that question.

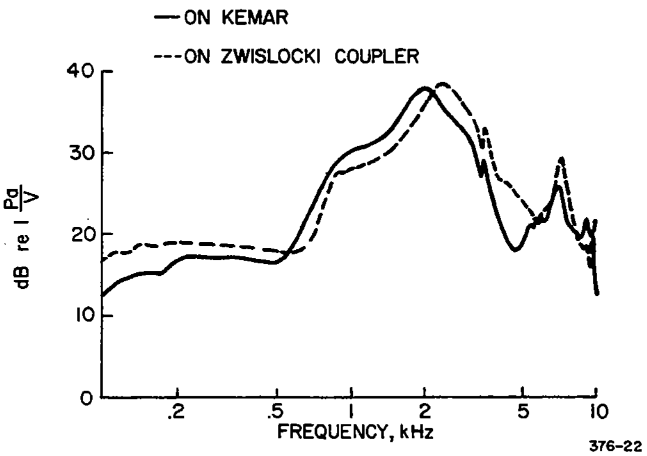


Fig. 13-3. Response of a Sharpe HA-10A circumaural earphone on a Zwislocki coupler and on the KEMAR.

Ear Defenders on the KEMAR Manikin

Attentuation of a compliant plastic foam type of earplug, namely an E.A.R. has been measured on the KEMAR manikin. The results are shown in Fig. 13-4, with the manufacturers data and data from Tobias. Attenuation on the manikin is greater than shown by the subjective data. There are apparently sound paths to the inner ear in people, that are not duplicated in the KEMAR manikin. The results are consistent with the theoretical predictions of ear plug attenuation made by Zwislocki (1955) and shown in Fig. 13-5.

Two effects are operative in the model. The first is gross motion of the plug in the ear canal due to

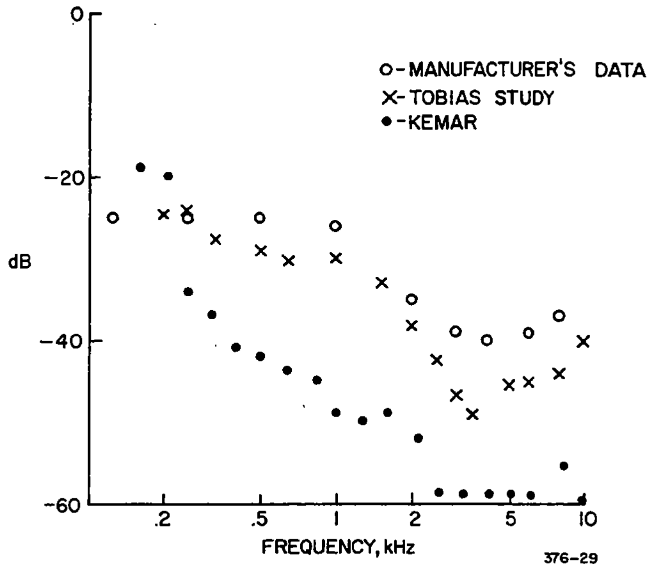


Fig. 13-4. Foam, EAR type, ear protector attenuation on a KEMAR manikin ear and published data. The manikin measurement was done with 1/3 octave bands of noise in an anechoic room.

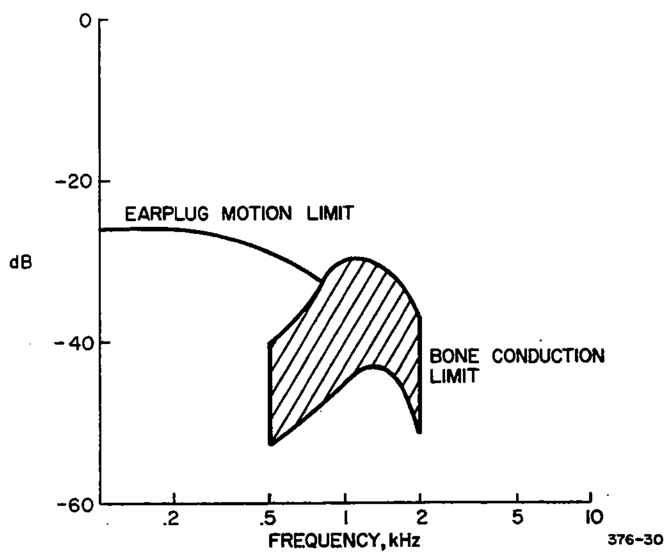


Fig. 13-5. Theoretical maximum attenuation for ear protectors according to Zwislocki.

the force of the sound pressure on it. This causes volume changes in the occluded space that translate to pressure on the eardrum. The second is a shunt path transmission by bone conduction. Since duplication of bone conduction paths of the human head were not considered in the KEMAR manikin design, attenuation at higher frequencies were expected to differ from both the human attenuation data and the theoretical predictions. The low frequency attenuation of the foam plug may be greater than in the model because it has compressibility which was not considered.

Bone Conduction

As just mentioned, bone sound conduction simulation of the human head was not included in the manikin design. The considerations are markedly different from air conduction and would add significantly to the complexity of the manikin if attempts were made to include them.

References.

Burkhard, M.D. and Corliss, E.L.R. (1954) Response of Earphones in Ears and Couplers, J. Acoust. Soc. Am. *26*, 679-685.

Tobias, J.V. (1975) Earplug Rankings based on the Protector—Attenuation rating (P-AR) Report FAA-AM-75-11. Federal Aviation Administration. (National Technical Information Service. Springfield, Va.)

Zwislocki, J.J. (1955) Semiplastic Earplugs. J. Acoust. Soc. Am. *27*, p 460

Chapter 14.

Considerations For Standardizing Artificial Ears

Ole Lauridsen

Toepholm - Westerman

Presented in Zurich, March 4, 1976

On behalf of the three Danish hearing aid factories, Oticon, Danavox and Widex, I should like to propose some requirements which the hearing aid industry feels a new artificial ear should provide. I am sure all of the participants in this meeting will agree that it is desirable that we achieve an artificial ear which more closely resembles the acoustic properties of the human ear than the 2cm³ coupler described in IEC R126. As nearly all the audiometic tests on hard of hearing people are done by supra aural earphones, it will be of great importance in the clinical fitting of hearing aids to have the possibility of relating pressure measured in the concha with supra aural earphones to actual ear drum pressures, so that the data given by the hearing aid manufacturers in terms of dB sound pressure levels measured at the ear drum can be of practical use, when applied to wide range hearing aids, in which the high frequency response goes far beyond the limits of the 2 cm³ coupler.

We feel that a time may arrive when different hearing aid manufacturers publish data on different unauthorized artificial ears, if we do not very soon succeed in standardizing a new ear.

In the following, we offer some comments on the ear mold simulator and afterwards on the coupler itself. Normally the peaks and valleys of the frequency response for a behind the ear hearing aid are governed by the connecting tubes. In other words if we choose the entrance of the 18 mm long, 3 mm diameter tube in the 2 cm³ coupler as the reference plane, one can hardly depend on the acoustic impedance at this place, when we consider frequencies higher than 1500 Hz, to be a measure of which coupler is at the other end of that tube. It's completely short circuited for these high frequencies. Of course, with a correction you would get a smaller peak to valley ratio in the coupler which contains some internal damping. Since it is the relationship of the diameter of the tubes at the interconnections that controls the standing wave ratio dramatically, we want the tube connection to the ear mold simulator to stay standardized. We would like them to be as at present and in accordance with IEC R126. These connection adapters have proved extremely practical even when large numbers of earphones and hearing aids have to be measured in succession.

In addition, it will be advantageous if the new coupler and the standard ear mold simulator can be screwed together into one mechanical unit like the old 2 cm³ coupler. This should, of course, not rule out the possibility of separating the coupler in the middle of the ear canal at a reference plane. That's where one would like to define the acoustic impedance and provide for connection of adapters for all the uses of an artificial ear for hearing aid measurements.

Also in the future, thousands of hearing aids will have to be checked in small sound boxes. The new coupler must, therefore, have small outer dimension so that diffraction problems will not arise. As we, at the same time, require good long term stability and reproducibility, we would rather accept an electrical correction and consider the impedance simulation as the main design goal to obtain the simplest possible construction.

We would also like the 2 cm³ coupler to be retained as well for use in cases where a reference measurement is adequate for testing a hearing aid whether or not it is a good simulation of a real ear.

We are aware of three alternatives to the 2 cm³ coupler in existence at the present time: Zwislocki and Diestel and B&K couplers, all of which use acoustic resistances in the side branches realized by scintered metal discs. Therefore, should a user want to check whether his coupler is all right, he must have access to equipment for measuring the transfer as well as the impedance function vs. frequency. It therefore seems logical to define a new coupler, not only by its mechanical dimension but also by its transfer and impedance function. In this context, it also seems reasonable to standardize a function for the transformation of pressure ratio between the concha and the ear drum if the coupler is equipped with an ear canal and a concha so that one can measure the resulting eardrum pressure from a supra aural earphone, if it is possible.

Chapter 15.

Ear Simulators, Designs, Stability, Etc.

M.D. Burkhard
Industrial Research Products, Inc.

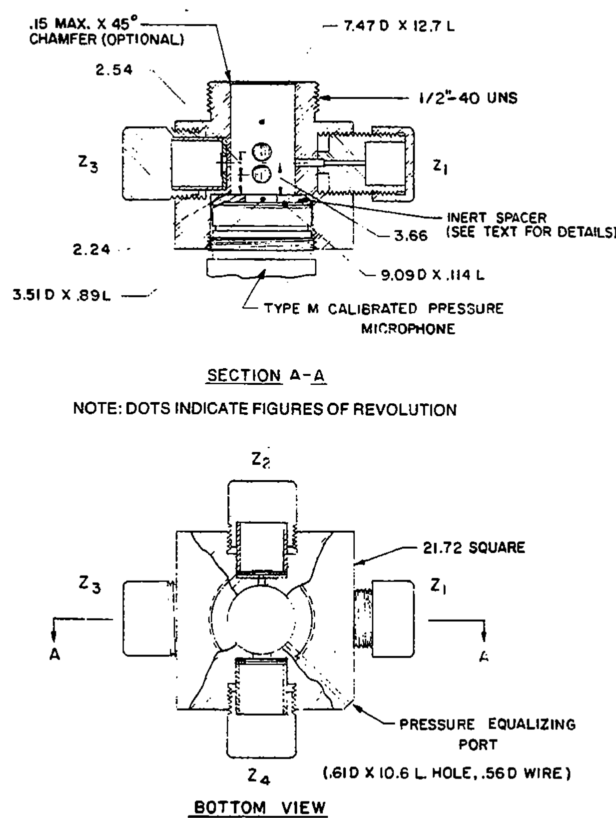
An important element of the KEMAR manikin that distinguishes it from other manikins used for acoustic measurements, is the inclusion of the ear canal and eardrum simulation structure. As used presently, the ear simulation is based on a construction proposed by Professor Zwislocki (1971). A number of improvements have been made which should be reviewed here. The changes significantly improve the stability and reproductibility of the structure. Methods have been devised to determine the quality of this type of simulator*. Stimulated in part, no doubt, by the ear simulator proposal of Zwislocki, several alternative constructions have recently been proposed and will be compared.

Zwislocki Type Ear Simulator

This ear simulator uses four side branch acoustic meshes, comprising inertance, resistance and compliance, to create an acoustic impedance approximating what Zwislocki estimated to be that of a normal human adult eardrum. The ear simulator also incorporated an ear canal 7.5 mm in diameter, this dimension having been based on the collective data from his own laboratory on a large number of people and on summaries of published data. A mechanical junction was placed in the simulator ear canal 12.7 mm from the microphone at a point corresponding to the depth of insertion of typical earmold in the typical or average ear. The portion of the simulator between this junction and the 1/2" condenser microphone, at the eardrum location is called the *occluded ear simulator* and is of most interest to us in hearing aid measurement work. A drawing of this part is shown in Fig. 15-1. The whole ear canal is simulated in the KEMAR manikin, of course, by a 7.5 mm diameter cylinder, 21.5 mm long, including the occluded ear simulator.

The most critical portions of ear simulator design involve the elements that reproduce the termination impedance corresponding to the eardrum. Of these, the resistance in the acoustic branches has been particularly troublesome. The "felt metal" used in the original design had poor uniformity, having flow resistance variations of 50% to 100% in a single sample. It was very difficult to keep the discs of felt metal flat and simultaneously seal them into the cavities. Flatness was needed because there was a strong interaction between the resistance elements and the tubes producing inertance in three of the four branches. In two of the branches, in particular, the tubes are very short and most of the inertance is contributed by end correction or what we have called "spreading inertance" as the sound spreads out into the volume, after having passed through the small hole. Because the felt metal resistance elements were in the region of sound spreading, slight changes in the location of the discs or in their flatness, made noticeable differences in both the transistion or resonance frequencies and the damping of the branches. The stability of felt metal as a precision acoustic resistance material for these small elements has been questionable.

In the version of Zwislocki simulator presently used in the KEMAR manikin, these uncertainties



(DIMENSIONS IN MILLIMETERS UNLESS OTHERWISE NOTED)
TOLERANCE ON ALL MILLIMETER DIMENSIONS +.02

Fig. 15-1. Zwislocki type ear simulator as modified by Industrial Research Products, Inc.

*This topic was discussed in some detail, but will be omitted here. The reader is referred to the December, 1977 issue of the Journal of the Audio Engineering Society for the complete article, "Measuring the Constants of Ear Simulators".

and instabilities have been eliminated. Changes in the branch construction are best illustrated by Fig. 15-2. While Fig. 15-1 shows a cross-section drawing of the simulator, Fig. 15-2, going from the top to the bottom, shows the actual elements of the individual branches. The low frequency branch is much like its original design, although a single long tube for inertance and resistance has been redesigned with two somewhat shorter tubes, to reduce the total size of the branch. In branches 2 and 3, it has been possible to use the same resistance element. This resistence element comprises a support disc, 0.56 mm thick, with a hole of 2.18 mm diame-

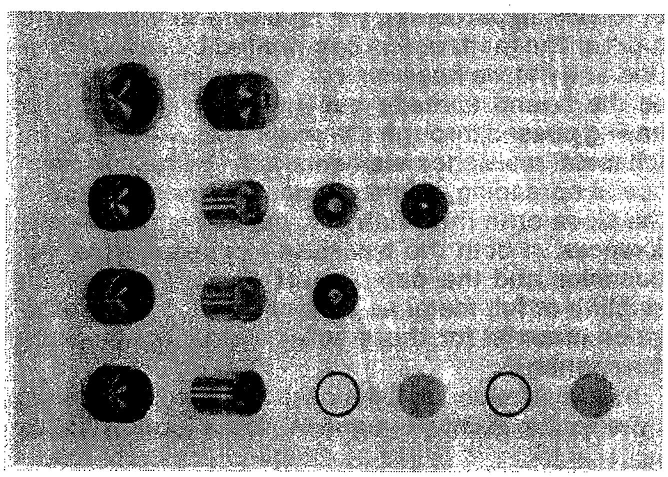


Fig. 15-2. Elements that are used in the four branches of a Zwislocki type ear simulator. Branches one through four are from top to bottom. Elements are inserted into the branch volumes beginning from right to left in the sequence shown.

ter. An electroformed nickel mesh screen with 13.11 wires per mm is cemented to the disc over the hole. The screen is approximately .03 mm thick, and has a light transmission of 15%. The volume of each of these branches and branch 4 are fixed by annular cylinders held in place by a screw-in cap. An inertance correcting disc, with hole diameter of .61 mm and length of .43 mm is also used in branch 2. In branch 4, large area resistance elements are required, because of the larger holes used for inertance. Here, two resistance screens are used and separated by a spacer .22 mm thick and with 5.18 mm diameter opening. The first screen has the wire spacing as above, but it is somewhat more open with 22% light transmission. The second screen is the same material used in branches 2 and 3. Nominal resistance of the elements is 62.5, 62.5, and 28 MPa sec/m^3, respectively for branches 2, 3, and 4. The resistance elements may be fabricated, trimmed, and measured independently and then inserted into the simulator without additional trimming, if the various mechanical operations and dimensions have been properly controlled.

Some of the parameters of the four branch occluded ear simulator have been studied with a computer simulation model as well as measured in the laboratory. Some results of the simulation are

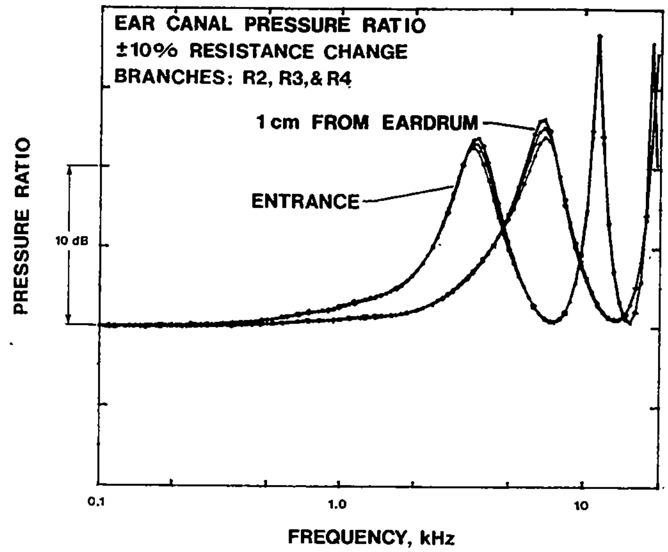


Fig. 15-3. Ear canal pressure ratio for a Zwislocki four branch type ear simulator, calculated from a computer model. The effect of resistance change in the branches is given also.

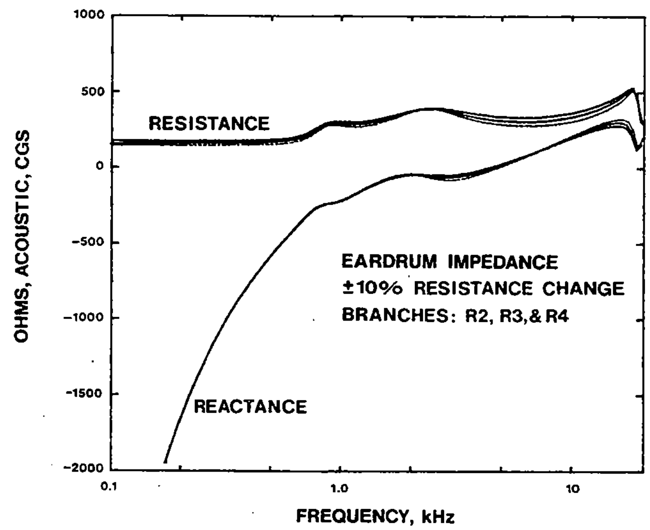
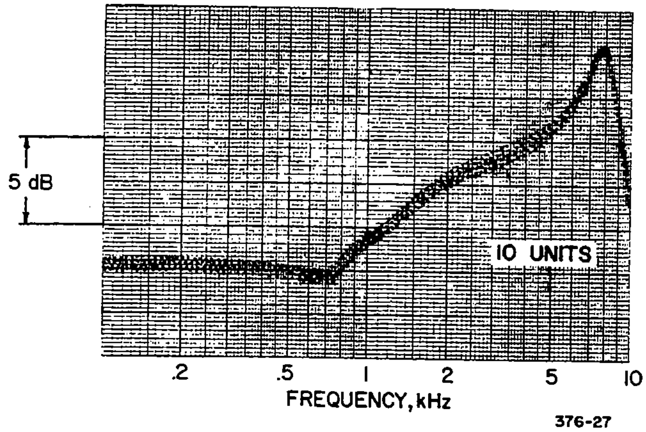


Fig. 15-4. Real and imaginary components of the eardrum impedance of a Zwislocki type four branch ear simulator calculated from a computer model. The effect of resistance changes in the branches is shown.

given in Figs. 15-3 and 15-4. The parameters of interest here are the resistances in each of branches two, three and four and their effect on the pressure ratio in one case and the effective eardrum impedance in the other. In each case a 10% increase or a 10% decrease of resistance change was introduced into all of the branches simultaneously. It can be seen that the effects of these changes although observable probably are not very significant.

A useful method for evaluating these ear simulators is to attach a very high impedance source transducer which produces a constant volume displacement as a function of frequency and observe the output of the microphone at the eardrum location. When this is done a function is plotted which is proportional to the transfer impedance of the occluded ear simulator multiplied by frequency. In Fig. 15-5 we show the reproducibility of 10 Zwislocki type four branch occluded ear simulators which have been measured with an experimental high impedance ceramic source. The range of variation is approximately 1 dB for frequencies up to 10 kHz. The resonance in the vicinity of 8 kHz in this figure is due to the transducer and not the ear simulator.



ZWISLOCKI COUPLER RESPONSE

Fig. 15-5. The response of a high impedance experimental ceramic sound source on 10 Zwislocki type occluded ear simulators.

New Ear Simulator Designs

The Zwislocki design for an ear simulator has been objected to as being unnecessarily complicated. Professor Zwislocki was very concerned about reducing the number of degrees of freedom in the design to the smallest practical number. His experiments with electrical analog circuits led to the four branch structure as being optimum. Gardner and Hawley, (1973) in their work with electrical analogs, found an acceptable fit to the preferred eardrum impedance of Zwislocki with both two branch and four branch systems. For an unexplained reason, one and three branch electrical networks could not be found that would fit the design criteria as well. Zuercher and Burkhard have studied the one, two, three and four branch systems with a digital computer model with conclusions similar to Gardner and Hawley. These results will be discussed below. The various suggestions for ear simulation are not particularly new, but perhaps have been approached with more confidence in view of the more extensive literature describing the external ear that now exists. Recently Diestel (1974) has built a two branch ear simulator, based in part on the Zwislocki and Gardner and Hawley results. Bruel et. al. (1975) have proposed a single branch simulator.

In our laboratory we have experimented with a two branch ear simulator construction with element values selected as a result of our computer model studies and the data on ear canal sound pressures together with the Zwislocki data and the more recent summaries by Shaw (1974).

The acoustical parameters of the external ear canal which were considered in our development were based on our own research as well as on published data. Sachs and Burkhard (1972) measured with a probe microphone the sound pressure at the tip of an earmold for one ear of each of 11 persons. Because the sound source was a very high impedance device, this pressure data could be considered as a close approximation at middle and low frequencies to the impedance of the ear at the point where the probe was located, namely, 5 mm from the tip of the earmold. The results were converted to pressure at the ear drum location by assuming a standing wave between the probe location and the end of the ear canal, and that the estimates of eardrum impedance by Zwislocki were reasonable. This then gives the data shown in Fig. 15-6.

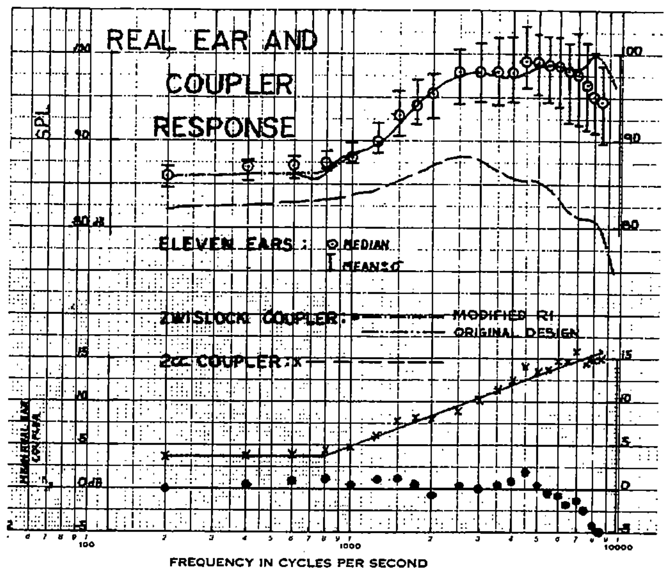


Fig. 15-6. Sound pressure by probe measurement in 11 ears of 11 persons. The figure also shows response of the transducer system on a 2 cm^3 coupler and a Zwislocki type ear simulator, and the differences.

The second important information is the eardrum impedance. Shaw (1974) used the Zwislocki and other data to arrive at an estimate which is given in Fig. 15-7.

The third observation is the ratio of pressure at the ear canal entrance to the pressure at the eardrum of the open canal. The primary source for this type of data has been the work of Wiener and Ross, which Shaw (1974) included in a summary as shown in Fig. 15-8. In this figure we are interested especially in curves A and B. Each represents the average ratio of sound pressure level at an ear canal entrance (A), at a location in the middle of the concha (C) and at a location corresponding to the tip of an earmold (B), to the sound pressure level at the human eardrum. It should be noted that because these curves are averages, peak values for individuals will be greater by 1 to 3 dB. In all cases the ear canal is open.

The various pieces of data on the acoustics of the external ear have varying degrees of accuracy but they must be assumed to represent the actual acoustical fields in an average ear and they must be considered together. As one evolves an ear simulator, one must therefore make a best fit compromise to all of the observed properties. In addition to the various impedance parameters considerable credence was given to the probe measurements in the 11 ears because they did extend over an appreciable portion of the frequency range of interest.

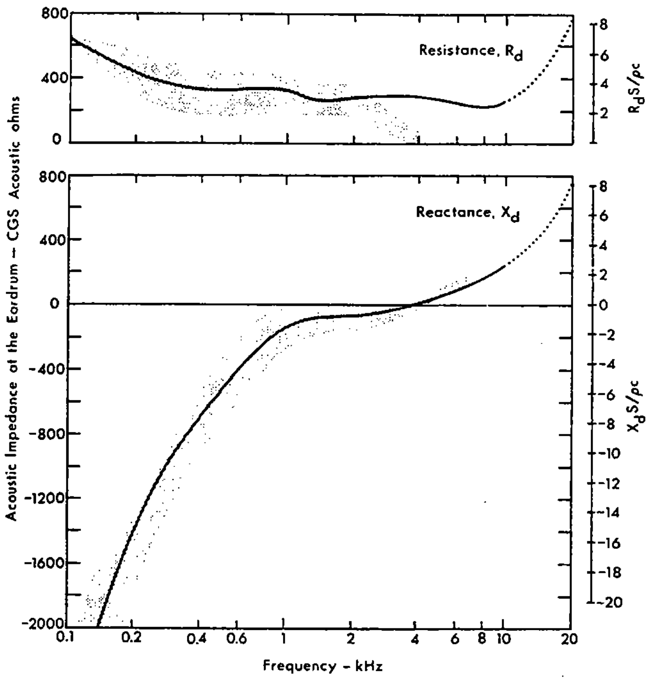


Fig. 15-7. Eardrum impedance design criteria (Shaw, 1974).

The Industrial Research Products ear simulator designated here as XD 1053, will be described first and then some comparisons to other designs will be given. The XD 1053 system is shown in cut-away manner in Fig. 15-9, and a cross-section drawing of its construction is in Fig. 15-10. The two branches are located concentrically with the central ear canal cylinder of the simulator. As in the Zwislocki structure, this ear canal has a diameter of 7.5 mm

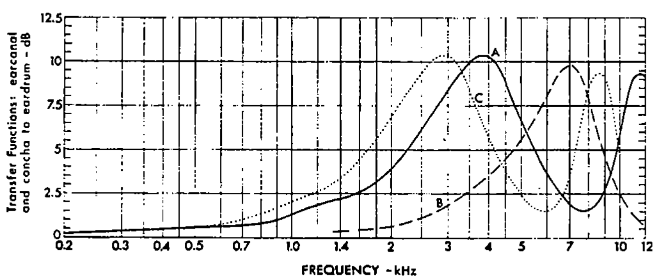


Fig. 15-8. Pressure ratio in ear canal (Shaw, 1974).

and the occluded ear portion extends 12.7 mm from the microphone. It is designed for use with a 1/2" condenser microphone. Resistance and inertance for the low frequency branch are produced by two holes connecting the central volume to the

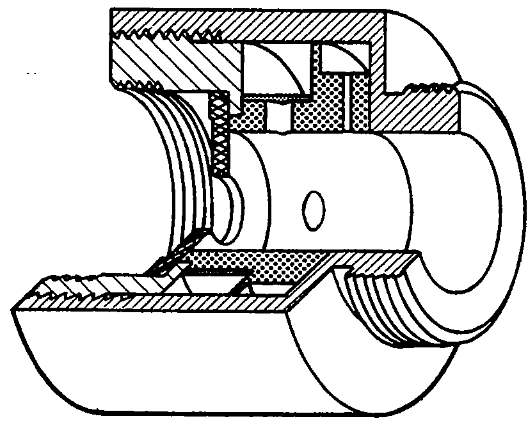


Fig. 15-9. Two branch ear simulator, type XD-1053.

branch volume. Inertance in the high frequency branch is obtained with four holes and most of the resistance is obtained by three layers of polyester monofilament bolting cloth. This bolting cloth is commonly used for fine particle filter applications.

Before deciding to locate the branches along the length of the canal of the ear simulator, and away from the microphone, the effect of branch location was evaluated with the computer model. It was found that the low frequency branch could be placed in the occluded ear simulator any distance from the microphone with impunity, but that some care must be given to the location of the high frequency branch. Branch locations satisfy the criteria derived in the analysis.

The uniformity of response of an earphone on six XD 1053 ear simulators was very good, as shown by Fig. 15-11. This earphone is an experimental ceramic unit that produces a volume displacement which is nearly constant with frequency, when driven with a constant voltage, over the frequency

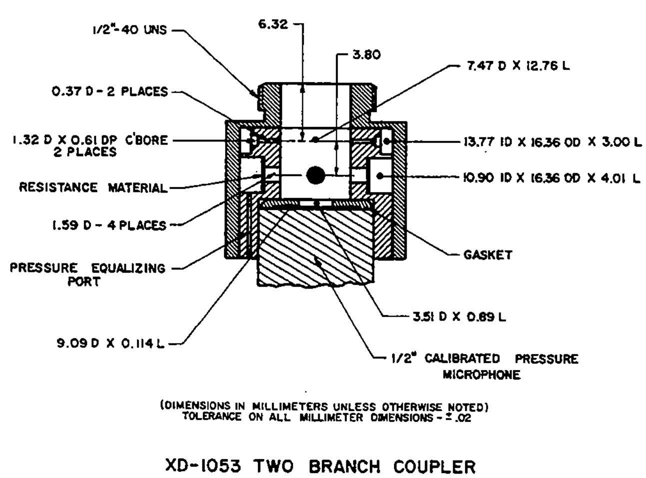


Fig. 15-10. Drawing of XD-1053 two branch occluded ear simulator.

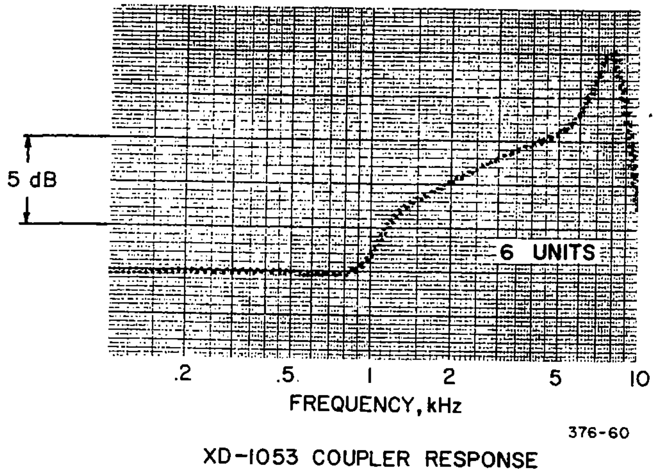


Fig. 15-11. Response of experimental ceramic earphone on XD-1053 occluded ear simulator, (6) six units.

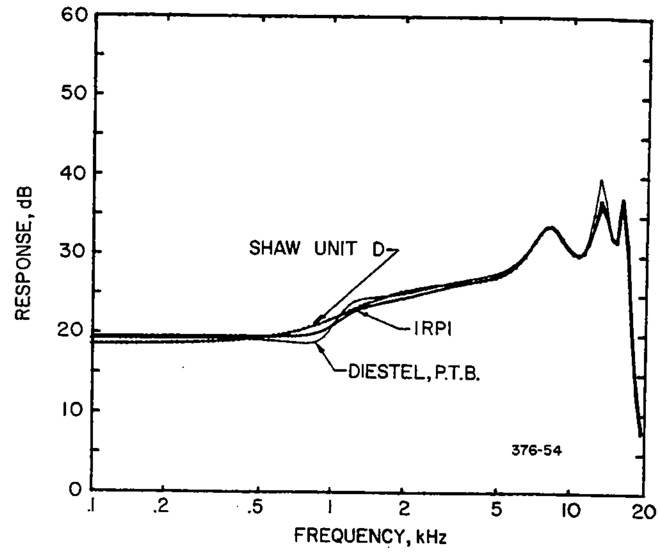


Fig. 15-13. Two branch occluded ear simulators: transfer impedance (multiplied by frequency) for a high impedance ceramic source. Computer calculated curves.

range up to 6 kHz. It has a highly damped resonance at around 8 kHz and is the same source used to obtain data in Fig. 15-5.

Some results of the computer modelling of the Shaw model, the simulator designed by Diestel, and the XD 1053 simulator are given in Figs. 15-12, 15-13, 15-14, 15-15, and 15-16. (More extensive analysis of these simulators, the Zwislocki type and a one branch system are given in Report 20022-2, to Knowles Electronics (April 1976), by Zuercher and Burkhard, Industrial Research Products, which was in preparation at the time of the conferences.) In general the differences among these designs are most pronounced in the frequency region from .5 to 3 kHz where the branches become resonant and undergo decoupling from the central cavity as frequency increases. The magnitude of input impedance is shown for the three occluded ear simulators in Fig. 15-12. The calculated response for the experimental ceramic source transducer is shown

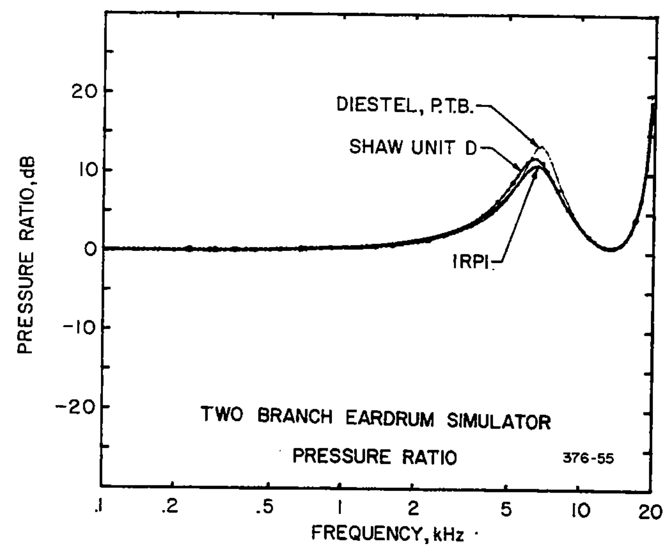


Fig. 15-14. Pressure ratio in three models of two branch ear simulators for pressures at the eardrum and at the location to be occupied by a standard earmold tip. Ear canal is open. Computer drawn curves.

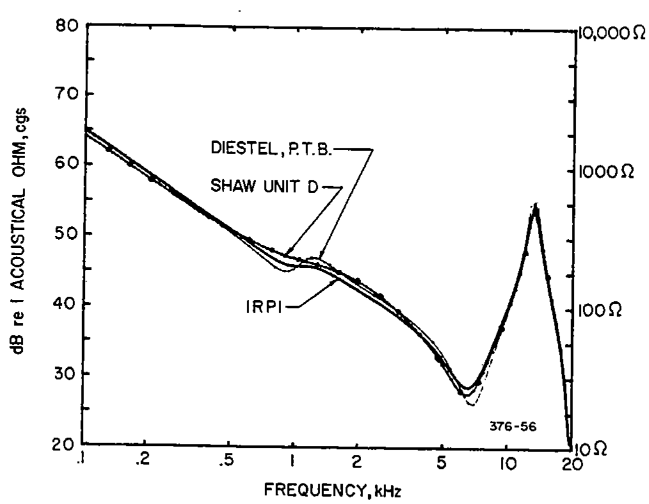


Fig. 15-12. Magnitude of input impedance for 3, two branch occluded ear simulator designs, calculated by computer model.

in Fig. 15-13 and may be compared with the data in Fig. 15-10 and Fig. 15-5. Resonance in the vicinity of 17 kHz is due to an anomalous behavior of the transducer and is not related to the ear simulator. There are also some differences in the damping of the pressure ratio maximum, Fig. 15-14, which are directly traceable to differences in the way eardrum impedance is synthesized, Fig. 15-15. The details of complex input impedance of these occluded ear simulators are shown in Fig. 15-16. The components are more alike, over much of the frequency range, because the ear canal volume is an appreciable part of the total impedance and at frequencies below the half wave length resonance, acts as a shunt on the eardrum impedance.

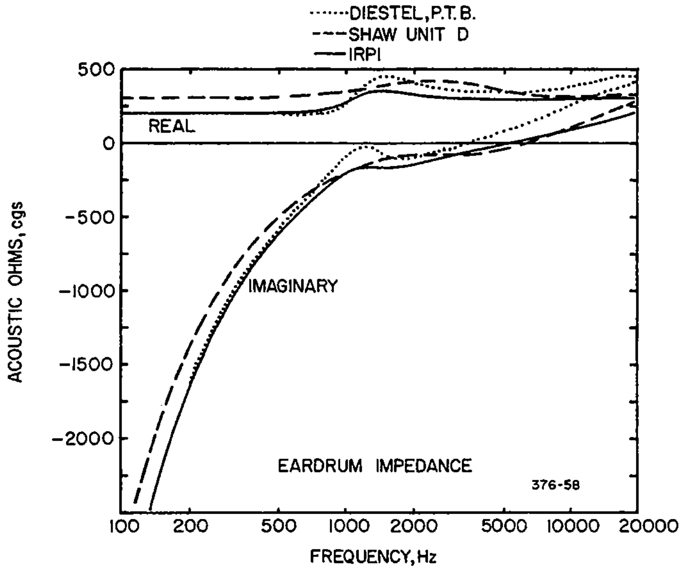


Fig. 15-15. Eardrum impedance for three different occluded ear simulator designs.

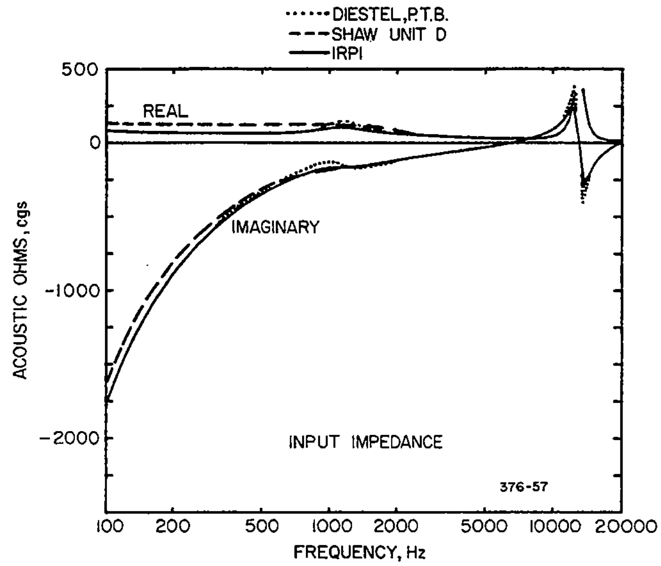


Fig. 15-16. Input impedance for occluded ear simulators of three different designs.

References:

Bruel, P.V., et. al. (1975) Impedance of Real and Artificial Ears. Bruel & Kjaer report.

Diestel, H.G. (1974) Measurements on New Couplers for Insert Earphones. Eighth International Congress on Acoustics. *1*, 217.

Gardner, M.B. and Hawley, M.S. (1973) Comparison of Network and Real Ear Characteristics of the External Ear. J. Aud. Eng. Soc. *21*, 158-165, April.

Sachs, R.M. and Burkhard, M.D. (1972) Earphone Pressure Response in Ears and Couplers. Report No. 20021-2 to Knowles Electronics, Inc. Also presented at 83rd Meeting of the Acoustical Society of America, 21 April 1972.

Shaw, E.A.G. (1974) The External Ear, a chapter in *Handbook of Sensory Physiology* Volume V/I edited by W.D. Keidel and W.D. Neff. (Springer-Verlag)

Shaw, E.A.G. (1975) The External Ear: New Knowledge. *Earmolds and Associated Problems,* S.C. Dalsgaard organizer. Seventh Danavox Symposium, G.L. Avernaes, Denmark, pp. 24-50.

Zwislocki, J.J. (1971) An Earlike Coupler for Earphone Calibration, LSC-S-9, Laboratory for Sensory Communication, Syracuse Univ., Syracuse, N.Y.

Chapter 16.

Helpful Accessories

M.D. Burkhard
Industrial Research Products, Inc.

From time to time, it has been found that some of the measurements on the manikin and on the ear simulators when removed from the manikin, can be aided with attachments or accessories. The more useful ones will be noted briefly here.

Lybarger suggested a machined connector cylinder that could be inserted into the ear canal of the manikin as a substitute for an earmold. The substitute adapter is shown in Fig. 16-1. It may be made with a hole diameter corresponding to the tubing

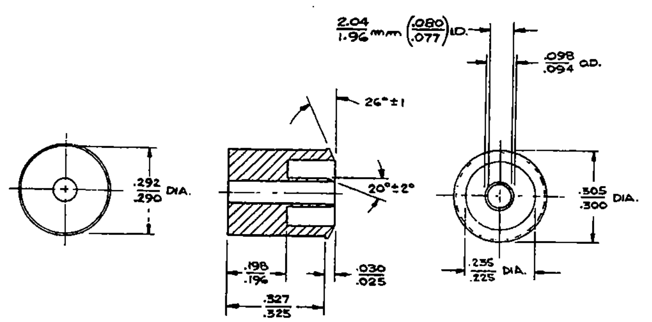
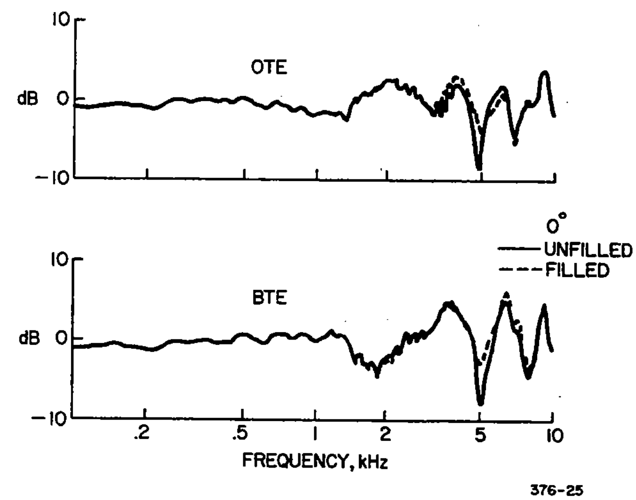


Fig. 16-1. Earmold substitute or adapter to be used in the KEMAR ear canal. (Dimensions in inches.)

being used. The piece extends into the manikin ear canal to the junction between the occluded ear simulator and the outer portion of the ear canal, thus giving a residual ear volume equivalent to that used when the hearing aid might be measured on the laboratory bench. To install the adapter, the rubber external ear of the manikin is removed, the substitute adapter inserted into the canal, and then the ear is replaced. The substitute adapter is trapped between the rubber ear and the metal of the ear canal by the small lip. It is recommended that after the piece is inserted, the concha of the ear be filled with putty or an ear mold. The resonances in the concha influence the sound field in the vicinity of the ear where hearing aid microphones may be placed, as shown in Fig. 16-2, and would not usually be present. The effect of filling the concha to eliminate its resonances is not large, but never-the-less may be preferred.

Examples of fittings to use with the occluded ear simulators, when they are removed from the manikin, are shown in Figs. 16-3, 16-4 and 16-5. The fitting in Fig. 16-3 provides a tubing connection. The hole would be made to correspond to the tubing on the hearing aid. For an in-the-ear hearing aid or an earmold, adapters like the one shown in Fig. 16-4 might be used. The tip of the earmold is sealed to a shallow conical adapter with wax so that there is negligible additional length added to the occluded ear simulator. This adapter is shown with a Zwislocki type occluded ear simulator.



OVER THE EAR AND BEHIND THE EAR SOUND PRESSURE-FILLED AND UNFILLED CONCHA

Fig. 16-2. Pressure on the head near the ear showing the effect of filling the concha, for a constant free field sound environment. OTE and BTE refer to typical microphone locations in ear mounted hearing aids.

For simulation of the 18 mm long by 3mm diameter earmold tube specified in ANSI S3.7 and IEC 118 and 126 standards, a device like the one shown in Fig. 16-5 could be used. This particular adapter provides a rubber grommet at the top to retain and seal the nub of a typical insert receiver or earphone. The adapter is shown here on an XD-1053 two branch ear simulator described in Chapter 15.

The open ear fitting such as in a CROS type hearing aid, appears to require a means of holding the sound tube in the ear canal in a fixed and controlled manner. Dr. Causey reports that the Veterans Administration uses a skeleton ear mold structure shown in Fig. 16-6. The tubing is retained in one part of the mold. It is essentially open to the ear canal and has a structure that allows it to be trapped around the perimeter of the concha and in the cymba region of the ear. This has proved useful in measurements on Veterans Administration supplied hearing aids. This is one of a variety of open ear molds that might be considered for standards with the KEMAR manikin. Other examples were shown by Birk Nielsen (1975).

The usage of body hearing aids has been steadily declining. However some provision may still be needed for testing of body hearing aids on the manikin. One suggestion would be to use a standard body pouch that has straps over the shoulder and holds the hearing aid in the center of the upper torso. Vertical location of the hearing aid would be measured down from the neck junction between the torso and the head on the manikin.

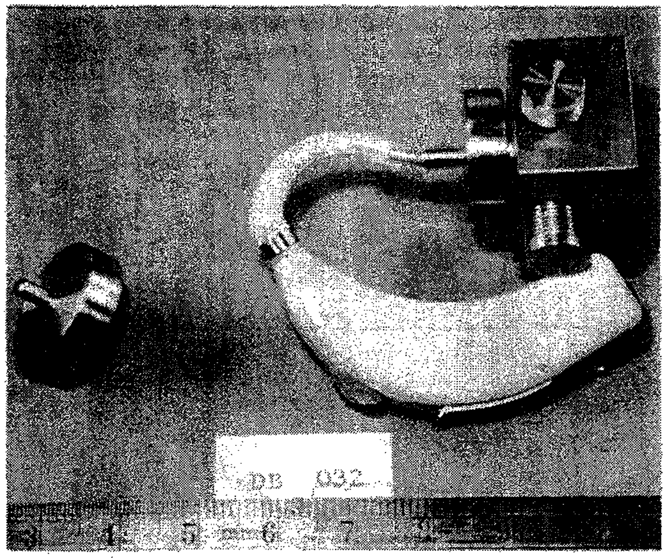


Fig. 16-3. Adapter for occluded ear simulator to connect to hearing aid tubing. The hole in the adapter would be equal to the inner-diameter of the tubing. The ear simulator shown is a Zwislocki four branch type.

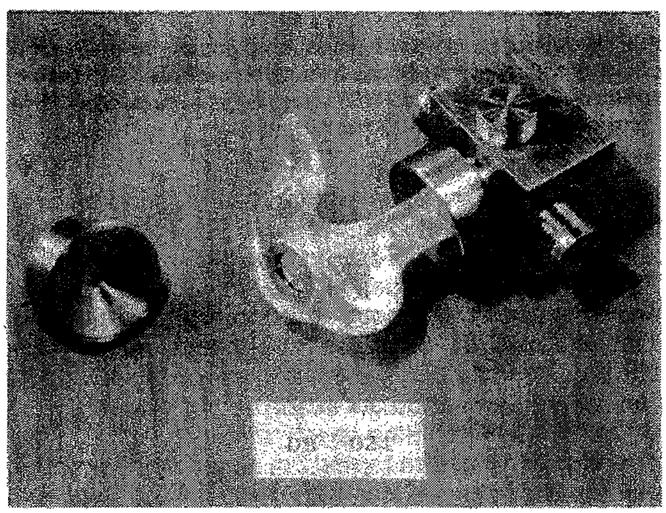


Fig. 16-4. Fitting for attaching an earmold or an in-the-ear hearing aid to an occluded ear simulator. The ear simulator shown is a Zwislocki four branch type.

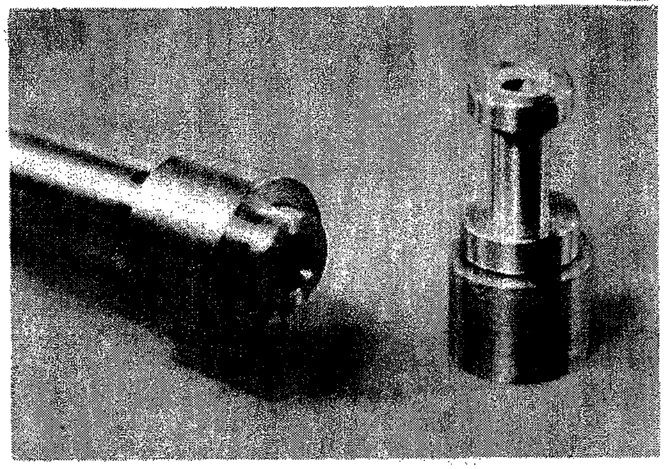


Fig. 16-5. An adapter for simulation of an earmold tube having internal dimensions of 18 mm length by 3 mm diameter. The adapter is shown with an XD-1053 two branch ear simulator.

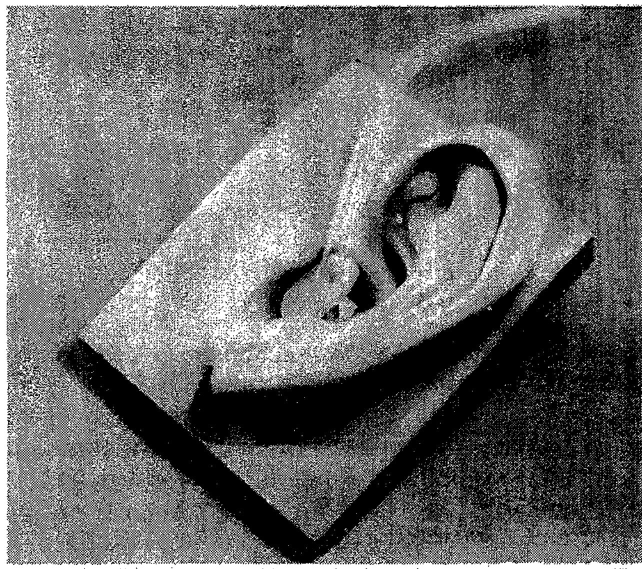


Fig. 16-6. Open ear mold used by the Veterans Administration.

References:

Birk Nielsen, H. (1975) The Effects of Open Versus Closed Earmold Treatment. *Earmolds and Associated Problems* organized by S.C. Dalsgaard. Seventh Danavox Symposium, Gl. Avernaes, Denmark.

Chapter 17.

Considerations For An *In Situ* Hearing Aid Measurement Standard

M.D. Burkhard
Industrial Research Products, Inc.

G.D. Causey
University of Maryland
and the
Veterans Administration

INTRODUCTION

Causey and Burkhard prepared an outline of requirements for manikin based measurements of hearing aid characteristics for Working Group S3-48 of the American National Standards Institute. A recently revised version is given here. A number of factors still remain to be decided. This list reflects many concerns one should have as he does various acoustic measurements with a manikin, and is especially directed to hearing aid evaluation.

PROPOSAL
REQUIREMENTS AND PROCDURES FOR MAKING HEARING AID MEASUREMENTS WITH A MANIKIN

PRELIMINARY OUTLINE

1. The uniformity of the sound field will be measured. A suggestion now is that the eight corners of a cube centered on the head, sans head, will have a specified maximum range of sound pressure, front to back, side to side, and up-down. Diagonal lengths of this cubic space would be 20 cm.
2. The measurement conditions will be specified relative to free field at the location in the sound field that will be the center of the head. The center of the head will be defined as being the center of a line between the ears.
3. The distance to the source will be 1 meter.
4. The source size will be restricted tentatively to less than 8" diameter. It will be a single simple loudspeaker source, i.e., no multiple loudspeaker or vented boxes (unless the vent is concentric with the speaker).
5. Horizontal for the manikin has been defined (see Chapter 2, p. 6). The effective center of the source will be 6.4 cm below this horizontal on the manikin, with the manikin centered on the source axis in all other aspects.
6. In reviewing the requirements here, it was decided that reference marks should be placed on the manikin.
7. Some attention to room size and/or possible definition of absorption in the room may be necessary. This must be evaluated.
8. The manikin KEMAR is supplied with two mounting positions. We will specify the recommended mounting position as the most forward one.
9. Clothing will be used on the manikin. The suggested clothing will be a T-shirt and a lab coat or equivalent.
10. For head-worn hearing aids using acoustic tubing, the hearing aid will be mounted on the manikin as it is expected to be worn on a person. Then the tube will be cut to a length that connects the hearing aid directly to the earmold in a conventional manner. This is a departure from ANSI standard practice permitting arbitrary specification of tube length. The tube length should be stated.
11. For open-ear and vented-earmold calibrations, the measurement should be made with gain depressed * dB relative to the acoustic feedback gain level, if observable (*number to be verified by experiment).
12. For open-earmold calibrations, the depth of insertion of the tube will be specified. Tentatively, it appears that this should be at the entrance of the ear as defined by the junction between the rubber-molded ear and the metal fitting to which it connects.
13. For body aids, a harness will be used which will support the hearing aid in the center of the front, ____ cm below the neck junction of the manikin (distance to be established).
14. There will be a note added about precautions in using control microphone signals with KEMAR, probably in connection with specification of the room, distance to source, etc.
15. Coordinates for directional measurements would be a clockwise rotation, with 90° indicating that the active ear is closest to the sound source.
16. Provision is to be made, by means of standard corrections, for converting non-manikin

data to manikin equivalents. This could permit reporting of data obtained in sound boxes to be compared with data obtained on manikins with the increased tolerance or increased uncertainty pointed out.

April 1974
Revised January 1976

Chapter 18
Discussion From The Manikin Measurements Conferences

This chapter contains part of the discussion which took place in each of the two conferences. The proceedings were recorded and, to the extent that they could be transcribed faithfully, material is nearly verbatim.

Zurich, Switzerland
March 4, 1976

Anonymous: You have made two rings to be placed in the neck of the KEMAR manikin. What neck length or how many rings are to be used for a measurement?

Burkhard: Our recommendation is that only one of the rings be used. This would then produce a median distance, based on all male and female persons, between the shoulder and the ear. Variations in this distance, as was pointed out, produce an observable effect on the pressure at the ear and in the eardrum of the manikin.

Anonymous: Certainly we know that the differences in the practical use of a hearing aid are much greater than any error we can introduce by measurement. I am talking about difference in fittings due to the individual differences you see in length of tubing or any other variable. But I think we should have a standard body or size of manikin that can be used as a female as well as a male so that we do not have to reproduce frequency response curves for fittings on men and for fittings on women individually.

Burkhard: I would like to comment on the length of tubing used for hearing aids on the manikin. The tube lengths specified in standards are too long for most hearing aids to fit in a normal position on the KEMAR manikin. It will be noted in the summary of test considerations for manikin measurements of hearing aids included in the conference (Chapter 17) that Causey and I recommend that the tube be cut to fit between the hearing aid as it is normally placed and the ear canal. With this definition there will be variations in the tube lengths for different hearing aids. This will, of course, influence the frequency responses. This, it seems to me, is in fact what happens when a hearing aid is actually used.

Helle: When the hearing aid is actually used it is on individuals. But if one is making a measurement it is important to be able to compare measurements of say different manufacturers. Therefore, I think we should have standard conditions for the measurements of the hearing aids. If we have a manikin where we all know the size, I think we should be able to provide a standard length of tubing that can be used with different types of hearing aids. There may be a set of lengths for eye-glass hearing aids and another one for behind-the-ear hearing aids, but I think for behind-the-ear hearing aids we should try to have one standard length.

Burkhard: Dr. Helle, in one of your slides you indicated that a coupler on the unused ear would provide a control signal for measuring the gain of the hearing aids on the other side. Do you use equalizing or change the spectrum in any way?

Helle: We have not tried this method. The method should be possible. There should be problems in the region around 8 kHz because, as you have shown, there are some differences at that frequency between the two ears.

Bergenstoff: Dr. Helle, how do you fit the hearing aid onto the KEMAR manikin?

Helle: We had an insertion of 18 mm length. It was like the insert for the 2 cm^3 coupler having an inner diameter of 3 mm. The adapter ended at the same point in the coupler as the new adapter Mr. Burkhard has shown here. The concha was not filled with an earmold. From the adapter we used 2 mm of plastic tubing.

Bruel: I have two small remarks. First, I think that using a tape recorder for the corrections is an excellent method. We have used it many times with good success. The problem of the wearing of the tape we have overcome by using an FM tape recorder. This solves such problems without any difficulty.

Burkhard: The FM recording method would be very similar to digital storage of the signal that I mentioned. One thing we found in regard to stability is that if we turn on the amplifier and apply power to the loudspeaker for half an hour or so, the loudspeaker will change its response because of heat from the coil.

Anonymous: Mr. Burkhard, do the polar response measurements that you showed with a hearing aid in place rotate with the hearing aid or the manikin at the center?

Burkhard: All of my data uses the center axis of the KEMAR for display of polar response of hearing aids. There has been some data reported by other people in which the axis of rotation was at the ear or at the hearing aid. I feel strongly that the axis for all polar response data of hearing aids *in situ* should be on the center of the manikin.

Helle: In our measurements comparing regulation with a microphone in the vicinity of KEMAR with the KEMAR substituted in a regulated field, we find differences of about 1.5 dB. I think in our investigation the interference of the control microphone with the sound field may not have been as critical as you suggested in your data.

Burkhard: The effectiveness of control by this method undoubtedly depends on the size of the room and how much space is available to do the experiment.

Helle: Yes, the room size is important and we know that this point of how the control is made is a critical one. Our room is somewhat larger than yours.

Helle: I am not sure that it is possible to have correction curves for different types of hearing aids measured in a sound box. The location of the microphone on the head and the influence of direction will certainly cause differences.

Burkhard: I would agree that correction curves must be used intelligently and that the accuracy of approaches involving correction curves will be less than if direct measurements are made. One should develop a correction curve for the particular hearing aid microphone system that is under investigation or development. The idea here was to suggest, with the correction curves applied to sound box measurements, that it would provide an inexpensive way of obtaining data for persons who did not have adequate facilities.

Helle: I think you must have a KEMAR manikin to do the measurements.

Burkhard: The reasons we have concerned ourselves with this question were two. At a meeting with some hearing aid manufacturers in Washington, D.C., November, 1975, the question was asked: How can we measure an in-the-ear hearing aid on KEMAR? As I pointed out then, KEMAR has a very specific ear. It can be thought of as one person's ear. A custom-made hearing aid or earmold that will fit one ear will not fit another ear in general. We felt that perhaps this box pressure control method provided a way of obtaining useful measurements of in-the-ear hearing aids without expensive rework and modification of the KEMAR manikin for each earmold. The other reason, of course, I have already stated as being an economical alternative in those situations where a KEMAR and the associated sound room could not be provided.

Rasmussen: In studying the question of the pressure on a microphone of an actual hearing aid, you used one microphone getting closer and closer to the hearing aid and certainly you get more equivalent curves. But the whole structure will have a different diffraction pattern. We have found 5 or 6 dB change of pressure due to the input impedance of hearing aids when they are placed in a specific location.

Burkhard: That really causes no problem. We know a standard condenser microphone responds to the pressure at its diaphragm, and that the diffraction effects of the microphone are added to its calibration to account for the free field and hence for the fact that the pressure at the diaphragm of the microphone is influenced by the total structure. But if the microphone is calibrated for its pressure response as in a calibrating coupler, and is used next to the hearing aid microphone, we are in fact controlling the pressure at the entrance to the hearing aid because they both are at the same point in the sound field. As far as the impedance of the hearing aid microphone is concerned, that really makes no difference, because the calibrating microphone senses the pressure at that point. Any loading of the sound field due to the finite impedance of the hearing aid microphone will be the same when the hearing aid is mounted on a person or is adjacent to the calibrated microphone. The important consideration here is that by making the measurement this way, all the diffraction effects are eliminated, both for the hearing aid and for the sensing microphone, and that the pressure is indeed the controlled parameter. Modern hearing aid microphones do have very high input impedance and should not affect the sound level appreciably. Of course, as I just implied, the fact that the impedance of the hearing aid microphone may be finite really does not cause any problems. The hearing aid response curves (Fig. 12-6) that were obtained by the two methods of measurement show good agreement, one being the comparison in the box and the other being the in situ measurement on the manikin. This data tells us that the impedance problems are not important.

Lauridsen: Mr. Burkhard, in the example you showed for controlling and measuring the pressure at the entrance at the hearing aid microphone, you showed the microphone on the outside of the hearing aid, very close to the compressor microphone. How can you do the measurement when the microphone is on the underside of the hearing aid next to the hook?

Burkhard: We have not made the observation with a hearing aid of that type but I see no reason why the microphone could not be placed adjacent to it. The microphone senses to a degree the average pressure over its diaphragm. There may be some uncertainty. If one is really concerned, he could use a 1/2" microphone to reduce the size of the cavity created in such a measurement situation.

Anonymous: I would say according to our experience it depends to a large degree on the shape of the hearing aid and the location of the microphone sound port.

Helle: We found, by practical measurements when we compared measurements in a sound box and then the anechoic room, that the position of the hearing aid relative to the pressure microphone depends to a large degree on the type of hearing aid. You can't say that one position is the best for measurement in a box. It might be best for one particular type of hearing aid but it might be poor for another one. We found that in some cases, and I have no explanation for it, that correlation of measurement in the box and in an anechoic room is much better if you put the hearing aid and the pressure microphone a few centimeters apart than if you put them closer together.

Bruel: In the test box you are trying to simulate free field conditions with the diffraction of the hearing aid included and that, of course, will work if the microphone is placed a distance away from the hearing aid. In the case Mr. Burkhard discussed, one really wants to have the pressure characteristic. What Rasmussen was discussing is that its not so easy to define the opening of the hearing aid because, first of all, you have sound going all over the place through the case and, therefore, you may not always have a high impedance device. There will be some impedance in the vicinity. In any case, the resulting measurement will be a pressure. Then by other means, as was explained by Mr. Burkhard, one would compensate for the diffraction and then you will be in business.

Lauridsen: Where in the system should the corrections be applied?

Burkhard: This is a matter of philosophy that hearing aid manufacturers and hearing aid users have to answer. It is obvious from what I have presented that the sound stimulus to the hearing aid will vary with frequency when it is worn in a constant free sound field. This is different from the concept used previously for hearing aid measurements and I think we have to relook at what we want to measure and how we want to apply sound to the hearing aid; and whether we, in fact, should include these diffraction effects before we make, for example, a distortion measurement. I am not prepared to answer what is good and what is bad at this time. Discussions at these conferences will hopefully develop a best procedure.

Helle: I think that an important point is to state what the condition was for the measurement to get a functional gain or an etymotic gain, and that the user should really see what measuring method was used. If you had the sound pressure level constant as a function of frequency or if you corrected before or after the coupler.

Nielson: You were talking about comparing the sound pressure at the two ears of the KEMAR manikin. Then you talked about differences. Have you deliberately made KEMAR unsymmetrical?

Burkhard: No, we did not do this deliberately. In fact, we tried to make the manikin symmetrical but at high frequencies it is very difficult to make things exactly alike on the two sides. Up to this time we have not found a shop that would make a mirror image mold of one side from the other at an acceptable price.

Dalsgaard: I know that an organization in Denmark has developed a technique for producing the mirror image of an ear for making prosthesis for people who have lost one ear. They use a special machine. I can probably give you a reference.

Dalsgaard: I put a question of where body hearing aids are mounted to some hearing therapists in Denmark. They told me that 90% of the males wear the pocket aid inside the left breast pocket, but the women wear the body hearing aids in the center of the body. This suggests that we should work with two body aid wearing positions.

Helle: You have shown an adapter, analagous to the HA-2 earmold simulator. (Fig. 16-5) Do you always use a 3 mm diameter hole?

Burkhard: At this time we have taken that connection because it has been used as a standard for the IEC and the ANSI 2 cm^3 couplers.

Helle: Some people are arguing that for hearing aids to be used behind-the-ear, the tubing is somewhat different and therefore they are saying that it would be better to use 2 mm tubing, not 3 mm tubing, for measurement.

Burkhard: I think a standard will depend on what the need is, what the practices are in the fitting of hearing aids. The earmold adapter part was quite arbitrarily machined in with a 3 mm hole because it copied existing standards. Obviously it can be made with any size hole.

Helle: I see two problems. One is you have to come close to the tube diameters actually used and the second is you have to have the same value for all people who are measuring the hearing aid.

Bergenstoff: Concerning the earmold substitute, I recently made some analysis on actual earmolds made by one of the Danish State clinics, and it is true that in earmolds meant for insert receivers the inside diameter very seldom exceeds 2.4 mm. On the other hand, the length seems to be much longer than 18 mm. In any future standards this has to be considered.

Dalsgaard: I think there has been some change in the earmold production. We made some statistical analysis covering 9 Danish hearing centers and a total of 900 earmolds. We found an average length of 22 mm with a standard deviation of .3 mm. We also found an average diameter of 3 mm at that time. It seems though that we can decrease the diameter.

Burkhard: I have asked some U.S. manufacturers what the practice seems to be in the clinics there and they say that diameter tends to be smaller than 3 mm now.

Helle: Mr. Burkhard, in your two-branch coupler design does the damping material, the monofilament cloth, go all the way around?

Burkhard: Yes, there are three revolutions and hence three layers of the material covering each of the four holes for the high frequency branch.

Lauridsen: Is this monofilament damping material stable and how would it compare to felt metal?

Burkhard: As far as we know the monofilament material is much more stable than felt metal. The polyester monofilament material does not absorb moisture and it is very stable chemically. The polyester is essentially Mylar, and Mylar is used for many purposes. The metal screen used as a replacement for the felt metal in the Zwislocki-type coupler is still better. The screen that I described is nickel so that it is very inert. The metal screen probably could be used in the two-branch coupler, if a satisfactory way could be found to form it and cement it around the holes.

Lauridsen: Did you say the basic screen parts in the four-branch coupler are electro-formed?

Burkhard: The metal screens are electro-formed by electrolytic deposition. They start with art work the same as integrated circuit manufacturers have art work for the patterns they put on silicon. A conducting wire pattern is placed on a substrate by photolithography. Metal is then deposited by electrolysis, i.e., by electrolytic deposition, the thickness of which can be controlled very precisely. The properties of the material are described in terms of its light transmission, and it is typically identified by the percentage of light transmission. We have measured samples of the material by both looking at it with a microscope to observe the wire spacing or grid size and by measuring the acoustic resistance with appropriate equipment. It is surprisingly uniform. The little discs with the metal screen on them can be made with acoustic resistance variations of 1% or 2%, except for occasional problems when cement will wick or run into the open grid area. The metal screen seems to be a very desirable material.

Anonymous: Mr. Burkhard, the branches of your new coupler are no longer near the microphone.

Burkhard: Yes, that is partially correct. We have placed the one that is most critical in location closest to the microphone and one that is not critical further away. As background, we did an extensive analysis of the Zwislocki coupler design to determine what economies could be made in its manufacture. One of the parameters investigated was the location of the branches along the ear canal. The location of each resonator was independently varied by computer analysis from a point adjacent to the microphone to a point at the remotest possible location, a distance of about 12 mm from the microphone. The only branch whose location affected either the transfer impedance or the standing wave pressure ratio in the ear was branch number 4. Also, the eardrum impedance, as deduced from a measurement of impedance at the entrance to the ear canal, was affected only by the location of branch number 4. Branch number 4 is the high frequency resonator and has large holes between the ear canal and the branch volume. The reason for this is that at low frequencies the pressure is essentially the same phase throughout the whole region of the occluded ear simulator. But at high frequencies the wave length is short enough that there will be relative differences in pressure from one end of the occluded ear simulator to the other and, hence, an impedance at a particular location can influence the net standing wave properties there. Using that background we looked carefully at the effect of branch locations in the two-branch coupler and found similarly that only the high frequency branch needs to be adjacent to the microphone.

Anonymous: What is the basis for the design of ear simulators?

Burkhard: First, virtually all of the data that is used in the design of ear simulators is based on measurements on normal persons. Most of our selection of design parameters is based on the analysis provided by Dr. E.A.G. Shaw from the National Research Council in Canada. He has made a comprehensive summary of data on the acoustics of the ear. This work is published in the *Handbook of Sensory Physiology* edited by Keidel and Neff published by Springer-Verlag in 1974. (See references to Chapter 15.) We were influenced considerably by the parameters and dimensions given by Zwislocki in his original design. We gave the most weight to our own observations on one ear of each of 11 persons, in which we measured the sound pressure 5 mm from the end of an earmold in an occluded ear.

Lauridsen: To what frequency limits do you consider that these ear simulator designs will work?

Burkhard: I see no reason why they should not work over the frequency range up to 15 or 16 kHz. Unfortunately, we have great difficulty in measuring input impedance and transfer characteristics of ears at these high frequency limits and, hence, to know exactly what performance we should reproduce in the ear simulators. I would like to ask you how high a frequency would the device be used for.

Helle: There we come to a problem. The coupler, when it is a real ear simulator, will be used for other purposes than for hearing aids.

Burkhard: Yes, I am well aware of that. Dr. Diestel and I have discussed this in connection with measurement of other types of earphones. We have been addressing ourselves here to the occluded ear simulator portion which is needed for hearing aid work. When the whole ear is simulated such as

Zwislocki attempted to do, you have additional acoustic elements that contribute to the response of an earphone for example, or to the response of the ear as an open ear receiver of sound. The structure is larger so that dimensional characteristics that give rise to standing waves or resonance frequencies obviously will be important. At the mid-low frequency region, however, the impedance that we introduce in the occluded ear simulator portion will be relatively less important because of the large volumes connecting between it and the earphone source.

Nielsen: It seems to me that we are approaching a difficult situation. Today you have described one more coupler or ear simulator. I think it is time that we try to put pressure on people sitting on the standardization committees to speed up the job before we have more good proposals. I think it is very important to the hearing aid manufacturer. We have seen our products being judged by different couplers and this creates confusion.

Grant-Salmon: You have to establish an area of agreement, perhaps, between manufacturers. The question is where is the agreement to start? Who should agree first?

Nielsen: We have seen here the latest versions, for instance, of a Zwislocki-type coupler. It has been very much improved as to reproducibility and how it would change with time now that Mr. Burkhard has put these new resistance metal elements into them. The other couplers are all good couplers too, but each design has a little different characteristic. Maybe it is time to decide that we should use one of them.

Grant-Salmon: My point is who would decide that.

Nielsen: The committee.

Burkhard: There is at least one person who is a member of the appropriate coupler committee here. All of us have some information for these committees and it is necessary to talk to the members of the committees to see that they get our viewpoints. Standardization is necessary.

Dalsgaard: I regret very much that Dr. Johannson, who is chairman of WG6 Hearing Aids, is not here today and also that Dr. Diestel of the other WG6 on Insert Earphone Couplers is not here too. I think we have to impress on these two chairmen the need for speeding up the work. The developments are running past us. It is essential that there be close collaboration between the two working groups.

Burkhard: In the U.S., an activity paralleling the IEC has the objective of specifying a standard ear simulator. The approach at this time is to create a performance specification for such devices. Then a description of one or two pieces of construction that meet the specification would be given. This turns out to be a very difficult task. The performance parameters that seem to be reasonable are the transfer admittance or transfer impedance, and perhaps an additional parameter that will describe the damping in the system such as the standing wave ratio. Another parameter as used in the design of these devices is the terminating impedance which I will call the effective terminating impedance. These three quantities would force any coupler that might be devised into a very narrow performance range.

Dalsgaard: I think that is the philosophy of WG6, Couplers.

Burkhard: If the specification is given then there must be a procedure by which manufacturers and users can determine whether the devices comply with the standard.

Dalsgaard: This was the method in the artificial audiometric ear. Was it not true, Dr. Bruel, that you decided first on the impedance data and then you constructed the coupler?

Bruel: The only specification is impedance data but it is expressed in volume, resistance and mass. How it's actually made is up to the manufacturer. By the way, I would like to see data for earphones on the IEC artificial ear in the comparison among the 9-A coupler, the KEMAR manikin and the Zwislocki coupler.

Helle: Do you think the result will be to standardize performance rather than dimensions? Is that not dangerous?

Dalsgaard: It is dangerous but I think we have to do it. We were exposed to this method with the audiometric ear and the approach was further confirmed in the development of the artificial mastoid. We were, in fact, in the latter case faced with two devices of completely different design, both claiming to have the same impedance characteristics. We could not reject one of them. The impedance can be obtained in many ways and in many different constructions.

Helle: Do those two artificial mastoids show the same results with the same bone vibrators?

Dalsgaard: As far as I know, they do not.

Nielsen: Will we see more couplers all claiming to be the same impedance?

Dalsgaard: I understand your fear. The 2 cm^3 coupler has the advantage of being extremely simple. It's specified in terms of a mechanical drawing and a skilled mechanic can make it. It should undoubtedly be retained for many uses.

Helle: Is it enough to specify the performance?

Burkhard: In principle, yes. If you specify the characteristic impedance of the ear canal that you are going to simulate, that fixes its diameter. Then if you specify either the standing wave ratio or the transfer function, these two parameters together fix the length of the ear canal. In fact, the standing wave ratio or the transfer impedance will also fix the termination. The termination is a complex impedance for which there may be a number of constructions that will work. The important thing is that the three or four parameters be specified in a way that they can be measured easily with conventional laboratory equipment.

Lauridsen: Mr. Burkhard, you mentioned before, when discussing the side resonators of a Zwislocki-type coupler, that it is possible to manufacture the resistance screens with good precision. Would it not be possible to define simply the mechanical shape and then incorporate the measures of these resistance screen elements to produce one well defined kind of coupler that could be standardized?

Burkhard: This would likely be possible. There is still a need, however, for performance specifications and ways of measuring the properties because the devices will be more complicated and will have potential for changes during use due to accumulation of dirt in the holes of the side branches.

Dalsgaard: We must face the fact that the Lord is not the standardizing person. Man is manufactured within a very wide tolerance. This is a fact that we must face when we are working with these devices and manikins. I am a little afraid of introducing manikins in measurements because I have the feeling that people in the clinics will then believe that the results you get on the KEMAR are the truth, the whole truth, and nothing but the truth for that particular hearing aid. Some years ago we had the experience in the Danish Hearing Centers that when they had a patient they looked at his audiogram and then went to the file of response curves to find the particular hearing aid that had the best fit. Fortunately, they do not do this any more. I feel that we might return to that situation when we get more realistic measurements on hearing aids as are provided by a manikin. The only solution, as I see it, is to continue to educate people in the clinics.

Anonymous: If we are to switch from reporting of hearing aid data in the present IEC standard methods and to the reporting of data on a manikin or an *in situ* condition, we face a real problem of how shall the transition be done.

Burkhard: I think that is a very important point. I see no alternative to recognizing that there will be a period of time in which double or dual reporting of hearing aid characteristics may occur. Even if manufacturers do not use the manikin, they likely will report data for public use on a different coupler.

Grant-Salmon: This relates to my comments earlier. My point is that in order to produce some commitment to uniformity, there has to be agreement among manufacturers.

Dalsgaard: That is perfectly right. We had such agreement when we prepared IEC documents 118 and 126 because we agreed in 1959 to exchange technical specifications based on measurements made in accordance with these procedures. In these documents it was stated that you would not use the results for clinical purposes. It could be possible that a standard would be used for the exchange of specifications, but in the development and research in the hearing aid field very elaborate artificial ears which need not be standardized could be used. One must then ask how bad is the 2 cc coupler if one's purpose is only to exchange technical specifications.

Helle: The $2cm^3$ coupler is very simple and it is easy to use in production. But for development, I believe we need something better.

Anonymous: Has there been experimentation on the size of the pinna for the KEMAR manikin?

Burkhard: We have not done any ourselves. This is a parameter that we were very much concerned about and that is why we made provision for substituting other pinnae.

Silbiger: Do you have data on the compressibility of the pinna? Is there any relationship between force as applied to KEMAR's pinna to give a certain deflection and the use of earmolds?

Burkhard: We have made a trade-off in the design and construction of the pinnae. We expected that it would be handled frequently. We picked a material that is reasonably robust, that wouldn't tear or crumble with repeated handling. The material is an RTV silicone rubber that is very flexible and yet has good tear resistance in thin sections. We added additional thickness to the portion of the pinna that forms the back wall of the concha to increase its strength. This thickness has made the ear somewhat stiffer for compression under earphones than you normally find on most people. We let the upper portion of the pinna be more or less of normal thickness to allow flexibility to accommodate a range of hearing aids that would be placed over and behind the ear.

Knowles: I would like to add a word to that because through omission the comments might be misunderstood. The rigidity of the pinna does not significantly influence the diffraction pattern and acoustic measurements that one would make in the ear with an open exposed free field condition or when a hearing aid is in place. The two occasions in which the compliance of the pinna and the material in particular become critical and in which this slightly stiffer model may be unsuitable are 1.) The case in which you use a supra-aural earphone. The pinna may not compress in a manner comparable to the way it would compress on a person with normal headband pressures. 2.) If one is using circumaural earphones and trying to take advantage of the slight amount of bulk compression of the pinna. I can generalize by saying that for conventional hearing aid use in which this first approximation design of a manikin is aimed, this trade-off is eminently satisfactory. But for the next steps of relating to earphone measurement, additional precautions are required.

Tedder: What is the effect of clothing on the results with the KEMAR manikin?

Burkhard: There is an effect which is shown in the illustrations in the paper on describing the development of KEMAR (Chapter 2). There is a rise in the pressure at the eardrum in the vicinity of 900 Hz and then a decrease in the vicinity of 1300 to 1700 Hz which is primarily due to an interference between sound arriving directly at the ear and sound arriving by reflection off of the upper torso. When clothing is added, this interference is diminished and less perturbation in the response at an eardrum will be observed. The effect is the order of a few dB and depends on the amount and type of clothing that is provided.

Kuhn: Mr. Burkhard, one figure that was of interest to me in particular was the one where you changed the surface impedance of the head itself. (Fig. 2-12) It suggests that basically the eardrum pressure is only a function of the local impedance around the pinna and not much dependent, until you get into the shadow, on the surface impedance of the head itself. This makes physical sense but I am quite surprised at the small differences. In the shadow, of course, you only have sound pressure due to the scattered sound wave.

Kuhn: Dr. Causey, in one of your slides you showed the effect of rotating the manikin 2½ and 5° and I think that was for frontal incidence. I think you stated that there was less than 2½ dB difference between frontal incidence and plus or minus 5°. This should be true only for frontal incidence. If the head is at 90° or some other angle the effect can be quite a bit more.

Rice: I am interested in the measurements at the eardrum or underneath an earmold. How much does the pressure between the end of the earmold and the drum vary? Where can you specify an eardrum measurement to be made? The pressure across the eardrum can vary and the problem of ambiguity of measurement is eventually coupled with the probe placement in that situation. My concern and question about using probe measurement techniques is how unique are they in a particular pressure measurement. Can one really relate them to the manikin measurements?

Burkhard: As you know, we have made a study of the reliability of the probe measurement under an earmold. We concluded that if a probe extended some distance away from the opening into the occluded ear, the reliability was certainly improved. The probe must be placed next to the eardrum if one is to measure correct eardrum pressure at high frequencies. The wave length must be very small or the frequency very high and nearly outside the range of audibility before the pressure variations across the eardrum become an important parameter. One can make estimates of this.

Villchur: As I understand, part of the question has to do with the possible influence of the measuring probe in the ear canal. Killion of Industrial Research Products, Inc., sent me a set of curves he took on KEMAR. One curve was with an earphone on the KEMAR and one was with free field. Each pair of curves showed one plot of the eardrum response with the probe microphone inserted approximately to the eardrum position, and one with the probe absent. The two curves lie almost on top of each other. The probe microphone used is one that was described in a letter to the editor in the Journal of the Acoustical Society. The diameter of the probe is the order of 1.3 mm and the diameter of the ear canal in KEMAR is 7.5 mm.

Shaw: I may say that in the classical study of Weiner and Ross, this question was answered up to perhaps 6 kHz. One probe tube was inserted into the ears of subjects and then a second probe was also inserted into the ear canal to determine how much change occurred. The changes observed with the addition of the second probe were very small.

Burkhard: Mr. Preves, I would like to ask you to go through your comments again about what you would recommend for the conditions to measure SSPL90 and the distortion measurement.

Preves: My feeling is that the substitution method with the tape or with some kind of equalizer should be used for both distortion measurements and SSPL90 and that they should be used with the KEMAR manikin because you want to represent what the median person is going to hear. That can only be done by taking into account whatever head diffraction there is, ear canal resonance, pinna effect, etc. So, yes, the measurements should be done on KEMAR using the substitution method, not the orthotelephonic method. When one looks at directional hearing aids, the orthotelephonic method becomes complicated because one must use a different recorded signal for each sound direction.

Tedder: Mr. Preves, have you tried plotting the difference between the substitution method and the control microphone? Obviously the control microphone would be the handiest.

Preves: No, nothing more than what I showed in my slides concerning directional hearing aids.

Punch: What methods do you use or recommend for fabricating an earmold to ensure minimum leakage in the manikin?

Burkhard: That has been a problem for us. We were not able to have custom molds made for the KEMAR ear in any quantity. This is why the alternate approach of an adapter has been proposed and shown today.

Studebaker: We have found, and I would like to suggest, that the ADCO Mold type of material is very satisfactory for leak-free molds on the KEMAR ear.

Cole: I have been using a brute-force method to measure directional hearing aids. It consisted of selecting speakers that can be made reasonably smooth with equalization. So far I have been able to get ± 1 dB from 300 to 4000 Hz. I use a 1/3 octave band filter set. It is probably not good enough at this point. One has to select the loudspeaker carefully. In my opinion there are fewer problems of repeatability, including the problem of synchronization between a tape recorder and a response plotter. The method has many advantages when testing directional hearing aids and when using noise, speech or other complex test signals.

Preves: There doesn't seem to be a definite trend on fitting hearing aids but usually the responses are based on supra or circumaural type earphone measurements which do not take head diffraction into account. If people fit from that type of audiometric data, then they should be adding the diffraction effects of the head. This also supports the substitution method of hearing aid measurement. If the threshold measurement is done in a free field, then the effect of the head is taken into account.

Causey: I think Mr. Preves is correct. You will have different distortion measurement results depending on the methods of testing. At the moment I think I would be happier with the orthotelephonic gain measurement in terms of free field excitation. But I also recognize that hearing aids will show up better on distortion measurements with that kind of measurement. By the way, when I say "relate to clinical measures" I change my clinical measures to a free field kind of assessment rather than earphone measurements.

Cole: When one talks about the SSPL90, the values measured with a KEMAR manikin are in general considerably higher in the high frequency region than we are used to seeing on a 2 cm^3 coupler. I have been asked a number of times whether this difference is significant. One has to ask whether the numbers one reads on an audiometer are realistic in terms of the sound appearing at the eardrum. Has anyone put an audiometer on the KEMAR manikin to see what sound appears at the eardrum when the dial says there is 100 dB hearing loss? If we are to relate the SSPL90 to real ears, we have to know the answer to that question.

Burkhard: There have been measurements with audiometer-type earphones on the manikin made by different persons. I do not have their results. Examples of data that we have taken are included in these proceedings.

Helle: Preves has shown us two different measurements: one where he had a substitution method in which he produced a free field sound pressure with a tape recorder and then placed the manikin with hearing aid there, and a second measurement where he had a microphone above the head of the manikin. The second method was in effect a comparison. You also have a substitution method when the reference is the sound level in the manikin ear and then the hearing aid is put on the manikin and the measurement is repeated. If you control the signals for the test in the other ear which would be acceptable for frequency below 8000 Hz, then you have a comparison method. I am not very happy about using the terms substitution method and comparison method, which to me are quite general, to describe the special measurement of etymotic gain.

Burkhard: We obviously need to define our terms very carefully. Dr. Causey, what is the substitution gain in your experiments?

Causey: It is the difference in gain from SPL at a test point and SPL at the manikin eardrum, with the hearing aid in place, thus the free field is the reference condition.

Burkhard: Mr. Preves, do you define your substitution method the same as Dr. Causey?

Preves: Yes.

Tedder: How much alike or different are the KEMAR manikins?

Burkhard: All of the manikin bodies and torsos are made from one mold. The ears are molded from renewable molds made from a single pattern for each ear. To the extent that objects can be reproduced from molds, the manikins are alike. This holds also for the differences between ears in the vicinity of 8 kHz.

Villchur: I have a question concerning the control technique in which the pressure at the hearing aid is controlled with an adjacent microphone. Please describe the signal from the speaker in the sound box. (Fig. 12-8).

Burkhard: If the auxiliary equipment consisting of tape recorder and divider were not in the circuit, the loudspeaker signal would be essentially constant with frequency or, more particularly, the pressure at the microphone would be constant with frequency. The signal from the tape recorder acting through the divider network causes the response of the microphone channel to vary with frequency which is analogous to a microphone with a response function inverse of the pressure that we intend to develop at the microphone. By receiving a signal from the microphone that is decreasing, or increasing, the servo control component tries to compensate the drive to the loudspeaker. As I indicated in the presentation, that divider function could be done with filters having the shape indicated in one of the illustrations. The divider circuit is quite inexpensive to manufacture with today's integrated circuits.

Kuhn: In view of the results of the paper by Burnett and myself, where we gave the differences in sound pressure level along the side of the manikin head, and your measurements of hearing aid gain with a reference microphone, I would have expected results to not be as good as they are. What I think may have happened in this experiment that is with a larger microphone a pressure is averaged across the diaphragm. We showed results where at over half a centimeter from the head one gets as much as 4 dB difference from one location to the next in the upper frequencies. Your agreement is much better than 4 dB. I would think that if one uses a small probe the results may be different.

Carlson: There may be a confusion here. The use of a 1" microphone was merely to control the pressure at the hearing aid sound port. It was not used for the measurement of a sound field, nor the sound on the KEMAR head.

Kuhn: Is it not the pressure over a 1" diaphragm, so that you have not really looked at what the pressure was before you placed the hearing aid in the sound field?

Burkhard: In this method it is not necessary to know what the sound field was before the objects were placed there.

Villchur: Was the 1" microphone used as a sound source?

Burkhard: No, it was used as a control.

Villchur: When the microphone is spaced a quarter of an inch from the hearing aid microphone, what is it doing then?

Carlson: The measurement shown in the three figures (Figs. 12-2, 12-3, 12-4) merely demonstrates that the spacing between the hearing aid and the control microphone and the direction of sound incidence is not an important factor. Perhaps the thing that is missing in the data in the figures is the response of the hearing aid microphone by itself. That response, however, would be negligibly different from the one shown here for the 90° orientation of the control microphone to the sound field.

Helle: Your conclusion was, then, that if you have a very small distance between the control microphone and the hearing aid and there is no difference for different directions up to 10 kHz, then you can go to the box.

Burkhard: Yes, that is correct.

Helle: Then my next question is: Have you done those measurements in a box with small controlling distances with a microphone smaller than one inch?

Carlson: We didn't do any of these in a sound box where we varied the distance. We did vary the angle. But this all comes down to an appreciation for how much the pressure can change in a given distance. The spacing has to be a fraction of a wavelength at the highest frequency to avoid problems with standing waves between the control microphone and the hearing aid.

Helle: In a sound box, the diaphragm is normally pointed in the direction of the progressing wave front. That means the wave front is perpendicular to the diaphragm. Would this control still be maintained?

Carlson: I don't think it will make any difference if you place the sound port of the hearing aid close enough to the diaphragm, and that is the intention here, but maybe you should not be closer than the diameter of the hearing aid sound port. Then you may start to influence the behavior of the hearing aid. The opening of the microphone in the hearing aid is small enough that the pressure across it is uniform even at high frequencies. However, with the controlling microphone there will be averaging when the wavelength is as short as or shorter than one to three times the diameter of the microphone. Fortunately, the average is quite similar to the pressure at the center even at these frequencies.

Participants

Manikin Measurement Methods Conference – Zurich

March 4, 1976

Attendees	Affiliation
Frau S. Bankau	Microelectrik
H. Bergenstoff	Danavox
O. Berland	Oticon
P. Bommer	Bommer
P.V. Brüel	Brüel & Kjaer
M.D. Burkhard	Industrial Research Products, Inc.
A. Constam	A.G. Constam
H. Dalsgaard	Oticon
S.C. Dalsgaard	Odense University
B. Diethelm	AG für Electro-Akustik
K. Fey	Bommer
D. Fischer	Willco
W. Grandjet	Ausbudungszentrum – Lubeck
A.K. Grant-Salmon	Knowles Electronics, Ltd.
R. Helle	Siemens AG
M. Houdred	N.V. Philips
Hr. Huber	Bommer
F. Hueber	Viennatone
Hr. Jakober	Gfeller
M. Kleiner	Chalmers Univ., Göteborg
O. Lauridsen	Topholm and Westerman
M. Lower	Southampton University
T. Nielsen	Oticon
S. Oloffson	Karolinska Institute-Stockholm
G. Piga	Amplifon
G. Rasmussen	Brüel & Kjaer
Dr. Richter	P.T.B. – Braünschweig
C. Starke	Siemens AG
J. Topholm	Topholm and Westerman
R.J. Wilton	Knowles Electronics, Ltd.

Participants

Manikin Measurement Methods Conference—Washington DC

April 5, 1976

Attendee	Affiliation
J. Agnew	Vicon Instruments
W.F. Balmer	Radioear Corporation
L. Beck	VA Hospital, Washington DC
N. Blaskovich Jr.	U.S. Dept. HEW-NIOSH
Klaus Brinkmann	P.T.B.—Braunschweig
M.D. Burkhard	Industrial Research Products, Inc.
E.V. Carlson	Industrial Research Products, Inc.
D. Causey	VA Hospital, Washington DC
M.R. Chial	University of Texas at Austin
W.A. Cole	Linear Technology, Inc.
S. Connors	Northwestern University
R. Cox	Memphis State University
J.L. Danhauer	Bowling Green State University
B.J. Edgerton	Bowling Green State University
E. Elkins	VA Hospital/Washington DC
W.G. Ely	Qualitone
R.O. Helle	Siemens AG
J.H. Janssen	
T. Kallberg	Audiotone
H.S. Knowles	Knowles Electronics, Inc.
G. Kuhn	National Bureau of Standards
B.W. Lawton	
N. Lewis	HC Electronics, Inc.
M.C. Martin	Royal National Institute for the Deaf
R. Miner	Wood County Educational Audiological Serv.
P.L. Michael	Penn State University
I.V. Nabelek	University of Tennessee
M. Pavel	Bell Telephone Laboratory
L. Pierce	Bell Telephone Laboratory
D. Preves	Starkey Labs
J.L. Punch	University of Maryland
C. Rice	Southampton University
E. Rindbo	Danavox Electronics A/S
H.R. Schechter	I.I.T. Research Institute
H.C. Schweitzer	VA Hospital/Washington DC
H.R. Silbiger	Bell Telephone Laboratory
Don Sims	Rochester Institute of Technology
J. Smaldino	VA Center, Temple TX
G. Studebaker	Memphis State University
H. Teder	Telex Comm/Minneapolis
A.T. Tutins	Knowles Electronics, Inc.
E. Villchur	Foundation for Hearing Aid Res.
H.F. Waller	Knowles Electronics, Inc.
D. Wark	Memphis State University
R.E.C. White	City University of New York
L. Young	Northwestern University