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MEASURING THE REAL EAR RESPONSE CHARACTERISTICS
OF HEARING AIDS

by

Michael Charles Lower

March 1980
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CONTENTS

Acknowledgement i
Contents ii
List of Figures viii

ABSTRACT

CHAPTER 1  INTRODUCTION 1

CHAPTER 2  MEASUREMENT OF HEARING AID RESPONSES 7
2.1  A BRIEF HISTORICAL PERSPECTIVE 7
2.2  CURRENT PROCEDURES AND THEIR LIMITATIONS 11
   2.2.1 Problems in comparing responses via the 2cc coupler 13
2.3  SOME RECENT DEVELOPMENTS IN OBJECTIVE MEASUREMENTS 14
   2.3.1 New couplers (ear simulators) 14
   2.3.2 Manikins for hearing aid measurements 16
2.4  SOME RECENT DEVELOPMENTS IN SUBJECTIVE MEASUREMENTS 17
2.5  POSSIBLE METHODS OF MEASURING AID RESPONSE 18

FIGURES

CHAPTER 3  METHODS, SUBJECTS AND FACILITIES USED IN THIS STUDY 19
3.1  INTRODUCTION 19
3.2  CHOICE OF A METHOD OF MEASURING THE REAL EAR RESPONSE OF A HEARING AID 19
3.3  THE LOUDNESS BALANCE METHOD 20
   3.3.1 Reasons for choice 20
   3.3.2 Outline of method 20
   3.3.3 The loudness balance audiometer 21
   3.3.4 Accuracy of loudness balancing 23
3.4  SUBJECTS 23
3.5  AIDS AND EARMOULDs 24
3.6  MEASUREMENTS ON COUPLER AND EAR SIMULATORS 25
   3.6.1 Couplers and ear simulators used 25
   3.6.2 Tubing and couplings 26
   3.6.3 Method 26
3.7  THE MANIKIN KEMAR 26
3.8  ANECHOIC ROOM 27
3.9  OTHER EQUIPMENT 27

FIGURES
CHAPTER 4 EXPERIMENTAL MEASUREMENTS OF HEARING AID GAIN BY LOUDNESS BALANCE, 2cc COUPLER, ZWISLOCKI EAR-SIMULATOR AND MANIKIN

4.1 INTRODUCTION
28

4.2 METHODS
28

4.2.1 Loudness balance measurements
28

4.2.1.1 Experimental design
28

4.2.1.2 Subjects
29

4.2.1.3 Equipment and facilities
29

4.2.1.4 Procedure
29

4.2.2 Measurements on coupler and ear simulator
31

4.2.3 Measurements on "Kemar"
31

4.3 RESULTS
31

4.3.1 Loudness balance measurements
31

4.3.2 Objective measurements
32

4.4 DISCUSSION
32

4.4.1 Earmould leakage
35

4.4.2 Differences between occluded and unoccluded listening
38

4.4.3 Summary of discussion and future work necessary
40

4.5 SUMMARY AND CONCLUSIONS
40

TABLES

FIGURES

CHAPTER 5 A PROBE-TUBE MICROPHONE AND ITS CALIBRATION

5.1 INTRODUCTION
42

5.2 DESCRIPTION OF THE PROBE MICROPHONE
42

5.2.1 Advantages of this microphone
43

5.3 GENERAL REMARKS ON PROBE MICROPHONE CALIBRATION
43

5.4 THE CALIBRATION OF THE MICROPHONE
44

5.4.1 Method
44

5.4.2 Frequency response and sensitivity
45

5.4.3 Stability of the microphone sensitivity
45

5.4.4 Noise floor
45

5.4.5 Directionality of the probe
45

5.4.6 Leakage of sound through the probe wall and at the connection of the probe to the microphone
46

5.4.7 Accuracy of the calibration under non-free field conditions
46

5.6 SUMMARY AND CONCLUSIONS
46

FIGURES
CHAPTER 6  SOUND LEVEL MEASUREMENTS IN AN AIDED AND UNAIDED MANIKIN'S EAR

6.1 INTRODUCTION

6.2 PUBLISHED DATA ON SOUND LEVEL VARIATIONS IN REAL EARS

6.2.1 Open ear, loudspeaker listening
6.2.2 With ear occluded by insert earphone or earmould

6.3 NECESSITY FOR MORE DATA

6.4 PROBE MEASUREMENTS OF SOUND PRESSURE IN THE EARCANAL OF THE MANIKIN KEMAR

6.4.1 Aims
6.4.2 Measurements in the unaided, unoccluded ear
  6.4.2.1 The effect of the probe on the existing sound field in the ear
  6.4.2.2 The probe microphone and the eardrum microphone compared
  6.4.2.3 Sound pressure level variations within Kemar's open earcanal
  6.4.2.4 Free field to eardrum pressure transfer ratio for Kemar

6.4.3 Measurements in the aided ear of Kemar
  6.4.3.1 The probe microphone and eardrum microphone compared
  6.4.3.2 Sound pressure level variations in Kemar's aided ear

6.5 SUMMARY AND CONCLUSIONS

FIGURES

CHAPTER 7  SOUND PRESSURE LEVEL VARIATION ALONG AIDED AND UNAIDED REAL EARCANALS

7.1 INTRODUCTION

7.2 SAFETY CONSIDERATIONS

7.3 METHOD
  7.3.1 Unoccluded, unaided real ear measurements
  7.3.2 Aided, occluded real ear measurements
  7.3.3 Other measurements

7.4 RESULTS

7.5 DISCUSSION
  7.5.1 Variations in sound pressure levels along unoccluded, unaided real earcanals
  7.5.2 Variations in sound pressure level along earmould occluded earcanals
  7.5.3 Implications for hearing aid gain measurements

7.6 CONCLUSIONS

TABLES

FIGURES

iv.
9.4 SUMMARY AND DISCUSSION TO CHAPTER 9
TABLES
FIGURES

CHAPTER 10 A GENERAL DISCUSSION
10.1 INTRODUCTION
10.2 COMPARISON OF HEARING AID RESPONSES MEASURED BY DIFFERENT METHODS
10.2.1 Insertion gain methods
10.2.2 Transmission gain methods
10.2.2.1 Zwischki and IRPI ear simulators compared
10.2.2.2 Zwischki ear simulator and 2 cc coupler compared
10.2.2.3 Zwischki ear simulator and real ear compared
10.2.3 Transmission gain and insertion gain compared
10.2.4 Keman and real subjects compared
10.3 IMPLICATIONS OF THIS STUDY FOR HEARING AID RESPONSE MEASUREMENTS
10.4 EFFECT OF AID CIRCUIT NOISE ON GAIN SETTINGS CHOSEN BY HEARING AID WEARERS
10.4.1 Feasibility and desirability of low noise hearing aids. Implications from these experiments for loudness balance studies
10.5 IMPLICATIONS FROM THESE EXPERIMENTS FOR LOUDNESS BALANCE STUDIES
10.6 SUGGESTIONS FOR FURTHER WORK
FIGURES

CHAPTER 11 CONCLUSIONS

APPENDIX A VALIDATION STUDY OF THE LOUDNESS BALANCE TECHNIQUE
A.1 INTRODUCTION
A.2 PROCEDURE
A.2.1 Experimental design
A.2.2 Subjects
A.2.3 Facilities and equipment
A.2.4 Experimental procedure
A.3 RESULTS AND ANALYSIS
A.3.1 Statistical analysis of early data
A.3.2 Analysis and results of complete experiment
A.3.3 A note on averaging balance errors
A.4 DISCUSSION
A.5 CONCLUSIONS
TABLES
FIGURES
APPENDIX B. SOUND ATTENUATION PROPERTIES OF HEARING AID EARMOULDLS

B.1 INTRODUCTION
B.1.1 Aims

B.2 THE EARMOULDLS

B.3 METHOD
B.3.1 Choice of method
B.3.2 Principle of the threshold shift measurement of attenuation
B.3.3 Subjects
B.3.4 Facilities and equipment
B.3.4.1 Test room
B.3.4.2 Equipment
B.3.5 Calibration
B.3.6 Experimental procedure

B.4 RESULTS

B.5 DISCUSSION
B.5.1 Threshold data
B.5.2 Factors affecting earmould attenuation
B.5.2.1 Type of mould - probe tubes
B.5.2.2 Left ear/right ear mould differences
B.5.2.3 Vaseline
B.5.2.4 Possible variations in earmould manufacture
B.5.3 Threshold shift and loudness balance measurements compared

B.6 CONCLUSIONS

TABLES

FIGURES

APPENDIX C PILOT STUDY ON THE USE OF PROBE TUBE MICROPHONE FOR REAL EAR MEASUREMENTS

APPENDIX D ESTIMATED MAXIMUM GAIN SETTINGS OF HEARING AIDS SUCH THAT CIRCUIT NOISE WOULD NOT MASK AIDED THRESHOLDS

REFERENCES
LIST OF FIGURES

Figures are numbered sequentially within each chapter or appendix.

2.1 West's G.P.O. artificial ear, 1930.
2.2 Inglis, Gray and Jenkin's artificial ear, 1932.
2.3 Littler's apparatus for threshold measurements of hearing aid response, 1936.
2.4 Romanow's designs for 2cc and 6cc couplers, 1942.
2.5 Insertion and transmission gains compared.
2.6 Real ear vs. 2cc coupler response for various receivers (after Nichols et al, 1946).
2.7 Real ear vs 2cc coupler response with various earmould vents (after Studebaker and Zachman, 1970).
2.8 Response of an aid on 2cc coupler and on real ears by loudness balance (after Starke, 1970).
2.9 Response of an aid on the 2cc coupler and on real ears by loudness balance (after Lower, 1975).
2.10 Response of an aid on the 2cc coupler and on real ears by a threshold method (after Pascoe, 1975).
2.11 Response of an aid on the 2cc coupler and on real ears by acoustic reflex threshold shift (after Tonisson, 1975).
2.12 Responses of two aids on the 2cc coupler and on real ears measured with a probe microphone (after Dalsgaard and Jensen, 1976).
2.13 Response of an aid on the 2cc coupler and on real ears by acoustic reflex threshold shift (after Pizarro, 1976).
2.14 Published real ear measurements of hearing aid insertion responses, compared via their respective 2cc coupler transmission responses.
2.15 The Zwislocki ear simulator, longitudinal section and bottom aspect.
2.16 The PTB ear simulator (2-branch).
2.17 The XD-1053 two branch ear simulator of Industrial Research Products, Inc.
2.18 Shaw's Unit-D ear simulator.
2.19 Tentative B & K design for Type 4154 artificial ear.
2.20 Keller's conical coupler.
2.21 A classification of methods of measuring hearing aid responses.

3.1 Philosophy behind the loudness balance method of measuring hearing aid gain.
3.2 Schematic diagram of the loudness balance audiometer.
3.3 Stimulus presentation sequence.
3.4 The ear simulators used in this study (photograph).
3.5 Connecting an aid to a coupler or ear simulator, dimensions used.
3.6 Equipment used to measure transmission responses of aids on coupler or ear simulator.
3.7 The manikin, Kemar (photograph).
3.8 Measured background noise in test room.
3.9 Comparison of sound field in room with ideal free-field conditions.

viii.
4.1 The Widex "Baritone" 641 H aid.
4.2 Block diagram of equipment used for the loudness balance measurements.
4.3 Subjective response of the Widex Baritone aid.
4.4 Response of the Widex Baritone on the 2cc coupler and Zwischocki ear simulator.
4.5 Response of the Widex Baritone on Kemar.
4.6 Difference between coupler and subjective responses.
4.7 Subjective response with allowance for head diffraction.
4.8 Subjective response with allowance for head diffraction and earcanal resonance effects.
4.9 Differences between coupler and subjective responses of the aid when head diffraction and earcanal resonance effects are removed from the comparison.
4.10 Response of the aid on Kemar with allowance for human head diffraction and earcanal resonance effects.
4.11 Comparison of aid gain and earmould attenuation.
4.12 Cumulative distributions of aid gain and earmould attenuation measurements at 250 Hz and 500 Hz.

5.1 a. The XL-9073 probe microphone (photograph).
b. The probe in Kemar's unaided ear (photograph).
5.2 Equipment used to calibrate the probe-tube microphone.
5.3 Orientation of the probe and reference microphone to the sound field.
5.4 Frequency response of the XL-9073 microphone unit without a probe tube.
5.5 Frequency response of the XL-9073 microphone with a medium length probe tube.
5.6 Frequency response of the XL-9073 microphone with the longest probe tube.
5.7 Directionality of the probe microphone.

6.1 Frequency response of the Goodmans Axiete loudspeaker.
6.2 Harmonic distortion of the Goodmans Axiete loudspeaker.
6.3 Change in Kemar's unaided 'eardrum' SPL on the insertion of the probe microphone to within 1 1/2 mm of the eardrum microphone.
6.4 Comparison of the probe and eardrum microphone measurements of SPL in Kemar's unaided ear.
6.5 Variation in SPL in Kemar's unaided ear relative to canal entrance.
6.6 Variation in SPL in Kemar's unaided ear relative to a position adjacent to the eardrum.
6.7 Variation in SPL along Kemar's unoccluded earcanal.
6.8 Freefield to eardrum pressure transfer ratio for Kemar.
6.9 Measuring sound pressure under a simulated hearing aid fitting on Kemar.
6.10 Comparison of probe and eardrum microphone measurements in Kemar's ear under a simulated hearing aid fitting.
6.11 Change in eardrum SPL in Kemar when the probe is withdrawn to the earmould from a position 1 mm from the eardrum.
6.12 Variation in SPL in Kemar's aided ear.
7.1 Schematic diagram of equipment used to map SPL in unaided, unoccluded ears.
7.2 Schematic diagram of equipment used to map SPL in ears occluded with earmoulds.
7.3 An earmould with a microphone probe tube in addition to the sound inlet tube (photograph).
7.4 Sound level variations along unaided earcanals: 4 subjects' left ears.
7.5 Sound level variations along unaided earcanals: mean of 4 real ears cf. Kemar.
7.6 Sound level variations under hearing aid earmoulds: left ears of 4 subjects.
7.7 Sound level variations in aided ears occluded with earmoulds: real ears and Kemar.
7.8 SPL generated in the occluded ear 5 mm beyond an earmould by a hearing aid receiver: mean of 4 real ears cf. Kemar (Zwislocki ear simulator).
7.9 Sound pressure transfer ratio from the earcanal entrance to a point 12 mm into the canal.
7.10 Sound pressure transfer ratio from the earcanal entrance to a point 10 mm into the canal.

8.1 The BE 11 hearing aid (photograph).
8.2 Equipment used for the loudness balance with simultaneous measurement of earcanal SPL.
8.3 Equipment used for the acoustic reflex measurements.
8.4 Subjective gain of the BE 11 aid measured by loudness balance.
8.5 Pressure insertion gain of the BE 11 aid measured with the probe microphone.
8.6 Gain of the BE 11 aid measured on individuals by acoustic reflex and probe microphone methods.
8.7 Comparison of loudness balance acoustic reflex and probe microphone measurements of insertion gain of the BE 11 aid: means of all data.
8.8 Response of the BE 11 aid on the 2cc coupler.
8.9 Response of the BE 11 aid on the Zwislocki and IRPI XD-1053 ear simulators.
8.10 The discrepancy between pressure insertion gain and subjective gain of the aid.
8.11 Simplified block diagram of the middle ear muscle reflex as a multipath feedback mechanism (after Dallos [12, 1973]).
8.12 Partial masking of a 980 Hz tone as a function of the SPL of the masking noise (after Scharf [17, 1964]).
8.13 Levels set by subjects to match loudness of reference tones for various levels of reference tone: aided listening.
8.14 Levels set by subjects to match loudness of reference tones for various levels of reference tone: unaided listening.
8.15 Subjective gain plotted as a function of the level of the reference tone.
8.16 Waveforms of the 1 kHz tone pulse measured aided and unaided on Kemar.
8.17 Leading edges of pulses shown in Figure 8.16.
8.18 Trailing edges of pulses shown in Figure 8.16.
9.1 Alto Acoustics Ltd. "Focus" aid (photograph).
9.2 Widex 691 aid (photograph).
9.3 Frequency responses of the BE 11 aids on the 2cc coupler.
9.4 Frequency response of BE 11 No. 3 on the Zwislocki ear simulator.
9.5 Frequency response of the Alto Focus on the 2cc coupler.
9.6 Frequency response of the Alto Focus on the Zwislocki and XD-1053 ear simulators.
9.7 Frequency response of the Widex 691 on the 2cc coupler.
9.8 Frequency response of the Widex 691 on the Zwislocki and XD-1053 ear simulators.
9.9 Variation in the SPL giving equal loudness in aided ears with the aid gain: individual subjects.
9.10 Mean variation in the SPL giving equal loudness in aided ears with mean aid gain.
9.11 Subjective gain vs. pressure insertion gain for the BE 11 aid: individual subjects.
9.12 Mean subjective gain vs. mean pressure insertion gain for all aids in this experiment (9.2). 
9.13 Relationship between the mean SPL in the aided ear at equal loudness and aid generated noise levels.
9.14 Aid generated noise level as a function of aid gain.
9.15 Mean subjective gain of every aid tested in this study plotted as a function of pressure insertion gain.
9.16 Partial masking of a 1 kHz tone - data from Zwicker [9, 1963].
9.17 Partial masking of a 1 kHz tone - data from Hellman and Zwislocki [10, 1964].
9.18 BE 11 aid with its microphone driven by sound from a receiver via a polythene tube (photograph).
9.19 BE 11 modified to allow external signals to be fed directly to its receiver (photograph).
9.20 a. Modification to the BE 11 aid enabling externally generated signals to be fed directly to the receiver. 
   b. The circuit used to supply such signals.
9.21 Equipment used for the experiment described in section 9.3.

10.1 Comparison between the response on the Zwislocki ear simulator and the response on the 2cc coupler of four different aids, this study.
10.2 Comparison between the response on the Zwislocki ear simulator and the response on the 2cc coupler of typical insert receivers, after Sachs and Burkhard [3].
10.3 Approximate average difference between SPL in real aided ears and SPL in the 2cc coupler.
10.4 Comparison of insertion gain on Kemar with transmission gain on the Zwislocki ear simulator for the BE 11 aid.
10.5 Comparison of insertion gain on Kemar with transmission gain on the Zwislocki ear simulator for the Widex Baritone.
10.6 Comparison of insertion gain on Kemar with transmission gain on the Zwislocki ear simulator for the Widex 691.
10.7 Comparison of insertion gain on Kemar with transmission gain on the Zwislocki ear simulator for the Alto Focus.
10.8 The difference between insertion gain (on Kemar) and transmission gain on the 2cc coupler.
A.1 Equipment used in loudness balance validation study.
A.2 Comparative frequency responses of the receivers measured on a Zwislocki ear simulator.
A.3 Variation in balance error with frequency and intensity.
A.4 Mean balance errors: 88 dB reference level.
A.5 Mean balance errors: 60 dB reference level.
A.6 Male-female differences in balance errors.

B.1 Equipment used to measure the sound attenuation afforded by hearing aid earmoulds.
B.2 Examples of an audiogram showing threshold with and without an earmould worn.
B.3 Pure tone attenuation characteristics of the earplug and muff used to occlude the non-test ear.
B.4 Monaural minimum audible field - this experiment compared with ISO R226.
B.5 Measured attenuation of the earmoulds averaged over all ears and replications.
B.6 Attenuation of normal unvaselined moulds averaged over subjects and replications.
B.7 Attenuation of probe-tube moulds without vaseline averaged over subjects and replications.
B.8 Attenuation of normal moulds with vaseline averaged over subjects and replications.
B.9 Attenuation of probe-tube moulds with vaseline averaged over subjects and replications.

C.1 Variation in sound pressure level along the unoccluded ear canal of a single subject.
C.2 Pressure levels relative to the ear canal entrance at depths into the canal of a single subject.
C.3 Pressure transfer ratio from free field to canal entrance of a single subject.
C.4 Pressure transfer ratio from free field to canal midpoint (14 mm) of a single subject.

xii.
A need was identified in hearing aid research for reliable, validated methods of determining the response of a hearing aid as experienced by the wearer, as opposed to the purely acoustical specification of an aid in isolation. Three methods were systematically investigated and their results compared. Two of these, an acoustic reflex threshold shift and a probe-tube microphone technique were validated, shown to give similar results and proved practical and reliable. These two methods may be used with confidence in research. The third method, a subjective loudness balance between aided and unaided loudness, was found to give considerably lower gain values than the other two methods, on occasion by up to 17 dB. Subsequent investigation showed these low values to be erroneous and attributable to very low levels of wideband noise generated within the hearing aid circuitry. This noise was partially masking the loudness balance test stimuli, even though the stimuli were of the order of 30 dB above the noise. This finding was totally unexpected. It was further found that low levels of wideband noise, even if only marginally audible to a subject, could affect the perceived loudness of tones several tens of decibels higher in level. The loudness balance method, despite its having been used in the past to measure aid responses, was therefore rejected as unreliable. The noise levels generated by the tested aids were not untypical for modern designs and although preventing the use of the loudness balance procedure, would not present any problems in normal day-to-day use.

Measurements were also made on couplers, ear simulators and a manikin to compare their performance against data from real persons. The Zwislocki ear simulator and the IRPI two-branch ear simulator were shown to give very similar results and were better representations of the average real ear than the current standard 2 cc coupler. When the Zwislocki ear simulator is incorporated in a Kemar manikin the free field to eardrum pressure transfer ratio is within 2 to 3 dB of the average measured for real subjects. However, although hearing aid gains measured on the manikin were within 5 dB of average real ear values up to 2 kHz, there was a serious discrepancy of 14 dB at 4 kHz. This discrepancy requires further investigation before a manikin can be accepted as giving an estimate of a typical real ear response of an aid.
CHAPTER 1

INTRODUCTION

Gain and frequency response are two of the fundamental descriptors of the performance of any electroacoustic system. This is no less so with hearing aids. An accurate and reliable knowledge of an aid's response is essential in hearing aid research and in aid dispensing, to the manufacturer, the researcher and the audiologist alike. Yet even today the measurement of aid responses and the interpretation and application of these measurements leaves much to be desired and causes untold confusion.

Much of this confusion arises because the response of a hearing aid is conventionally measured according to universally accepted standard procedures [1, IEC, 1959; 2, BSI, 1968; 3, ANSI, 1960]. In such a procedure a known sound pressure level (SPL) is applied to the hearing aid microphone and the aid's output sound pressure level is measured in a standard hard walled cavity incorporating a measuring microphone and having a volume of 2 cc (i.e., the "2 cc coupler"). The gain of the hearing aid is the difference between the output and input sound pressure levels and its variation with frequency is the frequency response.

This standard method of testing the aid in isolation was developed to enable any given response to be maintained once specified, for production testing and quality control by manufacturers, and for exchange of technical information. For these purposes simplicity and repeatability are more important than realism. The response so obtained is not the response of the aid when worn by the average, typical or representative wearer, nor does it readily relate to any average, typical or representative response. Firstly because the coupler does not represent a real ear, and secondly because no allowance is made for head diffraction effects which modify the input to the aid or the differences in ear canal acoustics between aided and unaided listening. Furthermore, the response of any given aid will differ considerably between individual wearers, perhaps by as much as 20 dB at some frequencies to judge from measurements by Dalsgaard and Jensen [4, 1976]. An aid could perhaps have as many responses as there are people to wear it.

1.
Recently Zwislocki has designed an improved "ear-like" coupler [5, 1970] which, it is claimed, accurately simulates the acoustical performance and dimensions of the median real ear canal, thus potentially overcoming one limitation of the standard coupler. Subsequently Burkhard and Sachs [6, 1975] have designed a manikin (an artificial head and torso to median human dimensions) which incorporates Zwislocki's ear-like coupler and enables head diffraction and ear canal acoustics, and their differences between aided and unaided listening to be included in response measurements, thus potentially overcoming further objections to the standard 2 cc coupler method. The response of an aid on a manikin differs considerably from that on a 2 cc coupler. The third objection, however, the differences between individuals, remains. But the Zwislocki and similar ear-like couplers and manikins are currently not widely used and limited to specialised and research laboratories.

Because the 2 cc coupler method is the only standard method it is used (or misused) for virtually all applications. Consequently a widespread misconception has arisen that the response of a hearing aid is an absolute and invariant quantity, and that the response measured on the 2 cc coupler is the response. This is despite the warnings printed in the standards documents that ...

"The results obtained by the methods specified herein express the performance under the conditions of the test, but will not necessarily agree exactly with the performance of the hearing aid under practical conditions of use. For this reason, the difference between practical and test conditions must be borne in mind in interpreting the test results."

There is an urgent need to dispel this misconception. Its implications for hearing aid research and dispensing are far reaching.

The benefit an aid wearer will obtain will depend on the performance of that aid on that wearer and "under practical conditions of use". Yet hearing aid research into the optimum response for a particular type or degree of hearing loss often relies on correlating some measure of benefit averaged over subjects with aid responses measured on a 2 cc coupler.

Inter-subject differences and interaction between subjects and hearing aids may be such that each aid really has a range of responses...
on real ears, a range of perhaps 20dB in the important speech frequencies above 1 kHz, each subject possibly receiving quite different responses from the same aid. Even with a relatively homogeneous sample of pathologies there will be considerable physical and dimensional differences between subjects such that no single aid will produce the same response on everyone. If the particular measure of benefit is averaged over subjects there will probably seem very little difference between aids, and yet for individuals, one aid may be very much better than another. Such an approach can easily lead to the assumption that subjects are insensitive to even quite large variations in aid response.

Such an approach was used in both the classic Medical Research Council (MRC) [7, 1947] and Harvard [8, Davis et al, 1947] studies and has led to the rationale adopted by the British National Health Service, namely that a few aids with particular, defined responses will adequately fit the majority of hearing impaired listeners. Whilst such an approach may provide the 'best aid' giving some benefit to a great many people, it is unlikely that any given individual will necessarily get the best of all possible aids or responses for him or her, or obtain the maximum benefit.

Assuming that the maximum benefit and optimum response for each individual is required then a more realistic approach to aid response measurement is needed. Such an approach would specify the real response of the aid as experienced by each individual with the aid worn normally (i.e., "in situ"). By comparing the real ear response for an individual with the benefit obtained or by comparing the same real ear response set up for a number of individuals having similar hearing problems with the average benefit obtained, a major source of random variation is at once eliminated. Each unique real ear response will now have associated with it a whole series of coupler responses, as many as there are subjects, but this is not important. Specifying in situ real ear responses would make aid research far more sensitive to changes in aid responses. In using the response which would correlate best with benefit and by cutting out uncontrollable variables this might enable hearing aid dispensing to be based for the first time on sound scientific principles rather than the current trial and error procedures. Measuring real ear responses in a way which will correlate best with benefit with
both individuals and with homogeneous groups of patients is the necessary first step towards prescription rather than dispensing. Indeed there is already some evidence that individual or group prescriptive fitting can be more effective than conventional treatments, [e.g., 9, Fournier, 1968; 10, Pascoe, 1974].

Specifying real ear responses rather than coupler responses enables the total response of the aid and earmould system to be taken into account. Full advantage may therefore be taken of developments such as vented earmoulds, wideband earmoulds, directional microphones, different microphone locations, etc.

But before research into optimum aid responses can proceed and before aid dispensing can be refined to a systematic and scientific prescriptive procedure there must be tried and tested methods of measuring the real ear response of an aid. Some methods do exist, for example, a probe-tube microphone comparison or threshold shift between unaided and aided listening, which were in use before the first couplers were widely available four decades ago. But it is disheartening that methods and their results have never been validated or compared with each other. There are many questions which have never been adequately answered. What methods are feasible? Do they all measure what the experimenter intends? Do they give similar results and, if not, why not? What advantages or disadvantages does each method have? Which method is suited to which application or which class of patient? How do objective measurements on real subjects or patients relate to the response perceived subjectively? These questions implicitly embody the main objectives of this research study.

In attempting to answer the above questions this thesis seeks to establish practical methods of measuring the real ear responses of hearing aids suitable for research and also prescription. These methods are validated by objective acoustic measurement and by comparisons one against another in a fully direct and systematic manner using as far as possible the same aids and subjects for each comparison. This systematic approach is considered to be of major importance and long overdue. Attempts at indirect comparisons from published data from different sources are, as argued below (2.2.1), inherently unreliable.
Further objectives of this study were to investigate the new Zwislocki earlike coupler and the manikin as possible improvements over the 2 cc coupler as potential bases for standardisation. These were in fact the initial reasons for the research but they assumed secondary importance as the experimental programme proceeded towards real ear measurements and the need for reliable real ear methods was more fully realised.

This is not to deprecate the manikin approach. A manikin should represent the closest approach using hardware alone to measuring an insertion gain comparable to that experienced by a real person. The manikin includes the repeatability simplicity and practical convenience of a hardware-based method with the realism of including head, body and ear acoustics and any directivity factors thereof, but in being an average, excludes the applicability of measured responses to a real person which can only be achieved by testing the aid on that wearer. Given the impracticability of testing every aid on every potential wearer, a manikin giving a response somewhere within the range of real ear responses would be as useful as an average real ear response or a typical real ear response, the manikin, average or real ear response being at present equally relatable or unrelatable to the needs of a given but otherwise unknown individual. Under these circumstances a manikin-measured response is preferable to the average or typical real ear response because of its repeatability. A manikin could consequently be an acceptable standardised device offering improvements over the current 2 cc coupler for specifying an aid response. The important questions are:— how closely does a manikin response fit in with the pattern of overall responses? Does a manikin provide a worthwhile improvement on the 2 cc coupler? Would a manikin response be close enough to an individual real ear response to provide useful information in aid prescription?

In response to the main and secondary questions outlined above a sequential series of experiments was conducted.

In order to determine the response of an aid as perceived by the wearer a loudness balance technique was developed. In the first main
experiment the technique was used to measure the subjective response of an aid on normally hearing subjects and to compare this response with objective manikin, 2 cc coupler and Zwislocki coupler responses. The response of the aid on the manikin was largely predictable from the Zwislocki coupler measurements showing that the role of head and ear canal acoustics could be quantifiably accounted for. But the subjective response perceived by the wearers was very different from that expected on the basis of the manikin and coupler measurements, the gain being generally lower than predicted. The results in fact suggested an effect akin to the "Missing 6 dB" whereby different sound pressures between occluded and unoccluded ears have been claimed to elicit the same subjective loudness.

Reasons for this unexpected effect were systematically investigated, possible artefacts being eliminated. The loudness balance repeated with a different aid, but this time supplemented with detailed probe-tube microphone measurements of sound pressures in aided and unaided ear canals together with measurements of aid gain by an acoustic reflex method. Careful preparatory work on suitable probe calibration methods and mapping of sound pressures in occluded and unoccluded ear canals was necessary to obtain results in which confidence could be placed.

The acoustic reflex and probe microphone measurements corroborated each other but the loudness balance still gave gain values considerably lower than expected. It was then known that a much higher sound pressure level was indeed needed in the occluded ear than unoccluded ear for a given loudness and that this was a central, psychoacoustic rather than a physical or physiological or peripheral phenomenon.

Further experiments revealed this phenomenon to be a manifestation of partial masking of loudness balance stimuli by electrically generated noise from the hearing aid. This was surprising considering that signal to noise ratios were generally of the order of 30 dB between test stimuli and the noise measured in the same third octave.

The experimental work is described with appropriate discussion in chronological order in Chapters 3 through 9. Each phase of the work reported in separate chapters is essentially self contained and may be read separately. A general unifying discussion can be found in Chapter 10 in which implications arising from this study both within and outside the hearing aid field are described.
CHAPTER 2
MEASUREMENT OF HEARING AID RESPONSE

2.1 A BRIEF HISTORICAL PERSPECTIVE

Information on the measurement of hearing aid responses prior to the 1940's is scant. The first artificial ears for measuring the output of earphones were designed by West [1] in 1930 and by Inglis, Gray and Jenkins [2] in 1932. As can be seen from Figures 2.1 and 2.2 each had an ear canal representation with a measuring microphone placed obliquely at the eardrum position and an acoustic network to produce the right impedance at the microphone.

They were designed specifically for measurements on telephones but were undoubtedly sometimes used for measuring the outputs of the electrical air conduction aids of the time which had telephone type external "receivers" or earphones. T.S. Littler [3, 1936] gives artificial ears a disparagingly brief mention in a 1936 article on testing hearing aids, most aid measurements at this time being subjective. Incidentally the rather perverse description by telephone engineers of earphones as "receivers" has persisted to the present day for hearing aids, and even the miniature internal output transducers of modern aids are referred to as "receivers".

The subjective measurements described by Littler were based on pure tone "aided" and "unaided" threshold measurements. For example, a patient would sit in a soundproof room lined with absorbent material and his threshold would be determined with and without an ear trumpet or speaking tube at each frequency of interest. The difference in the sound levels in decibels between aided and unaided listening was then the gain of the ear trumpet.

The method used with electrical aids was slightly more complicated. Figure 2.3 shows the set up. An attenuator was connected between the aid amplifier and the aid receiver and was adjusted before the measurements so that the electrical noise from the aid was inaudible or negligible to the patient. The patient sat outside the test room whilst his "aided" threshold was determined (the minimum intensity of sound in the room which

7.
he could hear through the aid). The patient entered the room and sat with his ear where the aid microphone had been whilst his unaided threshold was determined. In decibels, the aid's gain was calculated as the difference between aided and unaided thresholds plus the attenuation introduced between the aid's amplifier and receiver. Although the aid was tested on the ear on which it was to be worn, no allowance was made for the modification of the input to the hearing aid by the presence of the wearer's body acting as a baffle.

In 1942 Romanow [4] described a subjective calibration method which was in use in America. This was an aided and unaided monaural loudness balance method similar to Littler's aided and unaided threshold method. In the aided test the hearing aid microphone was placed in free field and no allowance made for body baffle. This was because some aid users would not always wear their aids but would perhaps place them on a desk, closer to a speaker for example. Romanow suggested that body baffle, if required, be measured separately. Littler described a similar loudness balance in 1944 [5].

Romanow's paper was to have far reaching consequences. In it he argued that, once an aid or earphone had been subjectively calibrated other aids or earphones of the same type, or even the same aid at different times or with, say, different battery voltages, could be compared simply by measuring their outputs generated in a cavity of pre-determined shape containing a measuring microphone. Since it was only to be used for comparative measurements, such a cavity did not need to be as complicated as an artificial ear. No acoustic networks would be needed, and the device could be made very simply and very cheaply. Such a device was known as a "coupler" - it simply coupled an earphone to a measuring microphone. Using a coupler a response of an earphone or hearing aid could be determined instantly, there was no need to employ subjects for repeat calibrations, and any comparisons, being objective, were more precise and repeatable.

Romanow published two designs of coupler and these are shown in Figure 2.4. The "6 cubic centimetre" coupler was for use with external (supra-aural) receivers, the "2 cc coupler" was designed for the new smaller "insert" receivers which were coupled to the ear with earmoulds. The volumes of each type of coupler were approximations to the air volumes
which would normally be trapped in the occluded ear canal.

The 2 cc and the 6 cc couplers quickly caught on. They were ideal for production or batch testing and their ease and simplicity of construction and use was extremely appealing especially during the exigencies of the war. These couplers were adopted by the Joint Radio Board and were often referred to as "JRB" couplers [e.g., 6, Glaser et al, 1948].

Many people began to use these couplers rather than subjective methods to compare earphones of similar types as well as earphones of the same type. This was an extension outside Romanow's original concept.

In 1945 the American Hearing Aid Association published a "Tentative code for measurement of performance of hearing aids" [7, Kranz et al, 1945]. The hearing aid microphone was to be placed in a free sound field of known level and the aid output was to be measured in the 6 cc or 2 cc coupler as appropriate. The ratio of output to input was the gain of the aid at the test frequency.

The contemporary literature contains much discussion of whether an aid should be mounted on a body-simulating baffle or whether the response of an aid on a coupler should be corrected for body baffle effects and/or aided-unaided ear differences [8, Carlisle and Mundel, 1944; 9, LeBel, 1944; 10, Hanson, 1944]. Many people used empirical data to convert objective coupler measurements to approximate subjective measurements so that coupler measurements could be used in aid dispensing [9], whilst others pointed out the inaccuracies of this [11, Nichols et al, 1945; 12, Martin and Anderson, 1947].

But before long the "Tentative Code" had become "the usual method" [13, Nichols et al, 1947] and was "in general use" [14, Corliss and Cook, 1948] and appeared to have superseded virtually all subjective testing in the USA. In the Harvard study, for example, [15, Davis et al, 1947] all responses were specified on the JRB 6 cc coupler. The accepted philosophy was that, once a basic response had been decided on as giving good speech discrimination, that response could be specified and maintained by a purely objective coupler calibration and that no further recourse to human subjects was necessary.

This view persisted and, despite accumulating evidence that the JRB couplers and real ears interacted differently with different receivers, these couplers with minor modifications became the standardised devices.
for earphone testing [16, ASA, 1949]. The American standards committee at the time, however, expected that the 2 cc coupler would have a useful life of no more than five years [17, Knowles, 1978].

In 1953 a procedure similar to the 1944 Tentative Code was adopted as the American Standard for hearing aid measurements [18, ASA, 1953]. As external receivers were replaced by insert receivers with earmoulds the 6 cc coupler became redundant and virtually all hearing aids were measured on the 2 cc coupler in free field. (N.B. the JRB 6 cc coupler was not used for audiometer calibration, a different type, the NBS 9A 6 cc coupler, was and still is used for this.)

During the fifties the American couplers were increasingly used in Europe. In Britain versions of West's artificial ear as modified by the NPL were used during the forties and fifties for aids with external receivers. These artificial ears could be used with insert receivers by fitting an adapter which reduced the cavity volume from 3 cc to 1.5 cc as described in the MRC 261 report [19, 1947]. But for insert receivers which were becoming the norm, the British ear was large and unwieldy and it was gradually displaced by the 2 cc coupler. Nevertheless it was still in use for hearing aid measurements as recently as 1958 [20, Morton, 1958].

The 2 cc coupler also supplanted the various national designs of coupler developed on the continent [e.g., 21, Chavasse and Lehmann, 1952], and in 1961 the IEC formally adopted the 2 cc coupler as the standard load for insert earphone calibration [22, IEC, 1961]. The coupler and the free field method of testing with all their inherent limitations have enjoyed a virtual monopoly in the field of hearing aids ever since. The coupler which the American standards committee thought would last five years is now in almost universal use thirty years later. Aid responses measured in free field on the 2 cc coupler are used for all applications from production testing and specification purposes to research and dispensing. Whilst the method is satisfactory for the first two applications it most certainly is not for the latter two.
2.2 CURRENT PROCEDURES AND THEIR LIMITATIONS

The current standardised methods of measuring the response of a hearing aid [23, IEC, 1959; 24, ANSI, 1960(1976); 25, BSI, 1968] are based on a free-field technique in which the microphone of the aid under test is placed in a plane progressive wave of known and constant sound level, and the output generated by the aid's receiver or earphone is measured in a standard hard-walled coupler.

The gain so measured, the ratio of the output sound pressure to the input sound pressure, or in decibels the difference between the output sound pressure level and the input sound pressure level, is a "transmission gain" [26, IEEE, 1972; 27, Collocott, 1971]. But the useful gain experienced by an aid wearer is an "insertion gain" [26;27] and is the ratio of the sound pressure at the eardrum when the aid is worn to the sound pressure at the eardrum with no aid worn, i.e., the ratio of the sound pressure in the "aided" ear to that in the "unaided" ear. This distinction is illustrated in Figure 2.5.

The transmission response on a coupler and the insertion response experienced by the wearer are quite different. When an aid is on a person the wearer's head and body modify the sound input to the hearing aid microphone. The input to the aid microphone depends upon the position of the microphone in relation to the wearer and upon the direction of sound incidence. Any directionality of the aid microphone is effectively modified too. Transmission gain measurements do not allow for these factors, insertion gain measurements include them automatically.

Furthermore there is no allowance in a transmission gain measurement for the change in earcanal acoustics from unaided to aided listening. This change is caused by occluding or partially occluding the earcanal with an earmould. An earcanal when unoccluded has a shallow resonance which gives an "open ear gain" over the range from about 1 kHz to 7 kHz which peaks around 3-4 kHz to 10 dB or more [28, Shaw, 1974]. But when the canal is occluded and partially filled with an earmould this resonance is destroyed. Consequently a hearing aid must compensate for the loss of the "open ear gain" in addition to overcoming any hearing loss.

The above limitations are a consequence of the difference between transmission and insertion response measurements and would apply even if the relevant standards were to specify a coupler which was a perfect model
of a typical or average real ear. But these limitations are compounded by measuring the aid output into the standard 2 cc coupler. This coupler was born of simplicity and convenience rather than of realism, and is adequate for its original intended purpose of comparative measurements on receivers or aids of the same type, though perhaps limited in bandwidth. When used outside of this well defined role it has severe limitations.

The shape of the coupler differs from that of a real ear. In addition the coupler has rigid walls whereas the boundaries of a real ear canal are more compliant. Consequently the sound level generated at the diaphragm of the coupler microphone by a hearing aid receiver will generally differ from the sound level at a real eardrum under otherwise similar conditions. The discrepancy will be greatest at higher frequencies: at lower frequencies (below say 800 Hz) the pressures measured in the coupler depend mainly on its volume.

More seriously two different types of earphone which produce the same sound pressure levels at a given frequency in a coupler may produce different sound levels from each other in a real ear. Figure 2.6 illustrates this with data from Nichols et al [13]. Because of this different interaction of receivers between ears and couplers a universal correction curve to convert a coupler measured response to an estimated wearer insertion response is ill advised. Nevertheless, such a conversion has often been suggested in the past [e.g. 9] and Richter and Diestel [29; 1973] argue that such a conversion is reasonably effective provided that there are no peaks or resonances in the response of the receiver, as resonances will differ in frequency as well as in level between the real ear and coupler.

The difference between coupler and real ear response varies not only with the type of receiver used but also with the various earmould treatments, such as venting. Figure 2.7 shows data from Studebaker and Zachman [30, 1970] which illustrates this aspect.

One further limitation of the 2 cc coupler, affecting its use for all purposes, including production testing, is that its useful bandwidth extends to approximately 4 kHz only [31, Eysbergen & Groen, 1959]. This is a consequence of its squat dimensions which were chosen to avoid resonances as far as possible. This restricted bandwidth may become an
increasingly acute problem as wider-band receivers and aids from the research labs come into common use.

2.2.1 Problems in comparing responses via the 2 cc coupler

The limitations of the 2 cc coupler have important repercussions when it is required to compare research findings from different sources. For example, it is possible to measure the response of a hearing aid on real persons by several different methods, as will be discussed below. But how do these methods compare and would they give the same responses under the same conditions?

To enable comparisons it is normal practice in research papers to include the response of an aid measured on a 2 cc coupler in addition to the response measured on a person. Whilst the sentiment is admirable the efficacy is questionable. Figures 2.8-2.13 show the results of some recent research projects [32-38]. Only studies published since 1970 are cited so that any techniques used should be beyond reasonable reproach or at least "state of the art". Also only data for headworn aids have been included.

Notice that each researcher has as usual confined himself to one and only one real ear method. Direct comparisons are therefore not possible between methods. Notice also that the only indirect means, in fact the only means, of comparing the different methods is via the 2 cc coupler response which is available in every case. This comparison is given in Figure 2.14. Clearly there are large differences between the various real ear responses when normalised to the standard 2 cc coupler response. But because of the limitations of the free-field/coupler method of testing explained above it is impossible to attribute these differences to any particular cause. They could be due to:-

(1) Different real ear methods giving inherently different responses.
(2) The real-ear-to-coupler sound pressure ratio differing with the type of aid receiver and with the particular earmould design employed.
(3) Differences between the real ears used in each study, e.g., with different pathologies, or merely between samples of a not very homogeneous population.
(4) Differences due to aid microphone position and directivity.
Or (5), and most likely, any combination of the above factors.

Since in general researchers use different hearing aids and different earmould tubing these factors cannot be disentangled.

The lesson from the above is that methods can only be compared directly. The free-field/coupler measurements are at best a floating standard and as such can provide no basis for reliable comparison of aids in use or methods of testing aids on a person. If real ear methods are to be compared they must involve the same hearing aids on the same people. This is the philosophy adopted during the work described in this thesis and in fact was one of the main reasons for the work.

2.3 SOME RECENT DEVELOPMENTS IN OBJECTIVE MEASUREMENTS

The early 1970's saw a renewed interest in hearing aid response measurements, particularly in objective measurements and improved couplers. It is too early for these new couplers to have become widely accepted: they are still effectively on trial or in limited use in research laboratories alongside accepted devices. The new couplers have also stimulated research into manikins or artificial heads as a means to measuring hearing aid insertion responses.

The new couplers and manikins are likely to be increasingly used in the future as they afford some considerable advantages over more traditional approaches and overcome some of the limitations of the 2 cc coupler. They also played a significant role in this research and consequently some detail is justified in their description.

2.3.1 New couplers (ear simulators)


Starting from a comprehensive collection of dimensional and impedance data on real ears, Zwislocki designed his "earlike" coupler as shown in Figure 2.15. The central cylindrical volume represents that part of the earcanal remaining when the average earmould is inserted in the average ear.
The half-inch condenser microphone diaphragm is reasonably similar in size to the average eardrum which it represents. Four damped Helmholtz resonators, each tuned to a different frequency lead off from the main earcanal cavity near the microphone, and these four "side branches" together produce an acoustic impedance next to the microphone which approximates the eardrum impedance of a median adult person over a wide frequency range. This part of the device is used for insert receiver measurements: for measurements of supra- and circum-aural earphones the earcanal is extended and a simple concha simulator is added.

The concept used by Zwislocki of side branches close to the microphone to produce the required impedance is basically that used by West in his 1930 GPO artificial ear, though the actual realisation is far more sophisticated and refined in the Zwislocki earlike coupler.

Sachs and Burkhard [41, 1972] state that sound pressures generated by insert earphones within a Zwislocki earlike coupler duplicate median sound pressure levels measured with probe microphones in real ears to within ±2 dB up to 7 kHz. Since Zwislocki produced his design there have been several attempts by other researchers to produce a similar performance from simpler designs. A theoretical study by Gardner and Hawley [42, 1973] suggests that both two- and four-branch couplers can give performance comparable to a real ear whilst one- and three-branch couplers are inherently poorer. A computer model by Zuercher and Burkhard confirmed these findings [43, 1972]. Practical designs of earlike coupler offering a performance similar to Zwislocki's include the PTB two-branch [44, Diestel, 1975], the IRPI XD 1053 two-branch [43, Zuercher and Burkhard, 1976] and the Shaw Unit-D device [45, Shaw, 1975] which has two resonators coming off the main canal with a further resonator off each of these. Bruel and Kjaer have also produced a tentative two-branch design, designated as Artificial Ear Type 4154 [46, B & K, 1977]. Figures 2.16-2.19 show the above earlike couplers.

These new earlike couplers are so radically different in concept and performance from the traditional hard-walled cavity that a new term "ear simulators" has been used to distinguish them from ordinary couplers or artificial ears. The following distinctions may be made:-
a coupler is a specified cavity which presents a standard load to an earphone under test. This load does not represent a real ear;

an artificial ear loads an earphone with an acoustic impedance which simulates the input impedance of an average ear. (It does not automatically follow that an artificial ear is similar to a real ear, nor that its microphone records the same sound level as an eardrum, but essentially, the earphone "looking in" cannot distinguish between an artificial ear and a real ear.)

an ear simulator, in addition to duplicating the input impedance of an average ear and thus correctly loading the earphone, also duplicates the transfer impedance to the eardrum. This implies certain geometric and acoustic similarities with a real ear and results in the measuring microphone at the eardrum position registering the sound pressure which would have been present at the drum of a real ear.

An IEC working group is currently working on performance standards to define artificial ears. All the above mentioned specific designs are likely to conform to these standards.

As well as new ear simulators there has been a design for a new coupler, Keller's conical coupler [47, 1974] shown in Figure 2.20. This is not an ear simulator but it does have an advantage over the 2 cc coupler of having a bandwidth extending to above 20 kHz, otherwise its limitations are the same as those of the 2 cc coupler. It is however far simpler and cheaper than a proper ear simulator.

The new ear simulators may be used in place of the 2 cc coupler to measure aid transmission responses. This gives a more realistic load on the aid receiver, enables different earmould designs including venting to be compared in a more valid manner, and gives an extended bandwidth. On the other hand it still does not allow directional microphone responses to be compared.

2.3.2 Manikins for hearing aid measurements

Perhaps the greatest advantage afforded by ear simulators is that both aided and unaided ears can be equally well simulated. This means
that they can be incorporated in anthropometric manikins or artificial heads. This in turn provides the opportunity of measuring in a repeatable, objective, but realistic way, the \textit{insertion} response of a hearing aid.

Manikins are not new. But those constructed in the past [e.g., 48, Bauer et al, 1967; 49, Kasten and Lottermann, 1967] whilst suitable for some comparative or directivity measurements on aids, have not been suitable for insertion gain measurements. This is because the available ear simulating devices have not been equally suited to aided and unaided listening.

The appearance of the Zwislocki ear simulator provided the first opportunity for designing a realistic manikin. Burkhard took this opportunity to design the manikin 'Kemar' (the Knowles Electronics Manikin for Acoustic Research). Considerable effort ensured that the dimensions of Kemar and the design of his external ears matched available and specially collected anthropometric data. Kemar is therefore a median-dimensional adult individual, with Zwislocki ear simulators making up the inner portions of his/her earcanals and with half-inch microphones as eardrums.

The enormous degree of success achieved is apparent from the extensive documentation comparing Kemar's acoustic performance with published human data [50–53, Burkhard and others, 1974–1977].

2.4 SOME RECENT DEVELOPMENTS IN SUBJECTIVE MEASUREMENTS

The appearance of the new ear simulators also renewed interest in real ear measurement methods. The probe microphone work of Dalsgaard and his collaborators, the threshold shift methods of Pascoe and his colleagues and the acoustic reflex methods of Tonisson are prominent, and have already been cited in section 2.2.1. However, as explained in that section, there is no method by which their various results can be compared or evaluated.
2.5 POSSIBLE METHODS FOR THE MEASUREMENT OF AID RESPONSE

With the latest developments described above there now exist several practical or theoretically possible methods of measuring the gain of an aid at a given frequency and which thus enable the response of an aid to be determined. These methods divide broadly into those giving a "transmission response" and those giving an "insertion response", with further subdivisions possible within these two categories. Figure 2.21 lists the various methods classified hierarchically to illustrate their similarities and differences. From the left to the right of this figure the methods get progressively more complex, progressively more realistic and approach more nearly the total response of an aid on an individual person.

Figure 2.21 effectively summarises and consolidates the discussion of this chapter, and serves as a convenient starting point in choosing the methods of measuring responses to be used in this research. This aspect is described in the next chapter.
FIG. 2.1 WEST'S GPO ARTIFICIAL EAR, 1930

to 8ft tube containing strands of wool

1/4 in \( \Phi \)  

\(- \_20 \_\)  

\( \theta \)  

silk gauze

ebonite block

condenser transmitter

diaphragm

dimensions in mm except where stated

FIG. 2.2 INGLIS, GRAY & JENKIN'S ARTIFICIAL EAR, 1932

acoustic leak

rubber seating

condenser transmitter

acoustic network
FIG. 2.3 LITTLE'S APPARATUS FOR THRESHOLD MEASUREMENTS OF HEARING AID RESPONSE, 1936.

FIG. 2.4 ROMANOW'S DESIGNS FOR 2 cc AND 6 cc COUPLERS, 1942.
transmission gain on coupler = \frac{P_3}{P_r}
transmission gain on real ear = \frac{P_s}{P_d}
insertion gain on real ear = \frac{P}{P_c}

FIG. 2.5 INSERTION AND TRANSMISSION GAINS COMPARED
FIG. 2.6  REAL EAR vs. COUPLER RESPONSE FOR VARIOUS RECEIVERS [after Nichols et al 1946]

FIG. 2.7  REAL EAR vs. 2 cc COUPLER RESPONSE WITH VARIOUS EARMOULD VENTS [after Studebaker & Zachman 1970]
FIG. 2.8 RESPONSE OF AID ON 2cc COUPLER AND ON REAL EARS BY LOUDNESS BALANCE [after Starke, 32, 1970]

FIG. 2.9 RESPONSE OF AN AID ON 2cc COUPLER AND ON REAL EARS BY LOUDNESS BALANCE [after Lower, 33, 1975]
FIG. 2.10 RESPONSE OF AN AID ON 2 cc COUPLER AND ON REAL EARS BY THRESHOLD METHOD [after Pascoe, 34, 1975]

FIG. 2.11 RESPONSE OF AN AID ON 2 cc COUPLER AND ON REAL EARS BY ACOUSTIC REFLEX THRESHOLD SHIFT [after Tonisson, 35, 1975]
FIG. 2.12 RESPONSES OF 2 AIDS ON 2 cc COUPLER AND ON REAL EARS MEASURED WITH A PROBE MICROPHONE [after Dalsgaard & Jensen, 36, 1976]

FIG. 2.13 RESPONSE OF AN AID ON 2cc COUPLER AND ON REAL EARS BY ACOUSTIC REFLEX THRESHOLD SHIFT [after Pizarro, 37, 1976]
FIG. 2.14 PUBLISHED REAL EAR MEASUREMENTS OF HEARING AID INSERTION RESPONSES, COMPARED VIA THEIR RESPECTIVE 2 cc COUPLER TRANSMISSION RESPONSES. see also figures 2.8 – 2.13
FIG. 2.15  THE ZWISLOCKI EAR SIMULATOR; LONGITUDINAL SECTION AND BOTTOM ASPECT.
See also Figure 3.4 for photograph.

FIG. 2.16  THE PTB EAR SIMULATOR (2-BRANCH)
FIG. 2.17 THE XD-1053 TWO BRANCH EAR SIMULATOR OF INDUSTRIAL RESEARCH PRODUCTS, INC.

FIG. 2.18 SHAW'S UNIT-D EAR SIMULATOR
FIG. 2.19 TENTATIVE B&K DESIGN FOR TYPE 4154 ARTIFICIAL EAR

FIG. 2.20 KELLER'S CONICAL COUPLER.
FIG. 2.21 A CLASSIFICATION OF METHODS OF MEASURING HEARING AID RESPONSES
CHAPTER 3

METHODS, SUBJECTS AND FACILITIES

3.1 INTRODUCTION

At the outset of this study it was apparent that for any serious hearing aid research the standard method of measuring aid response would be inadequate and that it would be necessary to be able to measure the response of an aid on real ears. The choice and development of a suitable real ear method was therefore of great importance, and this is described below.

What few published data were available suggested that a Zwislocki ear simulator or a Kemar manikin, both of which were accessible, might obviate the need for real ear measurements for some purposes. It was initially decided therefore to evaluate the Zwislocki ear simulator and Kemar against the real ear measurements and against the standard method using the 2 cc coupler.

The outcome of this first experiment suggested that far more fundamental research was needed into hearing aid response measurements and that response measurements were less well understood generally than was previously assumed. Consequently the rest of the project grew from the necessity to explain the results of this first experiment.

The first experiment is described in detail in the next chapter but many procedures and facilities were common to most of the experiments and are described below.

3.2 CHOICE OF A METHOD FOR MEASURING THE REAL EAR RESPONSE OF A HEARING AID

The method chosen for measuring the responses of aids on real persons was an aided versus unaided alternate binaural loudness balance by oto-logically and audiometrically normal subjects. The reasons for this initial choice are outlined below.

During the course of the research it was discovered that the combination of method, subjects and the hearing aids used caused some problems. However, there was no cause to suspect this at the outset.
3.3 THE LOUDNESS BALANCE METHOD

3.3.1 Reasons for choice

An alternate binaural loudness balance method was chosen because:

1) it would include the total response of the aided and unaided ear (previous comparisons of the Zwislocki ear simulator with real ears used probe microphones between the earmould and the eardrum, involving assumptions relating pressures in the ear canal to pressures at the drum and assumptions that such relationships hold for both aided and unaided listening);

2) it would allow supra-threshold measurements at comfortable levels, with options on different levels, and at levels well above the noise floor set by the aid;

3) it would allow the use of normally hearing subjects (a threshold shift, for example, would give a masked threshold for aided listening by normal subjects and the true gain of the aid would not be found);

4) it could be used with any design of earmould which fully occluded the ear, no special arrangements being required.

3.3.2 Outline of the method

The rationale behind the method is summarised in Figure 3.1.

Electronically generated pulsed pure tone reference stimuli were presented to one ear of a subject at a preset level via a hearing aid receiver and earmould. In the other ear the subject alternately wore the hearing aid or no aid at all, in either case adjusting the level of pulsed pure tones heard in this ear from a loudspeaker to match the fixed signal loudness in the reference ear. The pure tone frequency from the speaker was the same as the reference. In general the gain of an aid would be greater than unity, and the output from the loudspeaker would be reduced when the aid was worn to produce the same loudness as when no aid was worn.

The ratio of the loudspeaker driving voltages for the two balances (i.e., aided and unaided) was taken to be equal to the gain of the hearing aid as perceived by the wearer at the test frequency. This ratio was expressed in decibels and is referred to throughout this thesis as the "subjective gain". By repeating the test at various discrete frequencies a "subjective frequency response" was built up.
The pulses of sound from the reference receiver and the loudspeaker alternated and the subject heard each channel effectively monaurally. Thus there was no problem with relative signal phases as might have occurred had the tone in each channel been continuous and heard binaurally.

The exact level of the reference signal need not have been known exactly but had to remain constant, then any difference in the reference signal level from its nominal value would cancel between aided and unaided balances. Similarly, any difference in sensitivity between a subject's two ears was assumed to remain constant throughout each experimental session and this would also cancel in the calculation of the hearing aid gain.

Loudest balance techniques for measuring hearing aid gain were amongst the earliest accepted methods [e.g., 1, Romanow, 1942; 2, Littler, 1944]. The technique adopted here improved on the traditional method with the addition of the reference channel in the non-test ear. The reference signal was effectively an aid to the memory of the subject and helped to avoid the complications involved in a direct monaural comparison such as described by Littler ....

"A listener listens to the sound from the telephone and then quickly removes it to listen to the source in free air, without altering the position of his head. While so doing he quickly alters the reading of the attenuator to make the two sounds appear equally loud; in removing the telephone he covers the telephone orifice to prevent howling due to acoustic feedback."

The technique adopted with the reference signal to the non-test ear was based very loosely on a German standard for measuring the response of headphones [3, 1973]; a variation of this standard has been used by Starke [4, 1970] with hearing aids in an attempt to derive corrections for responses measured on the 2 cc coupler, whilst a similar technique has been proposed by the South African Bureau of Standards for measuring hearing protector attenuation [5, Meij et al, 1974].

3.3.3 The loudness balance audiometer

The tone pulses for the loudness balance were generated by a specially built loudness balance audiometer. A schematic diagram of the audiometer is given as Figure 3.2.
The audiometer generated two channels of pulses in the sequence shown in Figure 3.3. Pulses were 600 ms long and transient free with rise and fall times of 35 ms. The frequency of the tone was selected by switched preset controls, the presets being continuously adjustable over the range 180 Hz to 16 kHz.

The two channels of the audiometer were identical. An attenuator in each channel enabled the operator to control the output level of each channel. A further attenuator was controlled by the subject in the test room and this attenuator could be switched into either channel by the operator. The subject's attenuator was then in series with the operator's and the subject could vary the level of that channel from inaudibility up to the level set by the operator on the operator's control. The subject's attenuator was the "fader" type (Penny & Giles Ltd., type 1820, 10 kΩ log) as used on mixing desks in recording or broadcast studios and operated according to a logarithmic law. Moving the attenuator slider from the top of its range caused a change in signal level of approximately 10 dB for each 20 mm of travel down to 80 mm (40 dB) from the top of the range. Thereafter the range was compressed such that over the remaining 15 mm of travel the signal was reduced to zero.

In practice one of the two identical channels was designated the reference channel and its output was fed to the reference receiver in a hearing aid shell. The acoustic signals from the receiver were then fed via an earmould (3.5) to the subject's reference ear. The reference level was set directly by the experimenter.

The output of the second channel was generally fed via a power amplifier to a loudspeaker. The subject's attenuator was switched into this channel so that the subject had control of the level from the loudspeaker. The operator set his control such that the subject would find that equal loudness between the loudspeaker and the reference channel would occur within the range of the subject's attenuator. In setting his channel the operator had to allow for the gain of any hearing aid worn.

The audiometer provided an output to a voltmeter which could be switched into either channel. An acoustical calibration of the loudspeaker and reference receiver and an electrical calibration at low signal levels showed the output of the audiometer to vary linearly with the voltmeter reading over its entire range. Harmonic distortion in the audiometer output was approximately 0.6%.
The reference receiver was an experimental wideband design, a Knowles Electronics XD 1005, with a good response to beyond 10 kHz (Figure A.2). Bandwidth had been obtained at the expense of efficiency in this receiver and although it was capable of generating 120 dB into a 2 cc coupler, it required about 30 V rms to do so.

3.3.4 Accuracy of loudness balancing

A pilot experiment carried out to test the loudness balance audiometer and to determine the inherent accuracy of loudness balances is described in Appendix A. The loudness balance method used was shown to give consistent results with an average error of less than ±1 dB over the frequency range 500 Hz–5 kHz, and less than ±2 dB at 250 Hz and 10 kHz for reference sound pressure levels of 60 dB SPL and 88 dB SPL (measured in a Zwislocki ear simulator).

3.4 SUBJECTS

All subjects taking part in this study were otologically normal as defined in BS 5108: 1974 [6, BSI, 1974] and ISO R389-1964 [7, ISO, 1964] with hearing levels not exceeding 20 dB re: ISO R389 in the frequency range from 250 Hz to 6 kHz.

The subjects, most of whom were staff or students of the university aged between 20 and 30 years, were volunteers paid at normal ISVR permitted rates based on the duration of each experiment.

The decision was made to use normal subjects with healthy ears rather than hearing impaired subjects typical of aid wearers because:

1. Normal subjects would provide a better defined, more homogeneous group than hearing impaired subjects.

2. Consequently results were likely to be more consistent and would not be confounded with any particular pathology.

3. The Zwislocki ear simulator and Kemar simulate healthy ears with normal impedances (the only repeatable case) – a test against healthy ears is therefore fairer.

4. The results of different methods would be easier to relate if all possible unknown variables were eliminated: if hearing impaired
were originally chosen any variation in results could not be separated between method and subjects.

The study could have incorporated hearing impaired subjects at a later date. The reason it did not, as will become apparent, was that the experiments using normal subjects gave unexpected results and the research was directed towards explaining these results.

3.5 AIDS AND EARMOULD

Experiments were confined to behind the ear aids with downward or forward facing microphones. The majority of NHS aids and over 90% of aids sold privately in the UK are now of this type [see, e.g., 8, Price Commission, 1977].

Individual earmoulds were made for both ears of each subject. After an otological examination, an impression of each ear was made by the experimenter using "Otoform K" earmould impression material. A local hearing aid dispenser had the moulds made up by his usual earmould laboratory.

These final moulds were in acrylic and completely filled the concha of the subjects: skeleton moulds were not used since the best possible fit, the best seal, and the maximum sound attenuation were required. Each mould penetrated about 10 to 12 mm into the earcanal. The exact depth of penetration was left to the skill and discretion of the earmould laboratory as would normally be the practice. Any variation in depth of penetration was regarded as part of the intersubject variation in the size of earcanals, which in any case varied considerably. Furthermore, any attempt to artificially control the mould to eardrum distance or volume was deemed unnecessary since the ear simulator with which the real ears were to be compared was claimed to simulate the median of actually occurring mould to drum distances and volumes, assuming normal designs of earmoulds. The comparison between the ear simulator and real ears would therefore be more realistic if depth of earmould penetration were not controlled.

The earmould tubing dimensions in contrast were rigidly controlled and were identical for all earmoulds; the tubing and aid receiver together being regarded as a unit acoustic source. Furthermore, the same tubing dimensions were used for coupler and ear simulator measurements to enable direct comparisons with the real ear measurements.
The tubing was made of soft plastic and had an internal diameter of 2 mm. It ended flush with the earmould tip in the ear canal and was cut to a length of slightly more than 40 mm so that, when the tubing was pushed over the hearing aid hook, 40 mm of tubing remained between the end of the hook and the earmould tip. This length of tubing was a good fit for the subjects and also had the advantage, when used with a coupler (3.6.2), of conforming to the tubing arrangements shown in figure 2 of BS 3171: 1968 [9, BSI, 1968]. Whilst this is not the tubing arrangement recommended by the IEC [10, 1959] or ANSI [11, 1960(1976)], a variation from the recommendation is allowed in these standards. Measurements on the 2 cc coupler therefore conformed to the IEC, British and American standards.

During later experiments special earmoulds incorporating microphone probe tubes were used. These are described in Chapter 7 (7.3.2 and Figure 7.3) when they are first used.

In order to improve the sealing of an earmould to an ear during experiments a small smear of vaseline was applied to the mould. This was to prevent a low frequency roll-off of the hearing aid or reference receiver response due to leakage and also increased the earmould attenuation of the external sound field by a few decibels. The latter factor is important during loudness matching. The attenuation of the external sound field by the normal and the special probe-tube earmoulds was measured with and without the vaseline in a subsidiary experiment described in Appendix B.

3.6 MEASUREMENTS ON COUPLER AND EAR SIMULATORS

3.6.1 Couplers and ear simulators used

One 2 cc coupler and one Zwislocki-type ear simulator were used throughout this research to provide objective and repeatable response measurements of all the aids used. An additional Zwislocki-type ear simulator and an IRPI XD 1053 two-branch simulator were acquired during the study. The ear simulators are shown in Figure 3.4.

The 2 cc coupler was a B & K type DB 0138 used with a B & K 4144 one-inch pressure microphone. The Zwislocki ear simulator used throughout was type XD 960 manufactured by Industrial Research Products, Inc. (IRPI). The second Zwislocki ear simulator was also made by IRPI, type DB 100, but differed slightly from the other to make it more easily constructed and more stable. Measured responses were virtually indistinguishable between the two Zwislocki-type ear simulators.

25.
The IRPI XD-1053 is a two-branch ear simulator designed to give responses as close as possible to the Zwislocki four-branch simulators but at a fraction of the cost since it is a much simpler device. Its construction was shown in Figure 2.17 of the previous chapter.

All the ear simulators were used with the same recommended B & K 4134 half-inch pressure microphone.

3.6.2 Tubing and couplings

The tubing used to connect aids to couplers and ear simulators had the same dimensions as the tubing through the earmoulds used in the real ear measurements, i.e., 40 mm × 2 mm inside diameter. Earmould simulators were used as illustrated for the Zwislocki ear simulator in Figure 3.5. The same earmould simulator was used with the XD-1053 ear simulator whilst the B & K earmould simulator supplied as standard was used with the 2 cc coupler.

3.6.3 Method

All response measurements were made in a B & K 4212 Hearing Aid Test Box in accordance with BS 3171: 1968 [9]. The rest of the equipment is shown in Figure 3.6. The sound level at the aid microphone position was measured in the absence of an aid so that minor input level corrections could be made to the measured response. All measurement microphones used were calibrated against the same B & K 4220 pistonphone.

3.7 THE MANIKIN KEMAR

The manikin Kemar is a median dimensioned head and torso as shown in Figure 3.7. The Zwislocki ear simulator incorporated can clearly be seen in this photo. The only dimensional difference between men and women which is acoustically significant, though small in its effect, is the length of neck. Kemar's neck length is adjustable to give a median male length, a median female length or the average of male and female medians. The figure shows Kemar with two extension rings in place in his neck, which gives the median male length. For all measurements described in this study, Kemar was dressed in a tee-shirt to control the reflection of sound from the shoulders.
3.8 ANECHOIC ROOM

All subjective and many objective measurements were carried out in an anechoic room (Room 1.24, Rayleigh Building, ISVR).

This room has inner walls of brick, 11.5 cm thick, isolated by a 25 mm air gap from the outer structural walls. The internal dimensions, wall to wall, are 2.27 x 5.36 x 2.75 metres (w x l x h). The inner surfaces of the room are covered with coated polyurethane wedges 31 cm long. Areas of floor necessary for equipment or access were covered with sheets of absorbent foam rather than wedges. The volume of free space in the room is 20.1 m$^3$. The room has a double door forming a lead-wood sandwich.

The background noise level in the room, Figure 3.8, is below the limits specified in BS 5108:1974 [6], ASA Z24.22-1957 [12, ASA, 1957] and ANSI S3.19-1974 [13, ANSI, 1974] at frequencies at which measurements were not limited by the test equipment noise floor, and therefore below threshold for subjects with normal hearing levels at these frequencies. Threshold determinations for normally hearing subjects confirmed that ambient levels in the room were sufficiently low not to mask normal thresholds at any frequency tested (B.5.1).

The room is anechoic at frequencies above 200 Hz; Figure 3.9 shows that deviation from free-field conditions in the room is extremely small at 250 Hz and above.

3.9 OTHER EQUIPMENT

Other equipment used varied from experiment to experiment and is described in the appropriate chapter.
Definition: "subjective gain" = $20 \log_{10}(\frac{V_1}{V_2})$; dB

FIG.3.1 PHILOSOPHY BEHIND THE LOUDNESS BALANCE METHOD OF MEASURING HEARING AID GAIN
FIG. 3.2 SCHEMATIC DIAGRAM OF THE LOUDNESS BALANCE AUDIOMETER

4 ganged switches determine which channel is under subject's control.

FIG. 3.3 STIMULUS PRESENTATION SEQUENCE

Pulse duration 500 ms
Pattern repeats every 15 s
FIG. 3.4 THE EAR SIMULATORS USED IN THIS STUDY.
FIG. 3.5 CONNECTING AN AID TO A COUPLER OR EAR SIMULATOR; TUBING DIMENSIONS USED.

FIG. 3.6 EQUIPMENT USED TO MEASURE TRANSMISSION RESPONSES OF AIDS ON COUPLER OR EAR SIMULATORS.
Object: Noise floor in "Blue Room" (R 1.25) (electrical-acoustic) Date: 29/11/76 Range: 0 to 120

FIG. 3.8 MEASURED BACKGROUND NOISE IN TEST ROOM (PLUS THE ELECTRICAL NOISE OF THE EQUIPMENT USED TO MEASURE IT)

FIG. 3.9 COMPARISON OF SOUND FIELD IN ROOM WITH IDEAL FREE FIELD CONDITIONS
CHAPTER 4

EXPERIMENTAL MEASUREMENTS OF HEARING AID GAIN BY LOUDNESS BALANCE,
2cc COUPLER, ZWISLOCKI COUPLER AND MANIKIN

4.1 INTRODUCTION

The specific aims of this experiment were to develop a method of measuring the response of an aid worn normally on real ears; evaluate the Zwislocki ear simulator and the 2cc coupler against real ears; evaluate Kemar against real persons and to establish any relationships between the responses measured in these different ways in order to determine whether predictions are possible of one response from another.

4.2 METHODS

Three separate procedures were used to determine the response of a behind-the-ear hearing aid. These were (i) the subjective loudness balance technique, (ii) free-field measurements with Zwislocki and IEC 2cc couplers, and (iii) measurements on Kemar.

The same hearing aid, a Widex Baritone 641H with a downward facing microphone as shown in Figure 4.1 was used in all 3 parts of the experiment. The volume control was set to give a slight to moderate gain and then firmly sealed throughout the experiments.

4.2.1 Loudness balance measurements
4.2.1.1 Experimental design

Based on the findings of a trial loudness balance described in Appendix A, a fully counter-balanced design was evolved requiring eight subjects, to make aided and unaided loudness balances at eight frequencies, 250 Hz, 500 Hz, 1 kHz, 1.5 kHz, 2 kHz, 3 kHz, 4 kHz and 6 kHz, and for two directions of sound incidence, i.e., with the loudspeaker directly in front of (0°) or directly behind (180°) the subject. One reference signal was used, this being 70 dB SPL as measured in a Zwislocki ear simulator. This level was chosen to minimise the small but systematic errors described in Appendix A.

Each subject took part in four sessions, two with the loudspeaker in
front of him or her and two sessions with the loudspeaker behind. In each session all eight frequencies were presented for the subject both aided and unaided - in half the sessions all the aided balances were completed first as a group and in the other sessions the balances without the aid were first. The reference signal was presented to the right ear of four subjects and to the left ear of the other four to balance out any asymmetry in the listening room. The presentation order of the test frequencies was balanced over the whole experiment using a latin square design as shown in Table 4.1.

4.2.1.2 Subjects

Eight subjects conforming to section 3.4 took part. Five were male, three female. All had previously taken part in the experiment described in Appendix A.

4.2.1.3 Equipment and facilities

The equipment used is shown in Figure 4.2. The subject was seated in the anechoic room 2m from the loudspeaker with the high frequency unit at ear level.

One channel of the audiometer was connected to the wideband reference receiver mounted in a hearing aid shell and the pulses of sound transmitted at a preset level to the subject's ear through an earmould. The other channel was connected through a power amplifier to the loudspeaker, the level of this channel being controlled directly by the subject. Total harmonic distortion in the power amplifier and loudspeaker together was measured as less than 1% at 1 kHz and 70 dB SPL. The SPL at the centre-of-the-head position in the room varied linearly with the voltmeter reading of the audiometer output over the entire range used in the experiment.

4.2.1.4 Procedure

The nature of the experiment was explained to the subject at his first session and he was given the following written instructions:
INSTRUCTIONS

We would like you to help us evaluate a hearing aid using the loudness balance technique we tested in the previous experiment.

This experiment is slightly different from the last one in that you will hear all the constant tones in one ear, the same one throughout, while the adjustable tones will be presented to your other ear by the loudspeaker in front of, or behind, you; for part of the experiment we will ask you to wear a hearing aid.

However, all you need remember is:-

Move the sliding knob towards the yellow light whenever it comes on, as far as it will go.

Whether or not you are wearing a hearing aid, adjust the slider control until the tones you hear are equally loud in each ear. When you are satisfied the tones are equally loud press the "task complete" button.

Please try to keep your head still during the experiment, and face directly towards or away from the loudspeaker.

Thank you.

The intercom will be on all the time.

(Note: the "previous experiment" is that described in Appendix A.)

The subject sat facing towards or away from the loudspeaker and completed a total of 16 separate loudness balances against the reference signal with, or without the hearing aid in the test ear in the sequence determined by Table 4.1.

No head rest or clamp was used to avoid interfering with the sound field around the head, instead the subject was asked to keep his head still and look directly ahead while tones were being presented.

The audiometer controls were set so that the point of balance on the subject's control scale was never in the same place twice and was randomised over the scale but avoiding the extremes.

To eliminate any bias the subject started alternate balances from opposite ends of his control as indicated by signal lights. When the
subject had finished each balance he signalled with a buzzer, and the 
audiometer output voltage to the loudspeaker was recorded.

A complete session lasted between 25 and 50 minutes, with most 
subjects taking about 25–30 minutes.

4.2.2 Measurements on coupler and ear simulator

The response of the aid was measured on a 2 cc coupler and a Zwislocki 
ear simulator (3.6).

4.2.3 Measurements on "Kemar"

A constant level free-field frequency sweep up to 9 kHz was established 
in the anechoic room by replaying the tape recorded output of a beat 
frequency oscillator using the method described by Wansdronk [1, 1959]. 
Kemar was then seated in the room facing the sound source and the SPL at 
his "eardrum", the microphone diaphragm of his Zwislocki ear simulator, 
was graphically recorded over the frequency sweep, with and without the 
hearing aid worn. The unaided curve was subtracted from the aided to 
obtain the insertion response of the aid.

The procedure was repeated with Kemar facing directly away from the 
source. The measurement conditions were therefore directly comparable 
to those of the loudness balance experiments.

A plastic eartip, adapted from a Pacific Plantronics size No. 3 eartip 
by replacing the tubing, was used as an earmould for Kemar. Removing 
and replacing the earmould made less than a decibel difference to the 
measurements of sound pressure in Kemar's ear.

4.3 RESULTS

4.3.1 Loudness balance measurements

The mean subjective response of the aid (3.3.2) is plotted in Figure 
4.3 for each direction of sound incidence together with the 95% confi-
dence intervals. Each plotted point is the mean of 16 values (eight 
subjects twice).
4.3.2 Objective measurements

Figure 4.4 shows the transmission responses of the hearing aid on the 2 cc coupler and the Zwislocki ear simulator; Figure 4.5 shows the insertion responses of the aid on Kemar for front and rear sound incidence.

4.4 DISCUSSION

Both the real ear and the Kemar measurements show that the aid's insertion response varies with the direction of sound incidence. The gain of the aid is greater for sound from behind the wearer. This supports complaints from wearers of behind-the-ear aids with downward facing microphones that sounds from behind are heard better than sounds from in front. Reverberation from behind is amplified more than direct sound from the front thereby reducing intelligibility of face-to-face conversation.

Figure 4.4 shows the transmission responses on the 2 cc coupler and the Zwislocki ear simulator. Differences between them follow similar trends to those measured by Sachs and Burkhard [2; 1972]; a small roughly constant difference up to about 1 kHz with the SPL in the ear simulator greater than that in the coupler, followed by a divergence above this frequency.

Comparing the objective measurements with the subjective shows that over most of the aid's bandwidth the mean subjective gain is considerably less than the transmission gain measured on either the 2 cc coupler or the Zwislocki ear simulator - up to 11 dB less at 1 kHz than that measured on the coupler and up to 25 dB less at 1.5 -2 kHz than that on the ear simulator. These differences are plotted in Figure 4.6. Note that the transmission response on the ear simulator is further than the transmission response on the coupler from the mean subjective (insertion) response, even though the ear simulator will be a more accurate model of the average real ear.

The comparison between the transmission gain on a coupler or ear simulator and an insertion gain on real ears is valid in its own right as demonstrating that the two quantities are quite different and should not be confused in aid research or dispensing. However, to compare

32.
fairly the performance of an ear simulator with that of the real aided ear which it models then like must be compared with like; \textit{transmission} gains on each must be compared.

Referring back to Figure 2.5 it can be seen that the transmission gain of an aid on a real ear, $P_e/P_d$, can be derived from the insertion gain, $P_e/P_c$, if the pressure transfer ratio $P_d/P_c$ is known. This pressure transfer ratio is simply that between the aid microphone position and the eardrum of the \textit{unaided} ear. Though this transfer ratio is not available directly in any published literature it can be calculated. It is the product of the pressure transfer ratio from the aid microphone position to the unoccluded ear canal entrance and the pressure transfer ratio from the unoccluded ear canal entrance to the eardrum. Measured values of both transfer ratios have been published.

Table 4.2 gives the values of the pressure transfer ratio from a typical location of a downward facing aid microphone to the unoccluded ear canal entrance as measured by Olsen and Carhart [3, 1975]. This pressure transfer ratio differs with the direction of sound incidence and accordingly values are given in the table for sound incident from both the front and back. Table 4.3 presents the pressure transfer ratio from the ear canal entrance to the eardrum as measured by Wiener and Ross [4, 1946] and presented by Wiener [5, 1947]. According to Wiener and Ross these figures are independent of the direction of sound incidence.

So using Olsen and Carhart's and Wiener and Ross' data the transmission response of the aid on the average real ear may be estimated from its pressure insertion response, or the sound pressures existing at the aided and unaided eardrum. To estimate the real ear transmission response of the aid from the subjective gain measured here requires the assumption that the loudness elicited by a given sound pressure level at the eardrum is the same whether the ear is aided or unaided.

Figure 4.7 shows the effect of adding Olsen and Carhart's measurements on to the subjective response of the aid and Figure 4.8 shows the effect of adding Wiener and Ross' data as well. In both these figures the effect of sound diffraction around the head has been subtracted out. The curves for the two directions of sound incidence should therefore coincide. The two curves are in fact very close, within 3.5 dB from
250 Hz to 4 kHz.

What Figure 4.8 shows is an estimate from the aid's subjective response of the aid's transmission gain on the average real ear and therefore on an ideal simulator. Figure 4.9 illustrates the large differences between this estimated ideal response and actual responses on the 2 cc coupler and the Zwischen-Aar-Korn ear simulator. But how accurate is this estimated ideal response?

To test the validity of adding the head diffraction and ear canal resonance data to the subjective data to arrive at an estimate of the real ear transmission response, the same procedure was applied to the insertion response of the aid as measured on Kemar.

Kemar's eardrum and internal part of the ear canal is a Zwischen-Aar-Korn ear simulator. Therefore if the procedure adopted for converting an insertion gain to a transmission gain is applied to the aid response measured on Kemar, the result should be the same as the response of the aid measured on the Zwischen-Aar-Korn ear simulator in the aid test box.

Figure 4.10 shows the effect of adding the human head baffle and ear canal data to the aid response measured on Kemar, and comparison with Figure 4.4 shows a remarkably accurate agreement with the Zwischen-Aar-Korn measurements in the test box, marred only by slight irregularities around 3 kHz and 4 kHz. The agreement is within 2 dB at all frequencies up to 1 kHz and within 3 dB up to 2 kHz. The minor irregularities at 3 kHz - 4 kHz are due to slight irregularities in Kemar's ear canal resonances not apparent in the real ear data which is smoothed by averaging over subjects. This good agreement implies firstly that the procedure of adding the head baffle and ear canal effects to the insertion response to obtain the transmission response is both valid and complete, and secondly that Kemar accurately models the head diffraction of real subjects, at least for aids of this geometry with downward facing microphones.

Nevertheless, the prediction in Figure 4.8 from the subjective real ear data of what the response on an ideal ear simulator should be, although similar to the response measured on the Zwischen-Aar-Korn ear simulator, is not the same as the Zwischen-Aar-Korn curve, even though Sachs and Burkhard [6, 1972] have shown that the sound pressure level in this simulator will be the same as in a real average ear. The discrepancy is shown in Figure
4.9: the gain measured in the Zwislocki ear simulator is greater than that predicted from the loudness balance experiments.

A similar discrepancy was found progressively below 1 kHz by Richter and Diestel [7, 1973]. They used a loudness balance method to compare an electrically driven insert earphone with a conventional earphone with a known free-field sensitivity level as defined by DIN 45619 [8, 1973] and compared these subjective data with measurements using a German-built Zwislocki-type ear simulator.

Allowances were made for earcanal resonances in much the same way as described above, though they did not have to contend with hearing aid microphone locations. The Zwislocki simulator gave a greater gain measurement than would be expected from the subjective data by 6 dB or so at 500 Hz and 10 dB at 250 Hz. Richter and Diestel attributed this to variations in the fitting of the insert earphone and the possibility of acoustic leaks from the ear to the outside.

4.4.1 Earmould leakage

A vented or a loosely fitting earmould will reduce the low frequency gain of a hearing aid, but the precaution of smearing earmoulds lightly with vaseline would have ensured that any such effect would be minimal in this experiment. In any case, the frequency range and magnitude of the discrepancy between the aid transmission gain predicted for real ears and the aid gain on the Zwislocki ear simulator suggest that the discrepancy is not due to this effect.

However, more subtle effects of earmould leakage can occur, including the leakage of the external sound field generated by the loudspeaker into the aided ear and its mixing with the output of the hearing aid, and the leakage of the external sound field pulses into the reference ear in the quiet periods between reference pulses.

Sound leaking into the reference ear is unlikely to cause any problem. The loudness of the reference tones will be unaffected since the reference pulses do not occur at the same time as the loudspeaker pulses. If sound were to leak to the reference ear during an unaided balance so that the loudspeaker is heard binaurally instead of monaurally then the loudspeaker

35.

Note: Vaseline is a registered trade-mark: its initial letter should be upper case throughout.
might sound too loud due to the binaural advantage. However, in the worst case of the reference earmould providing no attenuation whatsoever this binaural advantage would be only 2 dB to 3 dB. Even a few decibels of attenuation by the reference earmould would reduce the binaural advantage to virtually nil. Since it would be an extremely poor earmould that could not provide 5 dB of attenuation when smeared with vaseline, this is not considered a problem. During aided listening, provided the aid has a gain greater than unity, the external sound field will be reduced by the subject and sound leaking to the reference ear will be correspondingly reduced causing even less of a problem. Sound leaking into the aided ear and mixing with the aid output is the greater potential source of inaccuracy. Two tonal signals at the same frequency and level will combine to give a level 6 dB higher than each separately if they are in phase. If they are in antiphase, total cancellation will occur. Antiphase signals present the worst case even if the signal levels differ.

Whilst any signal leaking into the ear will have a level below that of the aid output, if it is in antiphase to the output it could modify the sound levels in the ear significantly. A leakage signal must be 19.2 dB below the aid output in order to affect the measured output by less than 1 dB. To be sure of affecting the measured aid output level by less than 3 dB the leakage must be at least 10.7 dB below the output.

How large an effect could earmould leakage have had in the present experiment? Figure 4.11 shows the mean attenuation of normal vaselined earmoulds measured as described in Appendix B, together with the mean gain of the aid measured in this experiment. Both curves were measured under free-field conditions with the sound source in front of the subjects.

Each curve represents the mean increase (or decrease) in the loudness or threshold level of the sound field in the test ear when the hearing aid and mould is worn, relative to the loudness or threshold of the sound heard in the test ear when unoccluded and unaided, i.e., the zero dB line is the level in the unaided ear.

The upper curve can be regarded as the insertion gain of the aid-mould system when the aid is switched on while the lower curve is the insertion gain of the same system when the aid is switched off but when sound may still enter the ear by leakage. Both are relative to the
level in the unoccluded ear, so leaving aside any differences in results which may arise between loudness matching and threshold methods (see Appendix B), the difference between the two curves is the amount by which the sound in the ear from the hearing aid receiver exceeds the unwanted sound leaking through or around the mould.

Taking into account the differences apparent in some studies (Appendix B) between loudness balance and threshold shift results will hardly affect the comparison between the hearing aid receiver output and the leakage sound at 250 Hz and 500 Hz but the difference between the two curves may be overestimated by up to 6 dB at 1 kHz and above. Fortunately it is where the curves are furthest apart that the overestimation occurs and at the crucial frequencies where the two curves are already closest that no correction need be made.

At 500 Hz and above the curves are about 20 dB or further apart. So in general there will have been little or no error in the measurement of the hearing aid gain due to earmould leakage. But the spread of results is quite wide and in some individual cases earmould leakage may have influenced hearing aid gain judgements. This is illustrated by Figure 4.12 which shows cumulative frequency distributions of the results making up the two curves in Figure 4.11. Once again assuming the worst, in this case, that poorly attenuating earmoulds were those with which a hearing aid gave the least measured output, then at 250 Hz in at least 55% of cases the hearing aid output would have been 10 dB or more above any leakage sound. Thus in at least 55% of cases the loudness would have been affected by less than 3 dB. Similarly at 500 Hz at least 70% of individual balances would have been affected by no more than 3 dB.

The above worst case assumptions are very severe. Since the spread of aid gain measurements is quite small it is likely that the estimated percentage of "suspect" results has been overestimated or that any "suspect" results were very similar to the definitely unaffected results. The effect of earmould leakage on the aid's mean measured gain at 250 Hz and 500 Hz appears to have been extremely slight. At other frequencies any effect would have been even less.

It is concluded that earmould leakage was not a significant factor, but that higher aid gain settings in future experiments would remove any possibility of doubt.
4.4.2 Differences between occluded and unoccluded listening

The discrepancy found in this experiment between the expected and measured subjective gain of a hearing aid may be a manifestation of the enigmatic "missing 6 dB" or "closed ear" effect.

This effect was first described by Sivian and White [9, 1933] in relation to the difference between monaural minimum audible field (MAF) and minimum audible pressure (MAP), and is summarised by Munson and Wiener [10, 1952] as follows: "There is evidence that a pressure level in the ear canal about 6-10 dB higher is necessary to elicit a given sensation of loudness in a given observer when an earphone is worn as compared with the case where the stimulus is presented from a loudspeaker".

The existence of such an effect is controversial and disputed on both theoretical and empirical grounds. Nevertheless, Munson and Wiener measured discrepancies of 8-12 dB at 100 Hz, 4-6 dB at 200 Hz and about 0 dB at 1 kHz between MAP under a supra-aural earphone and MAF from a loudspeaker - both sets of data being measured as pressures in the ear canal.

Anderson and Whittle [11, 1971] claim that the missing 6 dB can be explained in terms of an increase in physiological noise under an earcap causing stimulus masking. But this may not be the only factor as Munson and Wiener have shown the effect to persist at least to a loudness level of 60-80 phons.

Tillman, Johnson and Olsen [12, 1965] report an average difference between MAF and MAP for a supra-aural earphone of 7.5 dB and for an insert receiver of 12.5 dB using speech as a stimulus. They suggest that although 3-4 dB of this was due to measuring the free field in the absence of a subject, 3.6 dB in the case of a supra-aural earphone and 9 dB in the case of an insert earphone remained as the extra sound pressure necessary in the closed ear to produce the same loudness as a free field. This interpretation is questionable since sound levels generated by the supra-aural earphone and the insert receiver were measured in standard couplers and not in real ears.

Similar results to these however led Weber and Lawrence [13, 1954] to speculate that "the drum membrane and its associated structures are better matched to the impedance of the open air than to the impedance of a limited volume of air entrapped in the meatus" and that the closed ear
effect is a manifestation of the impedance mismatch when an earphone is worn.

The recent trends in the literature are towards explaining differences between MAP and MAF data as either masking by physiological noise or inadequacies in the measurement or calibration procedures. Rudmose [14; 1962] found no evidence of missing decibels at 100 Hz when comparing insert earphones with a loudspeaker, though there was some evidence of the effect with a supra-aural earphone.

Since the physiological noise measured under the ear insert was significantly lower than that under the supra-aural earphone, Rudmose concluded that the auditory threshold at 100 Hz is elevated by masking in the case of the supra-aural phone. This suggestion was later supported by the work of Anderson and Whittle cited above [11].

However, as Rudmose points out, masking cannot adequately explain "missing 6 dB" effects at levels well above threshold. He suggests that no single answer is sufficient to explain all reported instances of "missing 6 dB" effects.

Villchur [15, 1969] has shown that the response of a TDH-39 earphone relative to a free field remains the same whether the measurements are made by threshold shift, by supra-threshold loudness balance or by a probe microphone within 3 mm of the eardrum. He found no evidence of any closed ear effect at frequencies between 100 Hz and 9 kHz.

In 1974, Morgan and Dirks [16] concluded that discrepancies between earphone and freefield data were apparent if artificial ears and couplers were used to specify sound pressure levels but that such discrepancies disappeared if probe microphones were used to measure sound levels in real ears. Their own data presented in their figures 4 and 7, however, would seem to contradict their conclusion.

These figures show, with discrete test frequencies of 500 Hz and above and also with wide band noise, that subjects consistently required a greater sound level from an earphone than from a loudspeaker for equal loudness. This was apparent at both reference intensities of 90 dB SPL and 70 dB. Similarly for loudness discomfort levels; at and above 500 Hz a higher sound pressure was required from the earphone than from the loudspeaker to cause discomfort. Thus, although Morgan and Dirks's

39.
data show no sign of any closed ear effect at the usual frequencies of 250 Hz and below, at higher frequencies there is a discrepancy of a similar nature of up to 10 dB. They do not comment on their data at 500 Hz and above.

Killion [17, 1978] has recently collated published earphone and free field threshold pressure levels. When the acoustics of the head and external ear are accounted for in the free-field data and when real ear versus coupler differences are accounted for in earphone data, the sound pressure levels at the eardrum for free-field and for earphone listening can be compared. Killion has shown that these estimated threshold sound pressure levels at the eardrum for free-field and earphone listening are in excellent agreement, with no evidence of any "missing 6 dB" apart from that expected at low frequencies from masking by physiological noise.

4.4.3 Summary of discussion and future work necessary

From the above discussion it seems very unlikely that earmould leakage or closing the earcanal with an earmould could alone be responsible for the low subjective gain of the aid, but clearly not all aspects of the differences between earphone and loudspeaker listening are fully understood. It is not clear whether deficiencies in the Zwislocki ear simulator or perceptual differences between aided and unaided listening or some experimental artefact is responsible for those differences between the subjective and objective measurements which are not attributable to head baffle or earcanal resonance. To resolve this dilemma it will be necessary to determine the sound levels existing in aided and unaided ears using a probe tube microphone. It will then be possible to relate the pressure insertion gain of an aid on real ears to the pressure insertion gain on a manikin and the pressure transmission gain on a coupler or ear simulator, and also to compare sound pressures existing in aided and unaided ears at a given loudness.

4.5 Summary and Conclusions

The subjective frequency response of a behind-the-ear hearing aid was measured using a loudness balance technique. The aid's subjective gain was less than that measured conventionally on either the 2 cc coupler
of Zwislocki ear simulator, by up to 11 dB and 25 dB, respectively. Much of the discrepancy between the subjective and objective measurements was apparent rather than real and due to comparing an insertion gain, with its inclusion of head baffle and earcanal resonances, against transmission gains. This was confirmed by measurements on Kemar.

Nevertheless, further work is necessary to explain remaining discrepancies not attributable to head diffraction or earcanal resonance effects.

The poor choice of microphone location of this aid was better suited to sound incident from behind the wearer than to sound from in front.
<table>
<thead>
<tr>
<th>Subject</th>
<th>Session</th>
<th>Direction</th>
<th>Balanced first</th>
<th>Ear as fixed reference</th>
<th>Frequency order*</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1</td>
<td>F</td>
<td>U</td>
<td>L</td>
<td>i</td>
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<td>2</td>
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<td>U</td>
<td>L</td>
<td></td>
<td>vii</td>
</tr>
<tr>
<td>3</td>
<td>F</td>
<td>U</td>
<td>L</td>
<td></td>
<td>vii</td>
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</tr>
<tr>
<td>4</td>
<td>F</td>
<td>A</td>
<td>R</td>
<td></td>
<td>viii</td>
</tr>
</tbody>
</table>

Subjects assigned at random to numbers.

F = facing loudspeaker
B = back to loudspeaker
A = aided
U = unaided
L = left
R = right

continued/
TABLE 4.1 continued...

*Frequency order: -

i = A B H C G D F E
ii = B C A D H E G F
iii = C D B E A F H G
iv = D E C F B G A H
v = E F D G C H B A
vi = F G E H D A C B
vii = G H F A E B D C
viii = H A G B F C E D

A = 1 kHz; B = 6 kHz;
C = 1.5 kHz; D = 250 Hz;
E = 500 Hz; F = 3 kHz;
G = 4 kHz; H = 2 kHz.
### TABLE 4.2

**PRESSURE TRANSFER RATIO FROM AID MICROPHONE TO EARCANAL ENTRANCE**


<table>
<thead>
<tr>
<th>dB SPL at ear canal entrance - dB SPL at aid microphone *</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frequency in Hz</td>
</tr>
<tr>
<td>-----------------</td>
</tr>
<tr>
<td>SPL difference for</td>
</tr>
<tr>
<td>(a) Wearer facing sound source</td>
</tr>
<tr>
<td>(b) Wearer's back to sound source</td>
</tr>
</tbody>
</table>

### TABLE 4.3

**EARCANAL ENTRANCE TO EARDRUM PRESSURE TRANSFER RATIO** from Wiener & Ross [4]

<table>
<thead>
<tr>
<th>dB SPL at eardrum - dB SPL at ear canal entrance *</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frequency in Hz</td>
</tr>
<tr>
<td>-----------------</td>
</tr>
<tr>
<td>SPL difference</td>
</tr>
</tbody>
</table>

*Values quoted to nearest half-decibel.
FIG. 4.1 THE WIDEX BARITONE 641H AID

FIG. 4.2 BLOCK DIAGRAM OF EQUIPMENT USED FOR THE LOUDNESS BALANCE MEASUREMENTS ON THE WIDEX BARITONE AID
FIG. 4.3 SUBJECTIVE RESPONSE OF THE WIDEX BARITONE AID.

FIG. 4.4 RESPONSE OF THE WIDEX BARITONE ON THE 2cc COUPLER AND ZWISLOCKI EAR SIMULATOR.
FIG. 4.5 RESPONSE OF THE WIDEX BARITONE ON KEMAR

FIG. 4.6 DIFFERENCE BETWEEN COUPLER AND SUBJECTIVE RESPONSES. SUBJECTIVE DATA IS AVERAGED OVER BOTH DIRECTIONS OF SOUND INCIDENCE.
FIG. 4.7 SUBJECTIVE RESPONSE WITH ALLOWANCE FOR HEAD DIFFRACTION.

FIG. 4.8 SUBJECTIVE RESPONSE WITH ALLOWANCE FOR HEAD DIFFRACTION AND EAR CANAL RESONANCE EFFECTS.
FIG. 4.9 DIFFERENCES BETWEEN COUPLER AND SUBJECTIVE RESPONSES OF THE AID WHEN HEAD DIFFRACTION AND EARCANAAL RESONANCE EFFECTS ARE REMOVED FROM THE COMPARISON.

FIG. 4.10 RESPONSE OF THE AID ON KEMAR WITH ADJUSTMENT FOR (HUMAN) HEAD DIFFRACTION AND EARCANAAL RESONANCE EFFECTS.
FIG. 4.11 COMPARISON OF AID GAIN AND EARMOULD ATTENUATION.
FIG. 4.12 CUMULATIVE DISTRIBUTIONS OF AID GAIN AND EARMOULD ATTENUATION MEASUREMENTS AT 250 Hz AND 500 Hz.
CHAPTER 5

A PROBE-TUBE MICROPHONE AND ITS CALIBRATION

5.1 INTRODUCTION

Without knowing the relative sound pressure levels in the aided and unaided real ears it is not possible to explain fully the results of the experiment in Chapter 4. It will therefore be necessary to repeat the loudness balance, making simultaneous measurements with a probe microphone in the aided and unaided real ears. Suitable probe microphones are available and this chapter describes the calibration and use of one which is particularly useful [1, Villchur and Killion, 1975].

Whilst papers describing measurements with probe microphones in the ear are not uncommon it seems that the calibration is rarely described. Since a poor calibration technique will ultimately cause misleading results, great care was taken with the calibration in this study and further measurements were made to assess the accuracy of the calibration when used in a confined space similar to an ear. These validating measurements are described in the next chapter.

5.2 DESCRIPTION OF THE PROBE MICROPHONE

Figure 5.1a shows the microphone used complete with probe tube.

The microphone, a Knowles Electronics XL-9073, is a flat response version of a ceramic subminiature hearing aid microphone and has a built-in FET preamplifier powered by two 1.5 Volt dry cells. The supply circuit used was that described by the manufacturer as the standard source follower circuit [2, Knowles Electronics, 1975].

The microphone was used with a selection of soft plastic probe tubes each having an inside diameter of about 1.3 mm but differing in length to allow measurements to be made at different depths into an ear canal. The probe tubes were formed by inserting 18 gauge (1.2 mm diameter) solder and bending them to shape in boiling water. The solder, which prevented the tubes from collapsing was then removed.

So that different probe tubes could be used on the same microphone

42.
unit the probes were not cemented in place over the microphone vent as would normally be done for a good leak-proof seal. Instead a length of fine fuse wire was coiled tightly around the end 2 mm of the tubing before immersion in the boiling water, but with the solder already in place. When the tubing was taken from the water and the solder and wire removed, the final 2 mm of tubing was constricted, and when trimmed, gave an airtight push fit onto the microphone vent as opposed to the fairly loose fit of the unconstricted tubing.

No acoustic damping was used in the tubes as repeatability and stability of the response was considered more important than minor improvements to an already smooth response. The rough edge at the open end of the tube was sanded smooth to prevent discomfort to subjects.

5.2.1 Advantages of this microphone

The microphone has the advantage of being very small and light and can therefore be attached to the subject's head with surgical tape just in front of the ear (Figure 5.1b). No complicated head clamp or mounting jig is required and since the microphone moves with the subject the risk of damage or discomfort to the subject is far less than with more traditional metal probe-tube microphones.

The small size also means less disturbance to the sound field. The absence of bulky head clamps or mountings is more reassuring to the subject who is not restricted in any way.

5.3 GENERAL REMARKS ON PROBE-MICROPHONE CALIBRATION

Probe microphones may be calibrated either in free field or in a rigid-walled coupler [3, Benson, 1953; 4, Delany, 1968] using a known microphone as a calibrating standard. Reciprocity methods, while possible in principle, are impractical with probe microphones [4].

Benson has shown that slight differences exist between a calibration in free field and the corresponding calibration in a coupler. But although these differences may be important if extreme accuracy is required, the differences found by Benson were not large and rarely exceeded 1 dB unless the probe tube was very long (100 mm) or narrow (0.6 mm).
In addition, Delany and Rennie [5, 1969] have pointed out possible faults in commercially available probe-calibration couplers and the precautions necessary to avoid them. Such faults can result in asymmetrical sound fields or very steep pressure gradients at high frequencies, requiring accurate microphone location.

For a mathematical modelling of a probe microphone and a discussion of the theory see Egolf [6, 1977].

5.4 THE CALIBRATION OF THE MICROPHONE

5.4.1 The method

The method chosen here for the calibration of the Knowles XL-9073 probe microphone was a free-field comparison with a standard Bruel and Kjaer 4135 quarter-inch microphone at discrete test frequencies. This method is quick, relatively simple and requires no special coupler or non-standard equipment. Figure 5.2 shows the equipment used. The reference microphone was calibrated initially against a pistonphone.

The sound source was a B & K artificial mouth in an anechoic room and the probe tube open end and the quarter-inch reference microphone were approximately 70 mm from the mouth on its axis. At this distance the sound field over the small area encompassed by the two microphones should approximate a progressive plane wave in free field.

A comparison rather than a substitution method was used with the reference and probe microphones close together rather than occupying the same space alternately. Figure 5.3a shows the orientation of the reference microphone and the probe for the calibration. The sound field from the artificial mouth was at grazing incidence to both the reference microphone diaphragm and the probe orifice to minimise diffraction and give a pressure calibration. The axes of the probe and of the reference microphones were perpendicular to minimise reflections between the microphones, with the probe about 1¼ mm from the reference microphone grid. The presence of the probe affected the reference microphone reading by not more than 0.1 dB.

The probe microphone output for a constant sound level at the reference microphone was measured from the dB scale of the measuring amplifier.
with allowance being made for the individual pressure response of the reference microphone as supplied by B & K. This was done for each probe length at each test frequency and from each reading a "K-factor" was derived which when added to the meter reading, gave the sound level in dB above 20 µPa. These are the figures used in all subsequent measurements.

5.4.2 Frequency response and sensitivity

Figures 5.4, 5.5 and 5.6 show the frequency response of the microphone unit without a probe, with a medium length and with a long probe tube, respectively. These figures are for illustration and no calibration data were measured from level recordings. The sensitivity of this particular XL-9073 without a probe was measured as 49 dB below 1V/Pa at 1 kHz, equivalent to 69 dB below 1V/µbar, which is the same value as on the individual response curve supplied with the microphone by the manufacturer.

5.4.3 Stability of the microphone sensitivity

The output from the microphone was stable both in the short term and from day to day within the resolution of the measuring amplifier scale as read by eye (i.e., nearest 0.1 dB approximately).

5.4.4 Noise floor

The noise floor of the microphone with all associated leads and equipment was 8µV (rms, slow, linear with 22.4 Hz high pass filter) or 2µV A-weighted. These are equivalent to sound levels of 41-42 dB(linear) or 30 dB(A) re 20µPa.

5.4.5 Directionality of the probe

As described above the orientation of Figure 5.3a was used to calibrate the probe. However as a check on the directionality of the probe, one probe tube was calibrated at each of the four orientations, a-d, shown in Figure 5.3.

The results, given in Figure 5.7, show the probe to be omnidirectional up to 4 kHz and omnidirectional within a 2.1 dB range up to 7 kHz. This is perfectly adequate for hearing aid measurements.
5.4.6 Leakage of sound through the probe wall and at the connection of the probe to the microphone

To test for sound leaking to the microphone through the probe wall and at the connection between probe and microphone, the open end of the longest probe tube (considered to represent a worst case) was closed with a small amount of "Otoform K" earmould impression material and recalibrated. The output from the microphone was reduced by 29 dB or more at all test frequencies.

Probe leakage is therefore considered not to present any problem, and the method of attaching the probe to the microphone is satisfactory.

5.4.7 Accuracy of the calibration under non-free-field conditions

The accuracy of the calibration was checked by comparing measurements in the occluded and unoccluded ear of Kemar as made with the calibrated probe microphone with the readings from the built in B & K 4134 1/4 inch "eardrum" microphone as described in the following chapter.

5.5 SUMMARY AND CONCLUSIONS OF CHAPTER 5

A free-field pressure calibration of the Knowles XL-9073 microphone with polythene probe tubes shows this probe-tube microphone to be omnidirectional below 4 kHz and almost so, to within 2dB, up to 7 kHz. The electrical noise floor is low, equivalent to 42 dB (linear) re 20 μPa or 30 dB(A). Sound leakage through the probe wall and at the tube to microphone connection is about 30 dB below the sound pressure level at the probe tube entrance.

These properties are well suited to hearing aid measurements.
FIG 5.1a THE KNOWLES XL-9073 PROBE MICROPHONE

FIG 5.1b THE PROBE IN KEMAR'S UNAIDED EAR CANAL
FIG 5.2 EQUIPMENT USED TO CALIBRATE THE PROBE TUBE MICROPHONE

FIG. 5.3 ORIENTATION OF PROBE AND REFERENCE MICROPHONE TO THE SOUND FIELD; 'a' chosen for probe calibration.
FIG. 5.4 FREQUENCY RESPONSE OF THE XL-9073 MICROPHONE UNIT WITHOUT A PROBE TUBE.

FIG. 5.5 FREQUENCY RESPONSE OF THE XL-9073 MICROPHONE WITH A MEDIUM LENGTH PROBE TUBE.
FIG. 5.6 FREQUENCY RESPONSE OF THE XL-9073 MICROPHONE WITH THE LONGEST PROBE TUBE.

FIG. 5.7 DIRECTIONALITY OF THE PROBE MICROPHONE (directions a-d are defined in Figure 5.3).
CHAPTER 6

SOUND LEVEL MEASUREMENTS IN AN AIRED AND UNAIRED MANIKIN'S EAR

6.1 INTRODUCTION

If the gain of a hearing aid is to be derived from sound pressure level measurements in an aided and unaided ear the ideal position for such measurements is the eardrum. But safety restrictions and practical difficulties especially within aided ears precluded probe microphone measurements that deep in this study.

The gain of the aid in terms of sound pressures at the drum must therefore be estimated from measurements at a more practical position in the aided and unaided ear canal, and this assumes a knowledge of the variation in sound pressure level within the aided and unaided ear canal right up to the eardrum at each test frequency of interest.

6.2 PUBLISHED DATA ON SOUND LEVEL VARIATIONS IN REAL EARS

6.2.1 Open ear - loudspeaker listening

Research papers describing probe microphone measurements in open ear canals are not uncommon; Shaw [1, 1974] for example cites twelve, and although open ear pressure transfer ratios from canal entrance to eardrum are often given [2, Wiener and Ross, 1946; 3, Yamaguchi and Sushi, 1956; 4, Jahn, 1960; 5, Djupesland and Zwislocki, 1972] and the transfer ratio from canal entrance to a point partway along the canal is occasionally given [2; 5] rarely is more detail available. The information which does exist is not always in good agreement and almost nothing is known regarding sound sources outside the horizontal plane.

Whilst it would be possible in this study to measure the sound level at the entrance of the unaided ear ear canal and estimate the sound pressure at the eardrum from the published pressure transfer ratios, it is likely that a greater accuracy would be obtained with measurements closer to the eardrum, and it would be preferable to measure at the same point in unaided ears as in the aided ears, especially if the sound pressure
variation close to the drum were found to be similar under both conditions.

A pilot test of the probe microphone was made on one real unoccluded ear (see Appendix C) and this suggests that sound pressure levels at the frequencies of interest were less variable nearer to the eardrum.

6.2.2 With ear occluded by insert earphone or earmould

Most probe microphone measurements in ears occluded with earphones relate to supra-aural earphones [e.g., 6, Corliss and Burkhard, 1953; 7, Villchur, 1969] and are usually concerned with transferring audiometric threshold data from a standard to a newer earphone. Since most audiometric earphones are of similar construction, shape and size, the pressure distributions in the ear will not vary from one type to another to any great extent and the positioning of a probe tube is not too critical. A probe located at the centre of the surface of the earphone or at the canal entrance has often been regarded as adequate, though Villchur does not agree with this approach.

Real ear probe microphone measurements under insert receivers and earmoulds are fewer. Many of these can be criticised for positioning the probe tube such that estimates of pressure at the eardrum are inherently inaccurate.

Among earliest measurements were those of Nichols et al [8, 1945] which are also described by Beranek [9, 1949]. For these measurements the probe tube was terminated flush with the tip of the earmould through which it passed.

More recently Ewertsen et al [10, 1957] used a similar technique, again with the probe tube ending at the earmould tip (see their Figure 3). Measurements by Dalsgaard published in 1971 [11] were also made with the probe tip flush with the earmould surface, although for his more recent measurements he has extended the probe beyond the mould by a few millimetres [12, Dalsgaard and Jensen, 1976].

Sachs and Burkhard [13, 1971] have shown theoretically that the sound field in the region of the earmould tip can vary quite drastically at higher frequencies, due to radial spreading of sound from the earmould tube into the wider earcanal. At each frequency above about 5 kHz there
will be a contour, centred on the sound inlet tube, where the sound pressure will fall drastically. In a hard walled cavity the sound pressure would fall to zero but even in a real ear there would be severe pressure gradients, and even slightly lower frequencies could be affected to some extent. Probe tube pressure measurements in the plane of the earmould tip may therefore be erratic and give a poor estimate of the pressure at the eardrum above, say, 3 kHz.

Extending the probe tube by about 3–5 mm beyond an earmould will avoid the anomalous sound distribution near the mould and will give reliable measurements up to 8 kHz at least.

6.3 NECESSITY FOR MORE DATA

The published data on the variation of sound levels within real ear canals as described above is neither sufficiently detailed nor sufficiently consistent to enable sound pressure levels at the eardrum to be predicted from measurements removed from the eardrum in both aided and unaided ears.

To obtain more detailed information measurements were made along the aided and unaided ear canal of Kemar, and such measurements as were considered safe from damage or discomfort were made along aided and unaided real ear canals. It would then be possible to compare Kemar's ear and real ears where similar measurements were possible on both, and to use Kemar as a guide to the acoustic behaviour of real ears close to the eardrum.

As well as being useful in their own right the measurements in Kemar's ear enable the accuracy of the probe calibration to be assessed under conditions not only similar to those in real ears but also repeatable. As such these measurements form an extension of the probe calibration of the previous chapter.

The measurements on real ears are described in the next Chapter.
6.4 PROBE MEASUREMENTS OF SOUND PRESSURE IN THE EAR CANAL OF THE MANIKIN KEMAR

6.4.1 Aims

To assess the accuracy of the probe microphone pressure calibration; to map the sound field within the occluded and unoccluded ear of a manikin; to develop and practice techniques suitable for real ear measurements, and to obtain data comparable with real ear data in order to be better able to extrapolate the real ear data to the eardrum.

6.4.2 Measurements in the unaided, unoccluded ear

Using the different length probe tubes, measurements of sound pressure were made at depths generally 2 mm apart along the length of Kemar's left ear canal whilst simultaneous measurements were made with the "ear drum" microphone for comparison.

All measurements were carried out in the anechoic room, using a Goodmans Axiette loudspeaker driven at a constant 0.5 volts from a beat frequency oscillator as a sound source.

A horizontal reference is defined for Kemar by the line connecting the lower eyelid of the open eye and the upper pinna-skull notch as viewed from the side [14, Burkhard and Sachs, 1975]. The loudspeaker was placed 64 mm below this horizontal directly in front of Kemar at 1 metre from the centre of his head. This is the position proposed by Burkhard [15, 1976] for consideration as a standard in manikin measurements, and is approximately at mouth level.

The loudspeaker box was covered with a thin layer of sound absorbent foam to minimise reflections and any standing waves between the subject and the loudspeaker. The diameter of the loudspeaker cone was 180 mm approximately which again conforms to proposed standards for manikin measurements.

The frequency response and harmonic distortion characteristics of the loudspeaker in its box are shown in Figures 6.1 and 6.2. The response of the loudspeaker off axis is similar to that on axis and consequently the sound field in the space to be occupied with the head is quite uniform.

The probe microphone was taped in position in front of the ear so that the probe tube passed through the notch above the tragus then turned sharply
inwards to enter the ear canal, as shown in Figure 5.1b.

To prevent damage to the 'ear drum' microphone diaphragm and to gain maximum accuracy in the positioning of the deepest probe tube, this tube was inserted before the "ear drum" microphone was screwed into position. The probe microphone output was measured using the same equipment as was used in the calibration (see Figure 5.2).

6.4.2.1 The effect of the probe on the existing soundfield in the ear

Figure 6.3 shows how the sound pressure levels measured at the 'ear drum' microphone were changed by inserting the longest probe tube into Kemar's ear canal. This probe was considered to represent a 'worst case' as its open end was only 1½ mm from the "ear drum" but the sound pressure level at the eardrum at 3 kHz did not change by more than 1½ dB and at other frequencies the change was even less.

6.4.2.2 The probe microphone and the eardrum microphone compared

This measurement serves as a check of the probe microphone (free-field) calibration described above. In Figure 6.4 the sound pressure level measured by the 'ear drum' microphone and the sound pressure level measured simultaneously at the probe microphone 1½ mm away are compared. The probe tube was in position for both sets of measurement.

Since the same preamplifier, measuring amplifier and pistonphone were used with the half inch 'ear drum' microphone as with the quarter inch calibration reference microphone, sources of error attributable to these items are self-cancelling in the derivation of Figure 6.4.

What Figure 6.4 shows therefore includes (i) the real SPL differences between the probe and the 'ear drum' positions, (ii) the residual calibration errors or directionality effects of the probe microphone and (iii) any change from free-field behaviour of the probe due to the change in input impedance seen by the probe when placed in a confined space (i.e., the difference between free-field and coupler calibrations identified by Benson [16, 1953]).

The net effect of all these factors is 0.6 dB or less at all frequencies below 3 kHz, and is 1.6 dB or less at all test frequencies.
Had the orientation of Figure 5.3d been used when the probe microphone was calibrated, the agreement between the probe and "eardrum" microphone measurements would have been improved slightly at 5 kHz, 6 kHz and 7 kHz but the maximum difference which is 1.6 dB at 4 kHz would have been unchanged. The agreement in any case is perfectly acceptable.

6.4.2.3 Sound pressure level variations within Kemar's open earcanal

Figures 6.5 and 6.6 show SPL vs frequency curves at various depths into Kemar's earcanal. In Figure 6.5 the values are relative to the earcanal entrance position and apply for a constant level sound source.

In Figure 6.6 values are relative to the deepest probe position, 1\half mm from the 'eardrum' microphone, and have been adjusted to apply for a constant level at the 'eardrum' microphone.

The former set of curves (Figure 6.5) can be related to measurements readily made in real ears, but the latter set (Figure 6.6) is perhaps the most useful if it is wished to predict the SPL at, or at least near, the eardrum from a measurement elsewhere in the earcanal. Unfortunately, while the measurements in Figure 6.6 are easy to make with Kemar they are not possible on real ears without the use of two probe tubes, one of them reaching almost to the eardrum - an inherently difficult and questionable procedure.

The curves in Figure 6.6 were normalised to the deepest probe at 1\half mm from the eardrum and not to the drum itself so as to avoid mixing measurements from the different microphones.

The adjustments from a constant level sound source to a constant level at the 'eardrum' applied between Figures 6.5 and 6.6 are small, of the order of 0.1 dB at 250 Hz and 500 Hz and therefore not reliably detectable, rising to just greater than 1 dB at 3 kHz at some probe depths. Such adjustments are due to the presence of the probe tube and should ideally be unnecessary. However, the adjustments are acceptably small.

Figure 6.7 shows the variation in SPL at each test frequency along the length of Kemar's earcanal with a constant eardrum SPL. At 2 kHz and lower frequencies there is a slight drop in SPL with increasing distance from the eardrum; at 3 kHz there is a more dramatic drop of 9 dB along the canal; whilst at higher frequencies, 4 kHz and above, the SPL falls to a minimum within the earcanal's length then rises again, with the position of the minimum moving deeper into the canal with increasing frequency. This is
consistent with the length of the ear canal being a half-wavelength at about 3.4 kHz, and the curves of Figures 6.5, 6.6 and 6.7 form self-consistent sets with no obviously anomalous readings.

Approval was given by the ISVR Safety and Ethics Committee [17, 1976] for probe measurements at up to 5 mm beyond an earmould tip and at an equivalent position in the unoccluded ears of volunteer subjects. The above results on Kemar suggest that at this depth, i.e., 9 mm from the eardrum, the SPL measured with a probe in the unaided ear would be within 2 dB of the SPL at the eardrum for frequencies up to 4 kHz.

6.4.2.4 Free-field to eardrum pressure transfer ratio in Kemar

Though not involving the use of a probe microphone this is a convenient point to include measurements of the free-field to eardrum pressure transfer ratio of Kemar.

For a constant electrical input to the loudspeaker the sound pressure level at each frequency was measured at the 'eardrum' microphone in Kemar, and with Kemar removed, at the position previously at the centre of Kemar's head.

The same half inch microphone which had been Kemar's eardrum was used for the free-field measurements so that any calibration errors were self-cancelling. Being a pressure microphone it was used at 90° (grazing incidence) to the sound source to give a flat free-field response. The free-field measurements were however made one day later than the in-ear measurements.

Figure 6.8 shows the measured values of the free-field to eardrum transfer ratio in Kemar and these are overlaid on curves published by Burkhard and Sachs [14]. The agreement is very good with the points measured here being within about 1 dB of the comparable curve. Such small differences could easily arise from slight differences in sound source directionality, differences in Kemar's clothing, or day to day equipment stability, and imply that the sound source, anechoic room and experimental technique used are satisfactory.

For a detailed comparison of Kemar's free-field to eardrum transfer with human data see Burkhard and Sachs [14]. Briefly the transfer ratio in Kemar agrees extremely well with the corresponding human data, the transfer ratio measured on Kemar being within 3 dB of the real ear curves

53.
published by Shaw [1] for all frequencies tested (up to 7 kHz).

6.4.3 Measurements in the aided ear of Kemar

Probe microphone measurements of sound pressure level were made in the occluded left ear canal of Kemar using the set up shown in Figure 6.9 to simulate the use of a hearing aid.

The earmould was simulated with 2 mm inside diameter plastic tubing passing through an EAR plug which had been cut to length. The tubing was attached to the hook of a BTE hearing aid shell containing a Knowles XD 1005 receiver, such that the standard 40 mm of tubing extended beyond the hook to end flush with the surface of the EAR plug. This surface was positioned inside Kemar's ear canal at the connection between the Zwislocki coupler and Kemar's own canal extension. The BTE aid was worn in the normal position by Kemar.

A probe tube of 1.3 mm inside diameter connected to the XL-9073 microphone passed through a second hole in the simulated earmould.

The XD 1005 receiver in the hearing aid was driven directly by a beat frequency oscillator at a constant voltage at each test frequency and the sound pressure generated in the ear canal was measured with the probe microphone at 1 mm increments between the 'eardrum' and the 'earmould' surface while sound level measurements were made simultaneously at the 'eardrum' microphone. The probe was positioned as centrally as possible for these measurements.

6.4.3.1 The probe microphone and the eardrum microphone compared

In Figure 6.10 the sound pressure level measured by the 'eardrum' microphone is compared with the sound pressure level measured by the probe microphone 1 mm away in Kemar's aided ear. This serves as a further calibration check.

The measurements in Figure 6.10 are very similar to those presented in Figure 6.4 which applies to Kemar's unaided ear and the comments made in Section 6.4.2.2 concerning Kemar's unaided ear may be taken to apply also to Kemar's aided ear. The probe calibration applies equally well for measurements in occluded or unoccluded ears.

54.
6.4.3.2 Sound pressure level variations in Kemar's aided ear

As the probe tube was withdrawn 1 mm at a time from Kemar's aided ear the SPL registered by the eardrum microphone fell progressively, though slightly, even though the hearing aid receiver was being driven at constant voltage. The maximum fall off was 1.0 dB at 6 kHz and 7 kHz with the probe being moved from 1 mm from the 'eardrum' back to the earmould surface. This is presumed to be due to the increase in the volume enclosed in the ear canal as the probe is withdrawn. Figure 6.11 illustrates the frequency dependence of this effect.

Figure 6.12 shows the variation in SPL with depth along Kemar's aided ear and applies for a constant sound level at the eardrum. The curve for each frequency has been normalised to the sound level measured at the deepest probe position. Any slight deviation from a smooth curve particularly at the higher frequencies will be due more to inaccuracies in measuring probe depth than to inaccuracies in measuring sound levels.

The distance from the earmould to the eardrum is a quarter wavelength at about 6½ kHz, hence the reduction in SPL towards the earmould at 6 kHz and 7 kHz in particular. At slightly higher frequencies than those measured the SPL would fall to a minimum close to the earmould and rise again towards the mould. However radial spreading of sound from the sound inlet tube into the wider canal will become a significant contributory factor which will influence high frequency sound level distribution close to the earmould. This has been discussed theoretically by Sachs and Burkhard [13] for a hard-walled ear simulation. However the practical case here may vary since the earmould simulator is not perfectly rigid, the sound inlet tube is slightly off axis, and the probe tube is present.

It is to avoid the anomalies in the sound field that the probe must be inserted well beyond the earmould tip if a reasonable estimate of sound pressure at the drum is to be made at higher frequencies. With the probe at a 5 mm depth beyond the earmould, i.e., about 9 mm from the eardrum, the SPL registered by the probe will be within 2 dB of the SPL at the eardrum up to at least 4 kHz.

In fact in both aided and unaided measurements the sound level measured with a probe 9 mm from Kemar’s eardrum was within 2 dB of the sound level at the drum for frequencies up to 4 kHz. In addition the variation in sound

55.
pressure in the 9 mm of earcanal nearest the eardrum was similar for the open and the closed ear and errors due to not measuring sound level at the ideal position of the eardrum will very nearly cancel in the calculation of a hearing aid insertion gain, to within 1 dB even up to 7 kHz, provided the end of the probe tube is accurately located at the same distance from the eardrum in both the aided and unaided ear. Any errors in calibrating the probe microphone will also cancel.

However, Kemar has no physiological noise and sits perfectly still without breathing or swallowing so it is unlikely that measurements to the same accuracy could be made easily in real ears. Also the pressure distribution in Kemar's ear may or may not be a good simulation of a median real ear especially close to the eardrum. In any case the above measurements cannot be regarded as substitutes for real ear measurements, but rather as useful comparisons with real ear data providing approximations to measurements that are difficult or impossible in real ears, and giving an idea of the errors and problems involved.

6.5 SUMMARY AND CONCLUSIONS OF CHAPTER 6

The free field pressure calibration of the probe-tube microphone described in Chapter 5 and the calibration in the coupler which constitutes Kemar's ear agree to within $\frac{1}{2}$ dB up to 3 kHz and to within 2 dB up to 7 kHz. The method described in the previous chapter is therefore sufficiently accurate for use in real ear measurements.

Measurements in the unaided ear of the manikin Kemar show that the introduction of the probe changes the existing sound distribution very slightly, less than 1 dB at all frequencies, except at 3 kHz where the change is $1\frac{1}{2}$ dB. This small change is easily avoidable if the probe is left in position during all measurements which are to be compared.

A probe measurement 9 mm from Kemar's eardrum will be within 2 dB of the sound pressure adjacent to the drum in both the aided and unaided ear up to 4 kHz. Since errors between aided and unaided measurements will tend to cancel, measurements of hearing aid insertion gain using a probe 9 mm from the eardrum would be well within 1 dB of the corresponding gain measured at the drum. These above conclusions relate to the manikin Kemar only—real ear measurements are likely to be more variable.

56.
FIG. 6.1 FREQUENCY RESPONSE OF THE GOODMANS AXIETTE LOUDSPEAKER (constant voltage).

FIG. 6.2 HARMONIC DISTORTION OF THE GOODMANS AXIETTE LOUDSPEAKER.
FIG. 6.3 CHANGE IN KEMAR'S UNAIDED 'EARDRUM' SPL ON THE INSERTION OF THE PROBE MIC. TO WITHIN 1\(1/2\) mm OF THE 'EARDRUM' MIC.

FIG. 6.4 COMPARISON OF PROBE AND EARDRUM MICROPHONE MEASUREMENTS OF SPL IN KEMAR'S UNAIDED EAR
FIG. 6.5 VARIATION IN SPL IN KEMAR'S UNAIDED EAR RELATIVE TO CANAL ENTRANCE for constant source spl.
FIG. 6.6 VARIATION IN SPL IN KEMAR'S UNAIDED EAR RELATIVE TO A POSITION ADJACENT TO THE EARDRUM for constant eardrum spl
FIG. 6.7 VARIATION IN SPL ALONG KEMAR'S UNOCCLUDED EARCANAL
FIG. 6.8 FREE FIELD TO EARDRUM (UNAIDED) PRESSURE TRANSFER RATIO FOR KEMAR

FIG. 6.9 MEASURING SOUND PRESSURE UNDER A SIMULATED HEARING AID FITTING ON KEMAR.
FIG. 6.10 COMPARISON OF PROBE AND EARDRUM MICROPHONE MEASUREMENTS IN KEMAR's EAR UNDER THE SIMULATED HEARING AID FITTING.

FIG. 6.11 CHANGE IN EARDRUM SPL IN KEMAR WHEN THE PROBE IS WITHDRAWN TO THE EARMOULD FROM A POSITION 1mm FROM THE EARDRUM.
FIG. 6.12 VARIATION IN SPL IN KEMAR'S AIDED EAR
CHAPTER 7

SOUND PRESSURE LEVEL VARIATION ALONG AIDED AND UNAIDED REAL EARCANALS

7.1 INTRODUCTION

Manikin measurements are no substitute for real ear data if such can be obtained. Firstly a manikin might not adequately or completely represent a real head and ear acoustically - the discrepancies in Chapter 4 for example raise some doubts - and secondly a manikin gives no idea of the likely inter-subject differences. A manikin must wherever possible be checked against real ear measurements.

But a manikin is necessary since some measurements are not possible for a variety of reasons in real ears. A manikin can then be a useful guide to the acoustics of a real ear, or an aid to extrapolating real ear data. If the manikin ear behaves similarly to a real ear for all measurements where direct comparisons are possible, if it shows no odd or unexpected variations in sound level otherwise, and if it is known that the free field to eardrum pressure transfer ratio of the manikin is similar to that of a median real ear (which for Kemar it is), then the assumption is reasonable that the real ear is likely to be equally free of odd or unexpected variations in sound level, and that the real ear data may be extrapolated with reasonable confidence.

This chapter presents and describes probe microphone measurements of sound pressure variations to a permitted depth in real ear canals when aided and unaided. The measurements are used to estimate the accuracy with which a measurement of hearing aid response at the permitted depth would represent the response of the aid measured at the ideal but impractical position of the eardrum.

7.2 SAFETY CONSIDERATIONS

Permission was granted by the ISVR Safety and Ethics Committee to make probe microphone measurements in volunteer subjects' ear canals to a depth of 5 mm beyond a hearing aid earmould and to the equivalent depth in unoccluded ears. In a typical ear the earmould tip would be about 14 mm from the eardrum and the end of the probe tube about 9 mm from
the eardrum, but the variation between ears would be wide. In the case of one person, (viz., the author who was both subject and experimenter) measurements under an earmould were easily possible with care to a depth of 13 mm beyond the mould (7.3.3). Although beyond this depth the probe could be felt touching the earcanal wall there was no discomfort [1, Lower, 1976].

7.3 METHOD

Sound pressure levels were measured at various depths in the unoccluded and earmould-occluded left earcanals of four otologically and audiometrically normal male subjects (as defined in section 3.4).

7.3.1 Unoccluded, unaided ear measurements

The technique was essentially the same as that used on Kemar. Measurements were made using the equipment shown in Figure 7.1 with each subject seated in the anechoic room.

The loudspeaker was in the standard position, 1 metre from the subject's head centre and at mouth height, and was driven at a constant voltage from the beat frequency oscillator.

The SPLs generated at different depths 4 mm apart in the subject's left ear were measured using different length probe tubes with the XL-9073 microphone. The probe tubes were individually calibrated as described in Chapter 5 and the microphone was attached to the subject as shown in Figure 5.1. The 4 mm increments were chosen as sufficiently closely spaced following the measurements on Kemar (Chapter 6) in which increments were 1 mm, and a pilot experiment on a single subject (Appendix C) in which increments were 2 mm. It was estimated from the pilot experiment that the probe tube tip could be positioned to ± 1 mm of the desired depth into the ear.

The probe microphone output was filtered to pass the third-octave band containing the test frequency in use. Without filtration the test signal was masked by low frequency physiological noise, mainly below 100 Hz and modulated at the subject's heart rate.
7.3.2 Aided, occluded real ear measurements

Figure 7.2 illustrates the equipment used. The sound source was a simulated hearing aid comprising a Knowles XD 1005 wideband receiver in a hearing aid shell connected into each subject's ear via a custom-made earmould with a 2 mm bore sound tube. However, unlike normal earmoulds, these moulds were equipped with a second, narrower bore (approx. 1.2 mm) soft plastic tube as shown in Figure 7.3.

This second tube functioned as a probe tube for the XL-9073 microphone and could be pushed, with difficulty, through the mould to enable measurement of sound pressure at different depths in the ear. These measurements were made at 1 mm increments up to the safety limit of 5 mm beyond the earmould with a constant b.f.o. output voltage driving the hearing aid receiver. Again each probe tube was individually calibrated and the microphone output third-octave bandpass filtered.

7.3.3 Other measurements

Probe measurements were made to a depth of 13 mm beyond the earmould in the occluded ear of one subject. Frequency sweeps instead of discrete test frequencies, and automatic recording of the probe microphone output on a level recorder with a synchronised filter set were used due to the limited mobility of the subject who was also the experimenter.

7.4 RESULTS

Figure 7.4 shows the variations in sound pressure level generated along individual unoccluded earcanals at each test frequency. In Figure 7.5 these results are averaged over subjects and compared with the measurements on Kemar. The mean results of Figure 7.5 are tabulated with standard deviations and confidence intervals in Table 7.1.

In the above figures all variations in sound pressure level are plotted relative to the sound pressure level measured at the earcanal entrance since the eardrum position and the sound pressure level at that position were not known accurately.
Individual results for the 4 subjects' left ears when occluded with a hearing aid earmould are illustrated in Figure 7.6. The means of these results are presented in Figure 7.7 (bottom) which also shows the results for the single subject upon whom measurements were made to a depth of 13 mm beyond the earmould (top) and comparative results for Kemar (middle). Variations in sound pressure level under an earmould are plotted relative to the sound pressure level measured with the end of the probe tube flush with the earmould tip.

Figure 7.8 gives a comparison of absolute sound pressure levels generated 5 mm beyond the 'earmould' in Kemar's occluded Zwislocki ear simulator with the mean levels generated 5 mm beyond the mould in real ears under the same conditions.

(Note that the vertical scales differ among graphs.)

7.5 DISCUSSION

7.5.1 Variations in sound pressure level along unoccluded, unaided real ear canals

As can be seen from Figure 7.4 variations between individuals are large even at frequencies as low as 1 kHz, though general trends can be picked out at each frequency. But even with as few as four subjects the mean variation in sound level along the ear canal at each frequency is very similar to that measured on Kemar, and although above 4 kHz the positions of the pressure antinodes in the real ears and Kemar's separate out, the similarity is excellent up to 3 kHz, generally within 2 dB, and good above. In fact the pressure transfer ratio in real ears from the canal entrance to the deepest probe position (16 mm - equivalent approximately to 5 mm beyond an earmould in an occluded ear) is within 2 dB of the comparable value for Kemar at all frequencies.

It should be remembered when comparing Kemar data with real ear data, that Kemar gives the response of a median individual, rather than the median or mean response of a group of individuals. Group averaging will smooth out details not occurring at exactly the same frequency in each individual whereas Kemar will preserve these but at a median frequency.
Although pressure transfer ratios from earcanal entrance to ear-
drum have been measured and published by many authors (see Shaw [2; 1974])
the pressure transfer ratio from the entrance to some intermediate
position is rarely available. Wiener and Ross [3, 1946] and Djupesland
and Zwislocki [4, 1972] however do provide notable exceptions.

Wiener and Ross measured the pressure transfer ratio from the ear-
canal entrance to the midpoint. Since they found the average length of
the auditory canal to be 23 mm the average midpoint would be 11 \(\frac{1}{2}\) mm from
the entrance. The transfer ratio from the earcanal entrance to the
12 mm depth in the present study (Fig. 7.5) should therefore be comparable
with Wiener and Ross' data. In fact as Figure 7.9 shows, agreement is
excellent below 3 kHz but only fair above, though with the small number
of subjects involved there will be no significant statistical differences.

The earcanal length of 23 mm quoted by Wiener and Ross is perhaps
shorter than typical and it is estimated that subjects in the present
study had canals longer than this, about 25-26 mm on average, a length
more in keeping with the 25.7 mm quoted by Bezold [5, 1882] and
Johansen [6, 1975], the 27 mm found by Bekesy [7, 1960], and the 26 mm
found by Courtois and Berland [8, 1972]. The resonant frequency of
the earcanals in this study would therefore be about ten percent lower
than that measured by Wiener and Ross, and furthermore if the midpoint of
the earcanal were taken more realistically as 13 mm, the sound levels at
the midpoint would be slightly increased at 4 kHz and 5 kHz (interpolation
from Figure 7.5). The resonant frequency of Kemar's ear canal also
appears to be slightly lower than the real ear value measured by Wiener
and Ross. Much if not all of the small difference between Wiener and
Ross' data and that of the present study will be due to real differences
between the actual ears rather than measurement artefacts or errors.

Djupesland and Zwislocki [4] provide a pressure transfer ratio from
the earcanal entrance to a point in the earcanal 10 mm from the entrance
as shown in Figure 7.10. This was derived by subtracting their figure 3
from their figure 2. Comparable real ear data points from the present
study also plotted in Figure 7.10 were interpolated from sound levels at
8 mm and 12 mm into the earcanal. The agreement is good, though again it
seems that the earcanals in the present study are a little longer than
those measured by Djupesland and Zwislocki. Measurements on Kemar agree closely with both sets of real ear data.

7.5.2 Variations in sound pressure level along earmould-occluded earcanals

As in the unoccluded ear data there are large intersubject differences. Although the variation in SPL along the ear canal in Kemar and in real ears is in good agreement, within 1 dB below 3 kHz, the agreement at 4 kHz is only to within 4½ dB as Figure 7.7 shows.

For the one real ear in which measurements were made at greater depth the agreement with Kemar is excellent, within 1.2 dB, at all depths at all frequencies up to 4 kHz with no anomalies close to the eardrum. Further similar measurements with more subjects would be very useful if permission were to be obtained for measurements closer to the eardrum.

The comparison, Figure 7.8, of the absolute pressures generated under the earmould shows nearly a 4 dB difference between the mean of real ears and Kemar at 3 kHz and 4 kHz. This agreement is apparently not as close as that measured by Sachs and Burkhard, + 2 dB below 7 kHz [4, 1972]. But since the variance is large and the number of subjects is small the figure of 4 dB is not reliable — the 95% confidence interval of the mean is 3.2 dB at 3 kHz and 3.0 dB at 4 kHz. The zero line is not within the 95% confidence limit at either frequency however. Once again more real ear measurements would be required to be more precise.

7.5.3 Implications for hearing aid gain measurements

The real ear measurements are inherently less accurate than those on Kemar. But, given the closeness in agreement of pressure transfer ratios between Kemar and the group averaged data, and making the assumption that this agreement is maintained closer to the eardrum, then it should be possible with care to measure the insertion gain of a hearing aid with the probe microphone at the maximum permitted depth and obtain values within about 2 or 3 dB of the actual gain at the drum below 3 kHz, and to better than 4 or 5 dB at 4 kHz. The assumption appears reasonable given (i) probe measurements in the unoccluded real ears agree well with those in Kemar and with such real ear data as are available; (ii) the deep probe measurements in the occluded ear of the single subject closely agree with
those in Kemar; (iii) the variation in sound pressure levels in the occluded and unoccluded ear of Kemar and in real ears show no irregularities, and (iv) pressure transfer ratios measured in the unoccluded ear of Kemar agree closely with published real ear curves.

7.6 CONCLUSIONS

Mean sound pressure levels relative to the level at the canal entrance in unoccluded real ears at the maximum permitted depth (16 mm from canal entrance) agree with previous measurements on Kemar to within 2 dB at all test frequencies, 250 Hz to 6 kHz.

At each frequency the mean variation in sound pressure level in real ears between the canal entrance and the maximum depth also agrees well with that of Kemar. At 3 kHz and below the agreement is very good and within 2 dB at all depths, except for one frequency at one depth (2.7 dB at 8 mm, 3 kHz).

Measurements of SPL, relative to the level at the earmould tip position, in a single earmould-occluded real ear to a depth of 13 mm beyond the mould are, without exception, within 1.2 dB of the level at the same depth in Kemar at all frequencies up to 4 kHz. Group average data at up to 5 mm from the mould are within 1 dB of the Kemar value up to 3 kHz only. At 4 kHz a discrepancy of 4.1 dB occurs between Kemar and the group average of real ears.

Mean absolute sound pressure levels generated under the earmould in real ears at a depth of 5 mm beyond the mould are within 4 dB of the levels generated in Kemar under similar conditions at and below 4 kHz.

There were large intersubject differences even at frequencies as low as 1 kHz for both unoccluded and earmould-occluded ears and measurements with more subjects and at a greater depth would be useful.
## TABLE 7.1

**SOUND PRESSURE LEVEL VARIATION ALONG REAL UNAIDED EARCANALS**

(See also Figure 7.5)

<table>
<thead>
<tr>
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<th>std. deviation of absolute SPL dB</th>
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**Notes:**
(i) Std error = std deviation/2
(ii) 95% confidence interval = ± 1.6 × (std deviation) about the mean
FIG. 7.1 SCHEMATIC DIAGRAM OF EQUIPMENT USED TO MAP SPL IN UNAIDED, UNOCLUDED EARS

FIG. 7.2 SCHEMATIC DIAGRAM OF EQUIPMENT USED TO MAP SPL IN EARS OCCLUDED WITH EARMOULD S
FIG. 7.3 AN EARMOULD WITH A MICROPHONE PROBE TUBE IN ADDITION TO THE SOUND INLET TUBE.

- Inlet tube ends flush with mould.
- Projection, x, of probe tube beyond mould may be varied.
- Sound inlet tube, 2 mm i.d.
- Probe tube 1.2 mm i.d.
- Probe tube smeared with vaseline where it passes through mould to prevent acoustic leaks.
FIG. 7.4  SOUND LEVEL VARIATION ALONG UNAIDED EARCANALS: 4 SUBJECTS' LEFT EARS.
FIG. 7.5 SOUND LEVEL VARIATION ALONG UNAIDED EARCANALS: MEAN OF 4 REAL EARS CF. KEMAR.
FIG. 7.6 SOUND LEVEL VARIATIONS UNDER HEARING AID EARMOULD: LEFT EARS OF 4 SUBJECTS.
FIG. 7.7 SOUND LEVEL VARIATIONS IN AIDED EARS OCCLUDED WITH EARMOULDS: REAL EARS AND KEMAR (ZWISLOCKI EAR SIMULATOR)
FIG. 7.8 SPL GENERATED IN THE OCCLUDED EAR 5 mm BEYOND AN EARMOULD BY A HEARING AID RECEIVER: MEAN OF 4 REAL EARS CF. KEMAR/ZWISLOCKI EAR SIMULATOR.
FIG. 7.9 SOUND PRESSURE TRANSFER RATIO FROM THE EARCANAL ENTRANCE TO A POINT 12 mm INTO THE CANAL.

FIG. 7.10 SOUND PRESSURE TRANSFER RATIO FROM THE EARCANAL ENTRANCE TO A POINT 10 mm INTO THE CANAL.
CHAPTER 8

A COMPARISON OF LOUDNESS BALANCE, PROBE MICROPHONE AND ACOUSTIC REFLEX MEASUREMENTS OF HEARING AID GAIN

8.1 INTRODUCTION

In Chapter 4 the response of a behind-the-ear hearing aid was measured by loudness balance with normally hearing subjects, and on a 2 cc coupler, a Zwislocki-type ear simulator and on Kemar.

The loudness balance data were not in as good agreement with the manikin and the Zwislocki ear simulator data (with suitable allowance for head diffraction and ear canal resonance effects) as might have been expected from available literature [1, Zwislocki, 1970; 2, Sachs & Burkhard, 1972].

It was evident that such discrepancies would not be resolved without a further loudness balance experiment in which sound pressure levels generated in the aided and unaided ear were also measured, and that for as full an understanding as possible an acoustic reflex method of measuring aid gain such as that described by Tonisson [3, 1975] might also be useful.

Consequently the response of one behind-the-ear aid was measured using all methods and devices available to the author but excluding auditory threshold shift [see e.g., 4, Pascoe, 1975] since this would require subjects having thresholds above the internally generated noise of the hearing aid. These measurements are described and compared in this chapter. The methods used were:-

1. Loudness balance comparing aided and unaided real ears;
2. pressure measurements in aided and unaided real ears;
3. acoustic reflex threshold shift between aided and unaided real ears;
4. Kemar, aided and unaided;
5. measurements on 2 cc coupler, Zwislocki ear simulator, and IRPI XD-1053 2-branch ear simulator.

The first four methods listed are measures of insertion gain, the fifth is a measure of the transmission gain of the hearing aid.

64.
8.2 THE AID AND EARMOULDS

It was considered that a different aid from the one already tested (Chapter 4) should be used, in case the discrepancies found between the different measurements were peculiar to the individual aid tested.

The aid chosen and used for all tests was an Oticon BE11 behind-the-ear model [see 5, RNID, 1975 for test report] as issued by the National Health Service. The aid is shown in Figure 8.1. It contains a Knowles Electronics type BL-1681 ceramic microphone and a Knowles type BK-2606 magnetic receiver (N.B. in later versions of the BE11 aid the ceramic microphone is replaced with an electret microphone).

The gain control was fixed with sealing wax to give a transmission gain on a Zwischen ear simulator of about 37 dB at 1 kHz.

The earmoulds used with this aid were those incorporating a microphone probe tube in addition to the sound inlet tube (see 7.3.2).

8.3 REAL EAR MEASUREMENTS OF HEARING AID GAIN

8.3.1 Subjects

The same four otoologically and audiometrically normal male students who had had the sound fields in their ears mapped participated in each of these experiments. All bar one subject had also taken part in the loudness balance experiment described in Chapter 4. Subjects were paid at standard ISVR rates.

Only four subjects were involved since these experiments were originally planned as pilots for larger versions. However, these pilot experiments gave sufficient information to make further experiments of the same type unnecessary.

8.3.2 Loudness balance method and probe microphone method

As before (Chapter 4) subjects balanced the sound from a loudspeaker heard aided and unaided, against a reference signal. Unlike before, when the subject indicated that equal loudness had been obtained, the sound level in this test ear was measured with the probe microphone. The equipment used for both the loudness balancing and the probe microphone measurements is shown in Figure 8.2.

65.
The procedural details were as follows.

A pulsed acoustic reference signal was fed via an earmould into the subject's right ear from a Knowles XD-1005 receiver in a hearing aid shell. The receiver was driven directly by the loudness balance audiometer to produce a fixed sound level in the ear of 75 dB (as measured in a Zwislocki simulator).

Pulses of the same duration and frequency were fed to the loudspeaker, so that pulses heard by the subject alternated between the loudspeaker heard in the left ear and reference signal heard in the right ear. The sound level produced by the loudspeaker was under the subject's direct control. The loudspeaker was at mouth height, at 1 m from the subject's centre-of-head position, directly in front of him, as in the earcanal sound pressure mapping.

The subject adjusted the level of the signal from the loudspeaker, which he heard in his left ear, the test ear, to match in loudness the signal heard from the reference receiver in his right ear. This procedure was followed with and without the subject wearing the hearing aid in his left ear (i.e., "aided" and "unaided"). The loudspeaker driving voltage was noted for both aided and unaided conditions when the subject indicated equal loudness. When the aid was worn the loudspeaker voltage required for equal loudness to the reference was less than when the aid was not worn since the aid was amplifying the sound from the loudspeaker. The "subjective gain" of the aid in decibels was then defined as $20 \log \left( \frac{V_1}{V_2} \right)$ where $V_1$ is the voltage to the loudspeaker set by the subject for the unaided loudness balance and $V_2$ is the voltage to the loudspeaker set by the subject for the aided loudness balance.

The experiment was conducted at 4 frequencies, 500 Hz, 1 kHz, 2 kHz and 4 kHz with one aided and one unaided balance at each frequency by each subject. A balanced presentation order was used. Other experimental details of the loudness balance were as described in Chapter 4.

During both the aided and unaided balances the sound pressure generated in the subject's test-ear canal was monitored using the calibrated Knowles XL-9073 probe tube microphone. The probe-tube tip was positioned 5 mm beyond the earmould in the aided ear and in the corresponding position in the unaided ear, i.e., about 16 mm from the earcanal entrance.
To avoid physiological noise the probe microphone output was filtered to pass only the third-octave band containing the test frequency. Tests on Kemar showed that such filtering does not affect the measured level of the test stimuli.

The "pressure insertion gain" of the aid as measured by the probe microphone was defined as the excess sound pressure level in the aided ear over the unaided ear at equal loudness added to the "subjective gain" in decibels as defined above. This is equivalent to measuring the increase in sound pressure level in the ear canal when the aid is inserted, for a constant loudspeaker output.

8.3.3 Acoustic reflex threshold shift method

8.3.3.1 The acoustic reflex: a brief introduction

When the ear is subjected to intense sound levels above, say, 80 or 85 dB, two small muscles in the middle ear, the tensor tympani and the stapedius muscles, contract. This increases the rigidity of the chain of bones in the middle ear and also changes the motion of the stapes to a rocking motion, and consequently the transmission of sound from the outer to the inner ear is reduced slightly. Although the acoustic reflex is traditionally ascribed a protective function there is some doubt about its effectiveness at this, and the function is really as yet unknown [6, Lutman, 1976].

The reflex contractions may be detected qualitatively and quantitatively since the impedance at the eardrum is increased by the reflex and the change in admittance may be measured with an impedance bridge or oto-admittance meter. The reflex is bilateral, that is, sounds presented to one ear will elicit muscle contractions in both ears. This simplifies experimentation somewhat since it is possible to present reflex-eliciting sound stimuli to one ear and measure their effect on the other ear, thus the operation of the test ear is not impeded or modified by the presence of an oto-admittance meter, and the oto-admittance meter is not affected by the sound stimuli.

8.3.3.2 Experimental procedure

The equipment used is illustrated in Figure 8.3. The impedance probe of the oto-admittance meter was supported on a tripod alongside the
subject's head and sealed using a rubber cuff into the right ear of the subject. The seal was checked by raising the static pressure in the ear canal, holding this pressure briefly and ensuring from the pressure gauge of the oto-admittance meter that the pressure was maintained. The oto-admittance meter had been modified so that signals representing conductance and susceptance of the ear could be recombined with suitable phase relationship to provide an output signal proportional to the modulus of admittance. This output was displayed on a directly coupled oscilloscope and a zero offset applied electrically to the signal to remove any steady component of the signal so that the maximum amplification of the oscilloscope could be used. The scope sensitivity was increased until random fluctuations were readily apparent on the screen.

The subject's acoustic reflex threshold due to sound stimuli presented to the left ear was then detected at each test frequency with the left, test, ear aided and unaided. The same test frequencies (i.e., 500 Hz, 1 kHz, 2 kHz and 4 kHz) were used as in the loudness balance method, with one aided and one unaided reflex threshold determination at each frequency for each subject. The procedure adopted was to increase the level of the sound pulses from the loudspeaker until the action of the acoustic reflex was just detectable above the random movement of the scope trace.

This was found to be a reliable and repeatable method. Since the magnitude of the reflex response on the oscilloscope grew considerably for slight increases in sound level, the onset of the reflex was easily and consistently detectable to within 1 dB or 2 dB, provided a reflex was present.

The audiometer output corresponding to the acoustic reflex threshold was noted in each case in the same way that the output for equal loudness had been noted. The gain of the hearing aid in decibels at each test frequency was then defined as $20 \log \left( \frac{V_3}{V_4} \right)$ where $V_3$ is the minimum voltage driving the loudspeaker at which the acoustic reflex was just detectable in the unaided ear and $V_4$ is the voltage driving the loudspeaker at which the acoustic reflex was just detectable in the aided ear.

Tonisson [3] used a similar definition of hearing aid gain based purely on acoustic reflex thresholds. Judging from results obtained by
Pizarro [7, 1976] a more complicated procedure taking into account the growth of the acoustic reflex is an unnecessary complication, giving mean aid gains virtually identical to those obtained from straightforward reflex threshold determinations.

8.4 NON-REAL EAR MEASUREMENTS OF HEARING AID GAIN

8.4.1 Manikin method

The manikin Kemar dressed in a Tee-shirt was located with its head in the same position as the human subjects had been, 1 m from and facing the loudspeaker which was 64 mm below the horizontal reference plane for Kemar defined as by Burkhard and Sachs [8, 1975]. Both neck extension rings were used.

The gain of the aid was measured at discrete test frequencies as the SPL in the aided ear referred to the SPL in the unaided ear, in decibels using the 'eardrum' microphone of the Zwislocki ear simulator.

8.4.2 Ear simulators and couplers

Responses of the aid were measured in the B & K hearing aid test box using the two Zwislocki-type ear simulators, the IRPI XD 1053 two-branch ear simulator, and the B & K 2 cc coupler. In each case 40 mm of 2 mm internal diameter tubing linked the aid hook to the coupler cavity so as to reproduce the earmould tubing dimensions used in the tests on real ears.

8.5 RESULTS

Figure 8.4 shows the hearing aid response measured by the loudness balance for each individual subject together with the mean response. The comparative probe microphone measurements made simultaneously with the loudness balance are presented in Figure 8.5.

The acoustic reflex measurements were possible on 3 of the 4 subjects at 1 kHz and 2 kHz and on 2 of the subjects at 4 kHz. Measurements on the fourth subject or on any subject at 500 Hz were not possible since the sound levels required would have been excessive. Such measurements as
were possible are given in Figure 8.6 where they are compared with the probe microphone measurements from the previous figure for the same subjects.

The mean results from the above three real ear methods are collected together for comparison in Figure 8.7 which also shows the response of the test hearing aid measured on Kemar. Figures 8.8 and 8.9 illustrate the aid's response into the available ear simulators and couplers. The two Zwislocki-type ear simulators used gave essentially the same response, differing only by the thickness of the line on a level recorder plot apart from some low frequency irregularities below 500 Hz.

8.6 DISCUSSION

As Figure 8.7 shows, the acoustic reflex threshold shift and the probe microphone measurements of aid gain give very similar results, differing on average by less than 0.5 dB at 1 kHz and 2 kHz and by about 2 dB at 4 kHz when all available results for each method are included. The larger discrepancy at 4 kHz may include some uncertainty in the probe microphone measurements which were not at the eardrum position.

Results for individual subjects (Figure 8.6) also show good agreement between the two methods with individual quirks showing up in both. The tendency for the acoustic reflex to give very slightly higher gain values than the probe microphone is more pronounced when the probe measurements are averaged over only those subjects on whom the acoustic reflex measurements were possible. Ignoring any interaction with frequency the gain measured on this reduced number of subjects by acoustic reflex averages 3.0 dB higher than that measured by the probe, and a Wilcoxon matched-pairs signed-ranks test - which does not require the normality assumption of a t-test - shows that this tendency is significant at the 5% level. In fact apart from the results of one subject at 1 kHz, all acoustic reflex gain measurements were slightly higher than their respective probe measurement for the same subject at the same frequency.

Thus, the acoustic reflex and probe microphone gain measurements give detectably different though very similar results. The measured differences are most probably due to the probe microphone not being at the eardrum. The subjective loudness balance, however, gives results very different from the above two methods as can also be seen from
Figure 8.7. Mean sound levels measured in the aided ears were up to 17 dB higher than those in unaided ears at equal loudness making the gain measured subjectively less than that measured with the probe microphone except at 4 kHz, by up to 17 dB as illustrated in Figure 8.10. Results for individual subjects all followed this trend. A difference of this magnitude is rather surprising, especially when it encroaches so far up the frequency scale. However, similar though smaller differences of about 8 dB at frequencies below 1½ kHz were identified with a different BTE aid in the experiment in Chapter 4, although since probe measurements were not available in this previous experiment real ear sound pressures had to be estimated from head baffle and ear canal resonance corrections to coupler data, and from responses measured on Kemar.

In the present experiment the response of the BEll on Kemar is generally similar to the real ear response measured with the probe microphone, suggesting that the estimates of real ear sound pressures in Chapter 4 were fair.

Why then do the acoustic reflex method and the probe microphone method give similar values for the real ear gain of a hearing aid, while a loudness balance method can give values up to 17 dB lower? Alternatively, why should the sound pressure generated in the ear by a hearing aid need to be on average 17 dB greater than the sound pressure generated by a loudspeaker, for equal loudness? And why should the acoustic reflex agree with the pressure measurements rather than the subjective measurements?

Seventeen decibels is far too large to be a "Missing 6 dB" type of effect. The current consensus of opinion is that "Missing 6 dB" types of effect are explicable in terms of calibration or experimental artefacts [9, Killion, 1978; 10, Rudmose, 1962]. The suggestion has been made by Wever and Lawrence [11, 1954] that impedance mismatching between the air in the ear canal and the eardrum and middle ear increases as the entrapped air volume in the ear decreases, requiring higher sound levels in the ear to be produced by an earphone than by a loudspeaker for equal loudness. If this were so then presumably the acoustic reflex gain measurements would be more similar to the loudness balance than the probe tube measurements. The close agreement between the acoustic reflex and
the probe results suggests that the ear drum itself is responding to SPL in much the same way whether or not the ear is occluded and that the 17 dB difference mentioned is a subjective loudness-evaluation effect. The slight difference between acoustic reflex and the probe results suggests if anything that the acoustic reflex occurs at a lower SPL in an occluded ear but that this is only a matter of 3 dB or so.

The pathways and mechanisms of the acoustic reflex are illustrated schematically in Figure 8.11 which is adapted from Dallos [12, 1973]. From this it can be seen that the acoustic reflex is a peripheral brainstem response elicited at the level of the superior olive. As a peripheral response the triggering of the reflex will correlate with an easily detectable acoustic variable such as SPL since comparatively little processing will have been carried out on the neural signals from the cochlea. Loudness on the other hand is a judgement formed in the higher auditory centres of the brain and as a central judgement it will be influenced by more variables than just the SPL in the ear. Other variables may be acoustic, such as small amounts of noise or distortion (see below) or apparent proximity of the sound source, or non-acoustic cues, such as comfort of an earmould and the feeling of the presence of a mould in the ear.

8.6.1 Influence of aid-generated noise on loudness judgements

Scharf [13, 1964] has shown that the loudness of a pure tone in noise of one "critical band" bandwidth is similar to the loudness of a tone of the same SPL without the noise, except when the tone-to-noise ratio is very small. The critical bandwidth of noise for a pure tone is the band-width "beyond which any further increase in the width of the pass band of the noise has little or no influence on the amount of masking produced on a pure tone at the center of the band" [14, Kinsler and Frey, 1962].

Scharf's results for a 980 Hz tone in noise in the critical band centred on 980 Hz and with a bandwidth of 900-1060 Hz (half-power frequencies) are shown in Figure 8.12. Interpolating between Scharf's curves suggests that a 75 dB tone at 980 Hz presented in a critical band of noise with a 55 dB band level, will be judged to have the same loudness as a 75 dB tone presented in silence. If the noise level is increased to 65 dB, the loudness of the 75 dB tone will be reduced by 3 or 4 dB.
The critical bandwidth for a 1 kHz tone approximates very closely to the third-octave centred on 1 kHz. The noise level in the output of the BE 11 aid in this third-octave band was approximately 50 dB measured in the Zwischocki ear simulator, and similar levels would be expected in real ears [1; 2]. Subjects consistently set the level of the 1 kHz tone pulse (as measured with the probe microphone in the ear) in the aided loudness balance to greater than the nominal 75 dB reference intensity for equal loudness. Thus when the aid was judged equal in loudness to the reference receiver, the signal to critical-band-noise ratio of the aid was at least 20 dB, often nearer 30 dB.

The internally generated noise of the hearing aid would, judging from Scharf's data, seem insufficient to influence the results of the loudness balance judgements, at least at 1 kHz. However, judging from data presented by Zwicker [15, 1963] wide band noise will have a greater partial masking effect than a critical band of noise. The following experiment was therefore designed to determine any effects due to the noise of an aid.

8.6.1.1 Supplementary experiment: The effect of varying the reference level on the gain of a hearing aid as measured by loudness balance

The influence of the electrical noise generated by the hearing aid on the loudness balance results may be determined by varying the level of the reference signal. Since the noise from the aid remains constant this is equivalent to varying the signal-to-noise ratio.

The four subjects who had taken part in the main experiment were asked to return. Each made one aided and one unaided balance against the reference receiver at each of four levels of reference signal. The reference signal was presented at one frequency only, 1 kHz, and the four levels were 55 dB, 65 dB, 75 dB and 85 dB SPL as measured in a Zwischocki ear simulator.

Two subjects performed the aided balances before the unaided; the other two performed the aided balances first. The four levels of reference signal were balanced in order over the four subjects.

The results are shown in Figures 8.13-8.15.

In the unaided condition a linear regression analysis shows that each 10 dB increase in reference level elicited an average 8.3 dB increase
in the signal controlled by the subject. For the aided condition the corresponding average increase was 6.9 dB for every 10 dB increase in the reference level.

Both these slopes are less than 1:1. However, the results taken at the 55 dB reference level are dubious, as some readings of the output of the loudness balance audiometer fell below the lowest meter range. In fact a fixed 20 dB increase in signal had to be switched in in order to measure the signal in some cases. Furthermore, for the aided condition the internally generated noise of the aid which was about 50-55 dB in the third-octave band centred on 1 kHz would be comparable to the reference level.

If the 55 dB reference level results are not included in the regression the slopes for the 'unaided' and 'aided' balances become 0.93 and 0.78, respectively.

Thus the subjective gain of the aid has an apparent slight tendency to increase as the reference level is increased. The increase in subjective gain is about 1.3 dB on average for each 10 dB increase in reference level from 55 dB to 85 dB but is not statistically significant

\[ t = r \sqrt{(N - 2)}/\sqrt{(1 - r^2)} = 1.09; \quad p >> 25\% \].

Although the subjective gain of the aid changes only slightly when the reference signal level is varied between 65 dB and 85 dB, since the graphs of test signal against reference signal have slopes of less than 1:1 it is not possible to completely rule out partial masking as influencing the results. Since the graphs for both aided and unaided conditions have slopes of less than 1:1, there may be some cancellation of partial masking effects in calculating subjective gain, which may be more dependent on signal-to-noise ratios than is apparent.

The mean subjective gain of the aid measured with a 75 dB reference signal level is 16 dB which is within 2 dB of the value measured in the main experiment at the same frequency. This is considered to show a good test-retest reliability especially with so few subjects.
8.6.2 Influence of distortion on loudness judgements

Distortion as well as background noise may affect judgements of loudness. With a 1 kHz tone at 95 dB in the output of the test aid as measured in the Zwischen ear simulator, the main harmonic distortion components were 2% at 2 kHz and less than 1% at 3 kHz. Such low levels of distortion are unlikely to influence judgements of the loudness of the aid, certainly not to the extent of 17 dB.

According to Plomp [16, 1976] the loudness of a complex tone depends solely upon the overall sound pressure level if all components of the complex are within a critical bandwidth but increases if the components are more widely spaced. The loudness is almost independent of any harmonic relationships between the components. Thus a harmonically distorted signal would, if anything, sound louder than an undistorted one of the same sound level. So if harmonic distortion in the hearing aid were influencing the loudness judgements subjects would tend to set a lower SPL from the hearing aid than from the loudspeaker for equal loudness. This is contrary to the experimental findings.

Since tones were presented singly during the loudness balance intermodulation distortion would not arise. However, one other form of distortion which may affect loudness judgements is transient distortion, or a change in the rise time of the sound pulses as they are amplified by the aid. Measurements on Kemar showed no such effect. Waveforms of sound pulses at 1 kHz received in Kemar's aided and unaided ear, as acquired using the ISVR computer, are shown in Figure 8.16 and the leading and trailing edges of these pulses are shown in Figures 8.17 and 8.18. No change in rise time is detectable between the aided and unaided pulses. (No filtering was used except for a 22 Hz high pass. The ripple in the envelope of the waveforms is due to the sampling frequency of the computer beating with the tone pulse and was not present when the tone pulses were examined on an oscilloscope).

8.7 CONCLUSIONS

The subjective gain of a hearing aid measured using the loudness balance technique was up to 17 dB less than the gain measured simultaneously with a probe microphone on the same subjects. Since the measurement of gain by the acoustic reflex threshold shift method gives similar results
to the probe microphone it is likely that the ear itself responds similarly to the sound pressure level whether aided or unaided, and that the 17 dB discrepancy is psycho-acoustic rather than physical in origin. Available literature suggests that this psycho-acoustic effect is unlikely to be related to harmonic or transient distortion generated by the aid since measured values of these factors would be sufficiently small to have little or no effect on loudness judgements, though the background electrical noise generated by the aid may influence loudness judgements.
FIG. 8.1 THE BE11 HEARING AID.
FIG. 8.2 EQUIPMENT USED FOR THE LOUDNESS BALANCE WITH SIMULTANEOUS MEASUREMENT OF EARCANAL SPL.

FIG. 8.3 EQUIPMENT USED FOR THE ACOUSTIC REFLEX MEASUREMENTS.
FIG. 8.4 SUBJECTIVE GAIN OF THE BE11 AID MEASURED BY LOUDNESS BALANCE.

FIG. 8.5 PRESSURE INSERTION GAIN OF THE BE11 AID MEASURED WITH THE PROBE MICROPHONE.
FIG. 8.6 GAIN OF THE BE11 AID MEASURED ON INDIVIDUALS BY ACOUSTIC REFLEX AND PROBE MICROPHONE METHODS

FIG. 8.7 COMPARISON OF LOUDNESS BALANCE, ACOUSTIC REFLEX AND PROBE MIC. MEASUREMENTS OF INSERTION GAIN OF THE BE11 AID; MEANS OF ALL DATA.
FIG. 8.8 RESPONSE OF THE BE11 AID ON THE 2 cc COUPLER.

FIG. 8.9 RESPONSE OF THE BE11 AID ON THE ZWISLOCKI AND IRPI XD-1053 EAR SIMULATORS.
FIG. 8.10 THE DISCREPANCY BETWEEN PRESSURE INSERTION GAIN AND SUBJECTIVE GAIN OF THE AID.

FIG. 8.11 SIMPLIFIED BLOCK DIAGRAM OF THE MIDDLE EAR MUSCLE REFLEX AS A MULTIPATH FEEDBACK MECHANISM. After Dallos [12, 1973]
FIG. 8.12 PARTIAL MASKING OF A 980 Hz TONE AS A FUNCTION OF THE SPL OF THE MASKING NOISE. — After Scharf (17, 1964)

The masked tone is heard, at the stated level, in noise one critical band wide. The comparison tone heard in the quiet is adjusted to match the loudness of the masked tone. The loudness of the masked tone decreases as the SPL of the masking noise increases.
FIG. 8.13 LEVEL SET BY SUBJECTS TO MATCH LOUDNESS OF REFERENCE TONES FOR VARIOUS LEVELS OF REFERENCE TONE: AIDED LISTENING.
FIG. 8.14 LEVELS SET BY SUBJECTS TO MATCH LOUDNESS OF REFERENCE TONES FOR VARIOUS LEVELS OF REFERENCE TONE: UNAIDED LISTENING.
FIG. 8.15  SUBJECTIVE GAIN PLOTTED AS A FUNCTION OF THE LEVEL OF REFERENCE TONE
FIG. 8.16  WAVEFORMS OF 1kHz TONE PULSE MEASURED 'AIDED' AND 'UNAIDED' ON KEMAR
FIG. 8.17 LEADING EDGES OF PULSES SHOWN IN FIG. 8.16
FIG. 8.18 TRAILING EDGES OF PULSES SHOWN IN FIG. 8.16
CHAPTER 9

VARIABLES AFFECTING SUBJECTIVE GAIN: TWO EXPERIMENTS

9.1 INTRODUCTION

The experiment described in the previous chapter has shown that there is a difference in loudness between a loudspeaker heard direct and the same loudspeaker heard through a hearing aid at the same sound pressure level in the ear. This difference is psychoacoustic rather than physiological, the higher brain centres above the level of the acoustic reflex are making a distinction between the aided and unaided listening conditions.

This distinction may be based either on acoustic differences or on non-acoustic cues. A possible acoustic cue could be the self-generated circuit noise of an aid in the aided ear. Distortion products are too small to give an appreciable effect on loudness. A non-acoustic cue may be the feeling produced by the earmould occluding the aided ear or even the knowledge of the presence and usual function of a hearing aid.

But until the explanation of the loudness differences is found their real importance to hearing aid listening cannot be judged. Will, for example, comfortable listening levels, discomfort levels and sound "quality" be affected? Will higher sound levels than necessary be produced in the ear of an aid wearer to the detriment of speech discrimination?

Furthermore implications of these loudness differences spread wider than hearing aid research. Loudness balancing is an accepted method in research and in calibration transfers between earphones and has been suggested as a possible though not ideal method of calibrating ipsilateral impedance probes [see, for example, 1, Leis, 1978; 2, Leis and Lutman, 1978]. Will loudness differences as found between aided and unaided listening also apply to other applications of loudness balance? In short, under what conditions is a loudness balance valid?

The following questions arising from previous experiments were considered important.

Does the background noise level generated by an aid affect loudness judgement? Do loudness judgements differ among aids of different types or designs? Do loudness judgements vary with the gain setting of an aid?
Does the degree of occlusion of the ear affect loudness? Does the presence of an external soundfield, which may leak into the test or reference ear, matter? Is a hearing aid receiver driven directly from the loudness balance audiometer perceived any differently from the same receiver driven normally by the hearing aid amplifier?

To answer these questions the two related experiments described below were carried out.

9.2 AN INVESTIGATION INTO THE EFFECT ON SUBJECTIVE GAIN OF AID GAIN AND OF DIFFERENT MODELS OF HEARING AID

The aims of the experiment were (i) to determine whether the observed differences between subjective gain and the gain measured by probe microphone were constant for a particular aid or dependent on gain settings; (ii) to determine whether these differences were dependent on the particular model or design of behind-the-ear aid tested and hence (iii) to establish whether the self-generated noise from a hearing aid, which will vary with gain settings and between aids, is a possible explanation for any such differences.

The gains of various hearing aids were measured using the loudness balance technique and probe microphone measurements were made of the sound pressure levels generated in aided and unaided ears at equal loudness.

9.2.1 The aids used

Six aids in all were tested. Four were Oticon BE 11's each with its gain control sealed at a different gain setting. The gains were 12 dB, 20 dB, 28 dB and 36 dB at 1 kHz as set up and measured on Kemar before the experiment. Thus most of the available range of gains was covered in equal decibel increments.

The aid set to 28 dB gain had been modified for use in the next experiment as described below. The modification did not affect the aid's performance in this experiment.

The two other aids tested were an Alto Acoustics Ltd. "Focus W" and a Widex type 691 "Compact", as shown in Figures 9.1 and 9.2. These aids were also set to 28 dB gain at 1 kHz on Kemar.

All three types of aid used single-ended class A amplifier output stages feeding Knowles BK-series magnetic receivers. The receivers, a
BK-2604, a BK-2606 and a BK-2618 in the Alto, BE 11 and Widex, respectively, differ only in their electrical impedance, i.e., the number of turns on their coils [3, S.C. Ewens, 1978]. Any interaction with the ear would be virtually identical with each aid.

The Widex 691 and the BE 11 had similar Knowles BL series ceramic microphones (Widex, BL-1671; BE 11, BL-1681) whereas the Alto had a magnetic microphone (BJ-2591). Thus the microphone and initial amplifier stage of the Alto aid would differ from those in the Widex and BE 11, giving in particular a self-generated background noise slightly different in level and perhaps different in quality, whereas the receiver and interaction with the ear would be virtually identical for all the aids. (N.B. Later models of BE 11 have a Knowles electret microphone).

The microphones in the Alto and Widex are forward facing, that of the BE 11 is downward facing.

Figures 9.3-9.8 show frequency responses of these aids measured in the B & K test box on the 2 cc coupler and Zwislocki and IRPI XD-1053 ear simulators.

9.2.2 Experimental design and procedure

Six otologically and audiometrically normal subjects, five male and one female took part and were given the following instructions:-

INSTRUCTIONS

We would like you to help us compare various hearing aids. In your right ear you will be presented with a pulsed tone at a fixed level.

In your left ear you will hear a similar tone presented through a loudspeaker or hearing aid; this tone may be varied in intensity using the slider on your control box.

PLEASE ADJUST THE INTENSITY OF THE VARIABLE TONE IN YOUR LEFT EAR UNTIL IT IS EQUALLY LOUD TO THE FIXED TONE IN YOUR RIGHT EAR. When you are satisfied that the two tones are matched please tell me over the intercom, leaving the control slider in the same position until I tell you otherwise. The sounds may continue for
a few seconds while I make some measurements. I will then enter the room to change hearing aids, etc.

During the experiment please look directly in front of you at the loudspeaker and keep your head as still as you can, consistent with comfort. Please base your judgement solely on the pure tones and ignore any background noises which may be present from time to time.

An intercom will be on at all times in case you have any questions, or wish to discontinue the experiment.

Thank you.

Each subject made six aided and two unaided loudness balances during the course of a single visit lasting about 40 minutes. In each case the left ear was the test ear and the right ear the reference. All balances were made at a single frequency of 1 kHz with a 75 dB reference level as measured in the Zwislocki ear simulator. It was considered sufficient to test at 1 kHz only since previous experiments clearly show aided-unaided loudness differences to exist at that frequency, and if no explanation were to be found at that frequency it would be unlikely to be found at other frequencies. This frequency is also almost a standard for partial masking experiments and is well documented.

As in previous experiments both the audiometer output to the loudspeaker and the sound pressure level indicated by the probe microphone in the test ear were recorded when the subject indicated that equal loudness had been achieved.

Table 9.1 shows the experimental design. The six aided balances, one with each aid, were made as a group with their orders counter-balanced across subjects. The aided group was preceded and followed by an unaided balance. The aided balances were grouped to make the experiment more sensitive to differences between aids. These differences were considered more important than the absolute subjective gain of each aid. Grouping also gave the practical advantage that earmould fit-refit error could be avoided by leaving the mould in place, and enabled the same six subjects to participate in both this and the next experiment.

The XL-9073 probe microphone unit was damaged before the experiment and replaced with a Knowles BL-1685 unit. This is identical to the XL-9073 with the exception that it is supplied from stock rather than
being specially selected [4, R.J. Wilton, 1977]. When calibrated the
new unit's sensitivity was within 0.6 dB of the XL-9073's, but its
noise output was 6 dB (linear) and 13 dB(A) higher. This did not affect
the accuracy of any measurements.

The rest of the instrumentation was unchanged from the previous
experiment (8.3.2). Other experimental details such as earmould tubing
lengths, reference levels, and loudspeaker positioning were also unchanged.

9.2.3 Results

Table 9.2 shows the results of this experiment. As in previous
experiments the sound levels in aided ears are greater than the levels
required in unaided ears for equal loudness and the subjective gain of
each aid is therefore less than that measured by the probe microphone.
The sound level required in the aided ear varies with the aid type and
gain. Figures 9.9 and 9.10 show the variation in the sound level required
for equal loudness with the gain setting of the BE 11.

The difference between the subjective gain and pressure insertion
gain of the BE 11 is therefore also dependent on the aid's gain setting.
Figure 9.11 illustrates the relationship between the two measures of gain
for individual subjects. There are large differences between subjects.

When these large differences are averaged out by plotting mean sub-
jective gain against the mean pressure insertion gain for the BE 11's the
linear relationship of Figure 9.12 emerges and the data points of the
Alto and Widex aids also fall close to the line. The slope of the line
is about 0.65-0.70 with a correlation coefficient of 0.9987 for the four
BE 11's or 0.9856 for all the points in Figure 9.12.

Table 9.2 also shows that the real ear pressure insertion gain
measured with the probe microphone is very similar to the gain measured on
Kemar in the case of each BE 11 but the real ear pressure insertion gains
of the two aids with forward facing microphones are both about 5 dB lower
than the respective gains measured on Kemar.

This suggests that the sound field about Kemar's head may be a
slightly poorer duplication of the sound field around a median real head
at the location of a forward facing microphone than at the location of a
downward facing microphone. This point is discussed below (10.2.4).
9.2.4 Discussion

It would be tempting to infer from Figures 9.11 and 9.12 that a linear gain dependent relationship exists between subjective gain and pressure insertion gain which holds for all the aids tested irrespective of circuit design, microphone position or the particular transducers used. Such a suggestion however is not reconcilable with the experimental method. It would require each subject to be able to estimate the gain of each aid and set the sound level in his ear accordingly. Since he is in an anechoic room listening simply to pure tones at an unknown level the subject has no way of doing this.

Subjects do however set different sound levels from different aids to give equal loudness. As far as a subject is concerned the only obvious difference between the aids is in the background noise level they produce in the aided ear. In Figure 9.13 the sound level set in the ear for a given loudness is plotted against the background noise level for each aid as measured in the Zwischocki ear simulator. The noise figures used are A-weighted since this weighting is generally accepted to correlate reasonably well with loudness judgements.

Again a straight line appears to fit the data well. But this time the Widex and Alto aids are not displaced from the line set by the four BE 11's as they were in Figure 9.12. All the aids are on the line and the correlation coefficient for the data of all the aids is 0.9723.

In other words the most reasonable explanation bearing in mind the subjects' task is also the explanation which best fits the experimental data. That is that the sound level required in the aided ear at equal loudness increases with the noise level in the ear; the test tones are being partially masked by the noise generated within each hearing aid, making them sound quieter.

The noise level increases with the gain of the aid and is similar for the Widex, Alto and BE 11 aids when they have the same real ear pressure insertion gain, as shown in Figure 9.14. Consequently the impression is given that the sound level set in the ear is a function of aid gain and that subjective gain varies with the pressure insertion gain, which indeed it does, but only because the noise levels vary with the pressure insertion gain.
In fact Figure 9.12 can be augmented by adding two further points to represent aids tested in previous experiments - the Widex Baritone (Chapter 4) and a BE 11 (Chapter 8). This gives Figure 9.15. The value of this exercise however lies only in demonstrating that, since these aids also lie on the line, the noise level produced by the Baritone is probably similar to that produced by a BE 11 with a similar gain. The pressure insertion gain used for the Baritone is incidentally an estimate based on measurements on Kemar.

9.2.4.1 Comparisons with published data on partial masking

The results discussed above suggest that during aided listening the loudness balance test tones may be partially masked by the noise generated within the aid. Noise levels in the reference ear and in the unaided ear from the loudspeaker were too low to measure in the Zwislocki ear simulator with the standard B & K 4134 microphone and were subjectively audible only with very careful listening in the quiet of the anechoic room.

The loudness balance therefore is essentially between the tone in noise and a tone in quiet, or in noise very near threshold.


Whilst different psychophysical functions have been fitted to the experimental data in these papers, the data themselves are generally similar. The results of Zwicker and of Hellman and Zwislocki shown in Figures 9.16 and 9.17, respectively, are typical. A tone in noise sounds quieter than the same tone in quiet.

Zwicker's data shows that a 1 kHz tone at 75 dB SPL (the reference level used in the present experiment) sounds equal in loudness to a tone of nearly 80 dB SPL or a tone of 83 dB SPL in wide-band noise at levels of, respectively, 40 dB and 60 dB SPL per critical band.

Hellman and Zwislocki quote sensation levels (SL in dB re: threshold) rather than sound pressure levels (SPL in dB re: 20 µPa). A 1 kHz tone
at 75 dB SL in quiet sounds equally loud to an 80 dB SL tone in 40 dB SL masking noise or an 85 dB SL tone in 60 dB of masking noise. Assuming that 0 dB SL is about 10 dB SPL at 1 kHz then a 1 kHz tone at 75 dB SPL would be equally loud to an 83 dB SPL tone and a 91 dB SPL tone in the two masking noise levels, which presumably are about 50 dB and 70 dB SPL in the critical band around 1 kHz.

To compare the results of the present experiment with the partial masking data cited above it is convenient to assume that the critical band centred on 1 kHz is approximately one third of an octave wide (8.6.1). The noise levels of each aid measured in the Zwislocki ear simulator in this critical band can therefore be read directly from Table 9.2.

The levels of pure tone required in the noise level generated by each aid to equal the 75 dB SPL reference tone in (approximate) quiet are in reasonably good agreement with the published data cited above or perhaps 3-4 dB higher. Given the assumptions made in comparing the results, the fact that noise spectra were different, that noise levels were measured in different ways with different equipment and different subjects and that in the references cited considerable care was taken to eliminate any bias by alternating the tone in quiet with the tone in noise as the adjustable tone, then any closer agreement could not reasonably be expected.

It is therefore considered that the changes in the sound level required in the ear for a given loudness may be entirely explained by partial masking and that results are consistent with published partial masking data.

9.3 AN INVESTIGATION INTO THE EFFECT OF THE METHOD OF PRESENTING TEST SIGNALS ON THE LOUDNESS-INTENSITY RELATIONSHIP

The aim of this experiment was to determine what variables other than aid gain setting would affect the relationship between loudness and the sound pressure level in the ear, and hence what variables influence the subjective gain of a hearing aid. The variables considered were

(i) the occlusion or non-occlusion of the test ear,
(ii) the presence or lack of the external sound field around the head from the loudspeaker during aided listening,
(iii) and the replacement of the loudspeaker/aid microphone/aid amplifier chain with a direct-wired link between the loudness balance audiometer and the hearing aid receiver.

This experiment was conducted simultaneously with, but separately from, the previous experiment.

Six subjects performed loudness balances between the adjustable level test signals presented in six different ways to the left ear and the fixed level reference presented via the standard reference receiver to the right ear. The sound pressure level was measured in the test ear at equal loudness.

9.3.1 Signal presentation conditions

The pulsed pure tone test signal was presented to the left ear of each subject in six different ways.

These were:

(i) from a loudspeaker at 1 m, the usual "unaided" condition;
(ii) via a BE 11 hearing aid and earmould, with the loudspeaker as a sound source, i.e., the usual "aided" condition: the aid was set to 28 dB gain at 1 kHz on Kemar;
(iii) from the receiver and earmould of the BE 11 of condition (ii), with the receiver driven directly from the loudness balance audiometer eliminating the need for the loudspeaker or aid microphone and amplifier (see below);
(iv) via a BE 11 aid and earmould but using a second hearing aid receiver rather than the loudspeaker as a sound source. The second receiver was connected via 1.2 mm diameter polythene tubing to the aid microphone as shown in Figure 9.18. This aid had to be used on a low gain setting to prevent acoustic feedback;
(v) as condition (iii) but replacing the earmould with a shaped tube of the same dimensions as the sound tube through the mould. The tube ended inside the ear canal but the canal was not occluded. (A rather extreme form of venting.)
(vi) As condition (iv) but with the sound tube only rather than the complete earmould.
A seventh condition consisting of the standard aided condition (ii) but with the sound tube replacing the complete earmould was not surprisingly impossible due to acoustic feedback.

All earmoulds contained a microphone probe tube in addition to the sound input tube and were lightly vaselined to provide a good seal.

The BE 11 aid used for the standard aided condition (ii) was modified so that a signal could be injected directly to its receiver from the loudness balance audiometer for condition (iii). This aid is shown in Figure 9.19. The modification consisted of breaking the leads from the amplifier to the receiver and bringing the ends out to a socket mounted next to the gain control as illustrated schematically in Figure 9.20a. The aid could be restored to normal operation by linking pins 2 and 3, or an externally generated signal could be fed to the aid receiver from the loudness balance audiometer using the circuit shown in Figure 9.20b.

The battery and resistor in this circuit produce a D.C. bias across the receiver which would normally be supplied by the hearing aid amplifier, whilst the capacitor isolates the battery circuit from the audiometer output stage. The receiver is designed to work with optimum linearity with this bias, about 200 mV D.C., superimposed on the audio signal.

In this experiment the bias across the receiver was 204 mV D.C., although at the sound output levels produced by the aid during the experiment the receiver was within its linear operating range even without the bias. Removing the bias affected the sound level output at 1 kHz by less than 0.1 dB and gave no measurable change in the levels of second or third harmonic distortion. The bias was always used nevertheless.

The receiver used as a sound source to drive the aid microphone in conditions (iv) and (vi) was of a different type and no D.C. bias was necessary.

9.3.2 Method

The six subjects who participated in the experiment described previously (9.2) also took part in this one and were given the same instructions as before. The equipment used is shown in Figure 9.21.
The standard pulsed reference tone was presented to the subject's right ear at a level of 75 dB measured in the Zwislocki ear simulator and at a frequency of 1 kHz. The comparison test tone was fed to the subject's left, test, ear via one of the six methods, (i)-(vi) and the subject adjusted the level of the comparison signal until it matched in loudness the reference signal in his right ear. This procedure was repeated with another signal presentation method and so on till all six methods had been tested. The counter-balanced presentation order of the six methods is shown in Table 9.3.

Once the subject indicated that equal loudness had been achieved for the particular presentation method being used, the output of the loudness balance audiometer was noted and the sound pressure level of the signal in the test ear was measured with the probe microphone.

Each subject attended for one session only of about 40 minutes.

9.3.3 Results

The mean sound pressure levels measured in the ear for each method of signal presentation when judged equally loud to the reference signal are listed in Table 9.4. Also given are the levels of statistical significance in percent of the differences between these sound levels as determined by Student's t-tests.

The mean sound level required in the test ear for a given loudness varies, according to the method of presenting the signal, from 72.4 dB to 84.8 dB. This is a range of 12.4 dB, with the extreme levels differing at the 1% level of significance \( t = 5.07; p < 1\% \). A range of this size could not be due to any calibration or measurement artefact of the probe microphone.

Ranking the methods of test signal presentation in decreasing order of sound levels required in the ear for equal loudness gives the following order:-

1: normal aided condition with earmould, condition (ii)
2: complete aid with mould, driven from second receiver; condition (iv)
3: receiver of BE 11 driven from audiometer, with earmould; condition (iii)
4: normal unaided condition, loudspeaker; condition (i)
5: receiver of BE 11 driven from audiometer, with sound tube only;
   condition (v)
6: complete aid, sound tube only, driven from second receiver;
   condition (vi).

9.3.4 Discussion

The rank order immediately above shows that the signal presentation
conditions in which the ear is fully occluded all require a greater
sound pressure level for equal loudness than the conditions in which the
ear is not occluded. In particular a comparison is possible in two
instances between presentation conditions which are identical apart
from the presence or absence of the full earmould. These two instances
are the BE 11 receiver driven directly from the audiometer and the complete
aid driven from the second receiver. In both cases the sound level with
the mould exceeds the sound level without the mould, by 5.9 dB and 7.9 dB
respectively, as Table 9.4 shows.

Both sound level differences are significant at a level between 5% and
2%, and are coincidentally close in magnitude to a "missing 6 dB
effect". Sound level differences of this order are real and could not
be artefacts of the probe microphone calibration or measurements, a
reason often invoked to explain the missing 6 dB as pointed out by
Killion [15, 1978]. Furthermore with both the directly driven receiver
and the complete aid driven from a second receiver there is no free
sound field around the head and therefore no possibility of leakage of
such a sound field into either the test or reference ear. There is no
question therefore of the external sound field influencing or being
responsible for the change in loudness between a fully occluded and an
open ear condition.

It is possible however that these differences are influenced by the
noise levels in the test ear varying and giving partial masking. This
aspect is discussed below.

A further observation arising from Table 9.4 is that all the condi-
tions in which the ear is not occluded require sound levels within a
2.5 dB range and do not differ significantly. In one case tones are
originating 1 m from the subject and in the other cases tones originate
in the unoccluded ear canal. Thus loudness of the test tones at a
given SPL in the ear is independent of the distance of the sound source from the ear. Some subjects did comment that they could not distinguish sounds coming down the sound tube from sounds originating at the loudspeaker.

9.3.4.1 An explanation in terms of partial masking

Superficially two distinct effects appear to be operating in this experiment. Firstly an "earmould effect" whereby fully occluding the ear canal with an earmould causes a 6-8 dB greater sound pressure to be required in the ear for a given loudness. Secondly, a "hearing aid effect" which also causes a greater sound pressure level to be required for a given loudness when a complete aid is used. The complete aid with earmould driven from the second receiver was set to a lower gain than the complete aid with earmould driven from the loudspeaker and also required a lower sound pressure level in the ear for a given loudness. The "hearing aid effect" is therefore consistent with the gain dependent effect found in the previous experiment.

The "hearing aid effect" may therefore be explained in terms of partial masking of the loudness balance tones by the circuit noise generated within the hearing aids. Is it possible that the "earmould effect" could also be due to "partial masking"?

Just as increasing the gain of a hearing aid will increase the level of circuit noise in the ear, so will sealing the ear by replacing a sound tube with a full earmould. Any noise energy emanating from a hearing aid or receiver is then confined to the smaller volume of the sealed ear canal and the noise level produced at the eardrum will necessarily be increased. Because of partial masking this increase in noise level at the drum will necessarily lead to an increase in the level of the loudness balance tones required for a given loudness.

Thus sealing the ear with an earmould will necessarily lead to an increase in the sound level required of the test tones for equal loudness if any circuit noise is audible in the ear. The "earmould effect" would thus be expected as a consequence of partial masking. Furthermore, it is possible to deduce the change in circuit noise level on sealing the ear from Table 9.5 and hence estimate by how much the required sound level will be raised by the "earmould effect".
Comparing conditions (iii) and (iv) in Table 9.5 shows that the signal voltage at 1 kHz required by the BE 11 receiver to produce 75 dB SPL in the ear was on average (47.2 - 25.7 =) 21.5 dB less when the ear was occluded with an earmould than when the sound tube only was present. Conversely if the signal voltage were held constant, sealing the ear with the earmould would cause at 21.5 dB increase in the sound level of the tone in the ear. Any circuit noise emanating from the receiver would similarly increase in level along with the signal, and for those frequencies around 1 kHz the increase would be 21.5 dB. At other frequencies the increase may be slightly more or less, but the frequencies around 1 kHz, more precisely in the 1 kHz critical band, are the most important here.

But whilst the subject can turn down the signal voltage when his ear is sealed to get back to equal loudness, the circuit noise voltage will remain constant. Thus there would have been a 21.5 dB higher level of circuit noise from the BE 11 receiver when a full earmould was used instead of a sound tube only. The increase was clearly audible subjectively.

A similar calculation for the complete aid driven from the second receiver shows that there was a 23.5 dB increase in circuit noise in the ear in this case when the sound tube was replaced by a full earmould.

Table 9.4 shows that these two increases of 21.5 dB and 23.5 dB in circuit noise in the ear were accompanied by measured increases in the signal level required for equal loudness of 5.8 dB and 7.9 dB, respectively. If these two increases in circuit noise had been produced by increasing the gain of a hearing aid then from the linear relationship shown in Figure 9.13 we would predict respectively a 7.0 dB and a 7.6 dB increase in required signal level. The figures are in very good agreement; a change in circuit noise level in the ear therefore changes the required tonal sound level for a given loudness by the same amount whether the change in circuit noise is caused by increasing aid gain or by sealing the ear.

It is therefore considered that the "earmould effect" as well as the "hearing aid effect" may be fully explained by partial masking of the loudness balance test tones in the test ear by circuit noise. The "earmould effect" and the "hearing aid effect" are different manifestations of the same effect. In Chapter 8 it was demonstrated that there was

90.
no acoustical basis for any "closed ear" or "missing 6 dB" effect. The experiments in this chapter suggest there is no psychological basis either and that, if background masking noise were eliminated from the test ear, loudness and sound pressure level at the ear drum would have the same relationship whether the ear were occluded or not. This is completely in accord with current thinking as propounded by Rudmose [16, 1963] and confirmed recently by Killion [15].

9.3.4.2 Rough estimates of noise levels

Rough estimates of average circuit noise levels in the ear for the different signal presentation conditions may be derived from Table 9.4 and Figure 9.13. Using the values in Table 9.4 as the ordinates in Figure 9.13, the corresponding abscissae are then the approximate noise levels in the ear. Thus for the directly driven receiver with earmould, condition (iii), the noise level would be about 35 dB(A) or more precisely, the circuit noise coming from the receiver would have the same partial masking effect as a noise level of 35 dB(A), which is equivalent to 27 dB third octave band level at 1 kHz, produced by a BE 11 hearing aid.

The reference level in these experiments was 75 dB SPL. Entering the graph with this value suggests a rough value of the noise level in the reference ear of 25 dB(A) or 17 dB in the 1 kHz third octave band. These figures are best regarded as not very approximate since many assumptions are involved.

9.3.4.3 Comparisons of sound pressure levels in real ears and Kemar

The output of the loudness balance audiometer was noted for each presentation method when the sound level in the real ears was measured. The sound pressure level produced in the ear for a given audiometer output by each method of signal presentation can therefore be calculated and compared with the sound level produced in Kemar's ear under the same conditions. Table 9.5 already referred to shows the results of such an exercise.

The sound pressure levels generated in Kemar's ear are slightly lower than the mean levels generated in real ears but only by between 0.6 dB and 3.1 dB. This small range of differences is a further indication that the probe microphone is reliably measuring sound pressure in the ear.

91.
regardless of the method of presenting the signal. The measured differences between sound levels generated by the same method in the mean real ear and in Kemar's ear will be due mainly to real differences between real ears and Kemar's ear simulator, which would be expected with a small number of subjects and in part to not making measurements at the eardrum in the real ears. The variation in these differences between different methods of presentation is likely to be due, at least in part, to small but inevitable systematic variations in the position of the probe tube tip with the different methods.

9.4 SUMMARY AND CONCLUSIONS TO CHAPTER 9

The first experiment described in this chapter (9.2) showed that as the gain of a hearing aid was increased a higher sound pressure level was required in the ear to produce the same sensation of loudness. Each increase in gain of 10 dB as measured by a probe microphone in the ear required a 3\frac{1}{2} dB greater sound pressure level in the ear for equal loudness. Consequently the subjective gain of an aid expressed in decibels increase linearly with the pressure insertion gain but at only 65\% of the rate.

The second experiment (9.3) revealed two aspects of the difference in the loudness of a given sound pressure level between aided and unaided ears. Firstly occluding the ear with an earmould necessitated a 6 dB increase in the sound pressure level in the ear for a given loudness; and secondly when a complete aid was used there was a further increase in the sound level required for a given loudness. The increase in sound level was linearly dependent on the aid gain, as in the previous experiment.

It was demonstrated that all these effects were explicable by partial masking of the loudness balance tones in the test ear. The degree of partial masking depended on the background noise level in the test ear which varied with hearing aid gain and according to whether or not the ear was occluded.
<table>
<thead>
<tr>
<th>Subject No.</th>
<th>1st</th>
<th>2nd</th>
<th>3rd</th>
<th>4th</th>
<th>5th</th>
<th>6th</th>
<th>7th</th>
<th>8th</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Unaided</td>
<td>Widex</td>
<td>BE11-3</td>
<td>BE11-2</td>
<td>Alto</td>
<td>BE11-1</td>
<td>BE11-4</td>
<td>Unaided</td>
</tr>
<tr>
<td>2</td>
<td>Unaided</td>
<td>BE11-3</td>
<td>Alto</td>
<td>Widex</td>
<td>BE11-4</td>
<td>BE11-2</td>
<td>BE11-1</td>
<td>Unaided</td>
</tr>
<tr>
<td>3</td>
<td>Unaided</td>
<td>Alto</td>
<td>BE11-4</td>
<td>BE11-3</td>
<td>BE11-1</td>
<td>Widex</td>
<td>BE11-2</td>
<td>Unaided</td>
</tr>
<tr>
<td>4</td>
<td>Unaided</td>
<td>BE11-4</td>
<td>BE11-1</td>
<td>Alto</td>
<td>BE11-2</td>
<td>BE11-3</td>
<td>Widex</td>
<td>Unaided</td>
</tr>
<tr>
<td>5</td>
<td>Unaided</td>
<td>BE11-1</td>
<td>BE11-2</td>
<td>BE11-4</td>
<td>Widex</td>
<td>Alto</td>
<td>BE11-3</td>
<td>Unaided</td>
</tr>
<tr>
<td>6</td>
<td>Unaided</td>
<td>BE11-2</td>
<td>Widex</td>
<td>BE11-1</td>
<td>BE11-3</td>
<td>BE11-4</td>
<td>Alto</td>
<td>Unaided</td>
</tr>
</tbody>
</table>

Widex = Widex 691 aid
Alto = Alto Focus
BE11-1 = BE11 aid set to 12 dB gain on Kemar
BE11-2 = BE11 aid set to 20 dB gain on Kemar
BE11-3 = BE11 aid set to 28 dB gain on Kemar
BE11-4 = BE11 aid set to 35.5 dB gain on Kemar

Subjects were assigned a number from 1-6 at random.
### TABLE 9.2
RESULTS OF THE EXPERIMENT DESCRIBED IN SECTION 9.2

(a) Aided and unaided sound pressure levels in the ear at equal loudness

<table>
<thead>
<tr>
<th>Unaided (mean of 2 replicates per subject)</th>
<th>Aided</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>BE11-1</td>
</tr>
<tr>
<td>SPL in ear at equal loudness mean of 6 subjects dB</td>
<td>72.1</td>
</tr>
<tr>
<td>(std dev. dB)</td>
<td>(4.7)</td>
</tr>
</tbody>
</table>

(b) Comparison of gain measurements, together with noise and distortion figures of the different aids

<table>
<thead>
<tr>
<th></th>
<th>BE11-1</th>
<th>BE11-2</th>
<th>BE11-3</th>
<th>BE11-4</th>
<th>Widex 691</th>
<th>Alto 'Focus'</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gain measured on Kemar, dB</td>
<td>12.0</td>
<td>20.0</td>
<td>28.0</td>
<td>35.5</td>
<td>28.0</td>
<td>28.0</td>
</tr>
<tr>
<td>Gain measured by probe mic. mean of 6 real ears, dB</td>
<td>11.4</td>
<td>19.0</td>
<td>27.4</td>
<td>33.2</td>
<td>22.9</td>
<td>23.5</td>
</tr>
<tr>
<td>(std. deviation, dB)</td>
<td>(4.4)</td>
<td>(4.9)</td>
<td>(3.5)</td>
<td>(6.0)</td>
<td>(4.0)</td>
<td>(3.4)</td>
</tr>
<tr>
<td>Subjective gain mean of 6 real ears, dB</td>
<td>3.9</td>
<td>9.7</td>
<td>14.9</td>
<td>19.8</td>
<td>10.5</td>
<td>13.6</td>
</tr>
<tr>
<td>(std. deviation, dB)</td>
<td>(5.1)</td>
<td>(4.2)</td>
<td>(3.7)</td>
<td>(3.9)</td>
<td>(3.2)</td>
<td>(4.2)</td>
</tr>
<tr>
<td>Self generated noise dB SPL measured in Zwislocki ear simulator</td>
<td>&lt;47</td>
<td>&lt;47</td>
<td>54</td>
<td>61</td>
<td>53</td>
<td>50</td>
</tr>
<tr>
<td>1kHz, 1/3-8ve</td>
<td>39</td>
<td>44</td>
<td>53</td>
<td>60</td>
<td>52</td>
<td>46</td>
</tr>
<tr>
<td>Harmonic distortion measured in Zwislocki ear simulator</td>
<td>2nd</td>
<td>0.7%</td>
<td>0.3%</td>
<td>0.3%</td>
<td>0.5%</td>
<td>0.3%</td>
</tr>
<tr>
<td>3rd</td>
<td>0.8%</td>
<td>0.6%</td>
<td>0.9%</td>
<td>0.3%</td>
<td>0.1%</td>
<td>0.6%</td>
</tr>
</tbody>
</table>
### TABLE 9.3

**Experimental design for the experiment described in section 9.3**

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>1st</th>
<th>2nd</th>
<th>3rd</th>
<th>4th</th>
<th>5th</th>
<th>6th</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>i</td>
<td>ii</td>
<td>vi</td>
<td>iii</td>
<td>v</td>
<td>iv</td>
</tr>
<tr>
<td>2</td>
<td>ii</td>
<td>iii</td>
<td>i</td>
<td>iv</td>
<td>vi</td>
<td>v</td>
</tr>
<tr>
<td>3</td>
<td>iii</td>
<td>iv</td>
<td>ii</td>
<td>v</td>
<td>i</td>
<td>vi</td>
</tr>
<tr>
<td>4</td>
<td>iv</td>
<td>v</td>
<td>iii</td>
<td>vi</td>
<td>ii</td>
<td>i</td>
</tr>
<tr>
<td>5</td>
<td>v</td>
<td>vi</td>
<td>iv</td>
<td>i</td>
<td>iii</td>
<td>ii</td>
</tr>
<tr>
<td>6</td>
<td>vi</td>
<td>i</td>
<td>v</td>
<td>ii</td>
<td>iv</td>
<td>iii</td>
</tr>
</tbody>
</table>

**Key:**

- **i:** Unaided balance - loudspeaker direct
- **ii:** Aided, BEll with earmould - input from loudspeaker
- **iii:** Receiver driven direct from audiometer, with mould
- **iv:** Aided, BEll with mould - input from receiver coupled to microphone
- **v:** As 3, but without earmould - sound tube only
- **vi:** As 4, but without earmould - sound tube only

The same subjects were assigned the same numbers as in the experiment described in section 9.2.
TABLE 9.4
RESULTS OF THE EXPERIMENT DESCRIBED IN SECTION 9.3

Sound pressure levels in the ear for equal loudness of the signal presented in various ways.

<table>
<thead>
<tr>
<th>signal presentation method</th>
<th>(i)</th>
<th>(ii)</th>
<th>(iii)</th>
<th>(iv)</th>
<th>(v)</th>
<th>(vi)</th>
</tr>
</thead>
<tbody>
<tr>
<td>normal unaided condition, loudspeaker; condition (i)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>normal aided condition, with earmold; condition (ii)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>BEll receiver driven from audiometer, with earmold; condition (iii)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>complete aid with earmold driven from second receiver; condition (iv)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>BEll receiver, driven from audiometer, sound tube only, condition (v)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>complete aid, sound tube only, driven from second receiver, condition (vi)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Sound pressure levels in ear for equal loudness, dB SPL</th>
<th>mean</th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(S.D.</td>
<td>(6.47)</td>
<td>(2.08)</td>
<td>(5.36)</td>
<td>(4.57)</td>
<td>(4.32)</td>
</tr>
<tr>
<td></td>
<td>n=6</td>
<td>74.9</td>
<td>84.8</td>
<td>78.7</td>
<td>80.3</td>
<td>72.9</td>
</tr>
</tbody>
</table>

Level of statistical significance of the differences between sound pressure levels (Student's t-test)

(NS = Not Significant)
<table>
<thead>
<tr>
<th>Signal Presentation Method</th>
<th>(i)</th>
<th>(ii)</th>
<th>(iii)</th>
<th>(iv)</th>
<th>(v)</th>
<th>(vi)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal unaided condition, loudspeaker, condition (i)</td>
<td>Normal aided condition, with earmould, condition (ii)</td>
<td>( \text{BE} \text{ II receiver driven from audiometer, with earmould, condition (iii)} )</td>
<td>Complete aid with earmould driven from second receiver, condition (iv)</td>
<td>( \text{BE} \text{ II receiver driven from audiometer, sound tube only, condition (v)} )</td>
<td>Complete aid, sound tube only, driven from second receiver, condition (vi)</td>
<td></td>
</tr>
</tbody>
</table>

| Audiometer output \( \text{dB re arbitrary voltmeter reference} \) | 25.5 | -3.1 | 25.7 | 25.4 | 47.2 | 48.9 |
| Mean SPL generated in real ears, \( \text{dB SPL at stated audiometer output} \) | 75.0 | 75.0 | 75.0 | 75.0 | 75.0 | 75.0 |
| SPL generated in Kemar's ear, dB SPL at stated audiometer output | 72.5 | 71.9 | 73.2 | 72.4 | 73.7 | 74.4 |
| SPL in real ears - SPL in Kemar's ear, dB (same audiometer output) | 2.5 | 3.1 | 1.8 | 2.6 | 1.3 | 0.6 |
FIG. 9.3  FREQUENCY RESPONSES OF THE BE11 AIDS ON THE 2 cc COUPLER

FIG. 9.4  FREQUENCY RESPONSE OF BE11 N°3 ON THE ZWISLOCKI EAR SIMULATOR
FIG. 9.5  FREQUENCY RESPONSE OF THE ALTO FOCUS ON THE 2 cc COUPLER

FIG. 9.6  FREQUENCY RESPONSE OF THE ALTO FOCUS ON THE ZWISLOCKI AND XD 1053 EAR SIMULATORS
**FIG 9.7** FREQUENCY RESPONSE OF THE WIDEX 691 ON THE 2cc COUPLER

**FIG 9.8** FREQUENCY RESPONSE OF THE WIDEX 691 ON THE ZWISLOCKI AND XD 1053 EAR SIMULATORS
Fig. 9.9 Variation in the SPL giving equal loudness in aided ears with the aid gain; individual subjects.

Fig. 9.10 Mean variation in the SPL giving equal loudness in aided ears with mean aid gain.
FIG. 9.11 SUBJECTIVE GAIN vs. PRESSURE INSERTION GAIN FOR THE BE11 AID; INDIVIDUAL SUBJECTS.

FIG. 9.12 MEAN SUBJECTIVE GAIN vs. MEAN PRESSURE INSERTION GAIN FOR ALL AIDS IN THIS EXPERIMENT (9.2).
FIG. 9.13 THE RELATIONSHIP BETWEEN THE MEAN SPL IN THE AIDED EAR AT EQUAL LOUDNESS AND AID GENERATED NOISE LEVEL.

FIG. 9.14 AID GENERATED NOISE LEVEL AS A FUNCTION OF AID GAIN.
FIG. 9.15 MEAN SUBJECTIVE GAIN OF EVERY AID TESTED IN THIS STUDY PLOTTED AS A FUNCTION OF PRESSURE INSERTION GAIN. (AVERAGED OVER SUBJECTS.)
FIG. 9.16 PARTIAL MASKING OF A 1 kHz TONE — DATA FROM ZWICKER [9, 1963]

FIG. 9.17 PARTIAL MASKING OF A 1 kHz TONE — DATA FROM HELLMAN AND ZWISLOCKI [10, 1964]
FIG. 9.18
BE11 AID WITH ITS MICROPHONE DRIVEN BY SOUND FROM A RECEIVER VIA A POLYTHENE TUBE

FIG. 9.19
BE11 MODIFIED TO ALLOW EXTERNAL SIGNALS TO BE FED DIRECTLY TO THE RECEIVER
FIG. 9.20  a  MODIFICATION TO THE BE11 AID ENABLING EXTERNALLY GENERATED SIGNALS TO BE FED DIRECT TO THE RECEIVER.

b  THE CIRCUIT USED TO SUPPLY SUCH SIGNALS.

FIG. 9.21  EQUIPMENT USED FOR THE EXPERIMENT DESCRIBED IN SECTION 9.3
CHAPTER 10

GENERAL DISCUSSION

10.1 INTRODUCTION

Individual experiments have been discussed in detail in previous chapters. This chapter summarises and extends the previous discussion and describes some implications of the experimental findings.

10.2 COMPARISON OF HEARING AID RESPONSES MEASURED BY DIFFERENT METHODS

The methods used to measure the response or gain of hearing aids in this study were:

1. Comparison of sound pressures in aided and unaided real ears.
2. Loudness balance between aided and unaided real ears.
3. Acoustic reflex threshold shift between aided and unaided real ears.
4. Comparison of sound pressures in a manikin Kemar's aided and unaided ear.
5. Measurements on Zwislocki and IRPI ear simulators and on a 2 cc coupler of the output and input of the aid.

The first four methods listed measure the insertion gain of an aid whilst the fifth gives the transmission gain into various loads.

10.2.1 Insertion gain methods

Measurements of the pressure insertion gain of an aid gave very similar results to acoustic reflex threshold shift measurements both for individuals and on average (8.6 for exact values). Gains measured on Kemar were very similar to those given by these two real ear methods when averaged over subjects (8.6). Kemar is compared more fully with real subjects below (10.2.4).

The acoustic reflex threshold was at the same sound pressure level in an ear canal whether the canal was open and unaided or sealed and aided. This demonstrates experimentally that there is no physical-acoustic basis for any "missing 6 dB" or "closed ear effect" - the eardrum responds to the sound pressure however that pressure is generated.
Gain measurements by loudness balance differed consistently from the measurements by probe microphone or by acoustic reflex. The loudness balance always gave lower gain measurements at frequencies below 3 kHz, and gains about the same or slightly higher at 4 kHz. At 1 kHz the gain of an aid measured by loudness balance was usually about 60% of the pressure insertion gain expressed in decibels.

The reason for this difference was that a higher sound pressure level was needed in an aided ear than in the same ear unaided for any given loudness. Since the acoustic reflex measurements showed the ear to be responding to sound pressure level whether the ear was aided or unaided some difference between aided and unaided listening was influencing loudness judgements at a level in the brain more central than the acoustic reflex arc.

This difference was shown to be the noise levels occurring in the ear as a background to the loudness balance test tones. The test tones were being partially masked by noise generated within the hearing aid electronics even though signal-to-noise ratios were 20–30 dB or more. The more noise the aid generated the higher was the sound level of the test tone needed for a given loudness: the greater the gain of an aid the more noise it generated.

The loudness balance method therefore was giving an indirect measure of the noise level produced by an aid rather than a direct measure of its gain.

The aids which were tested are typical of aids in current production and use and have typical noise levels. No available aid will have a noise level sufficiently low (i.e., below threshold) for the loudness balance method to be used with normally hearing subjects. Measuring aid gain by loudness balance on subjects with impaired hearing should be possible but might present some difficulties with subjects having vastly different losses in each ear. Measurements of aid gain by auditory threshold shift methods also would suffer from inaccuracies due to aid generated noise if the subjects' thresholds were not sufficiently above the circuit noise level in the ear.

Estimates from Appendix D suggest that if a hearing aid insertion gain is artificially kept below 20 dB at 1 kHz then aided pure-tone
thresholds will be reliable only if participating subjects have hearing levels of at least 40 dB. If subjects are allowed to choose their own gain settings to approximate normal realistic use-gains then their aided thresholds will be free of masking only if their hearing levels exceed about 50 dB at 1 kHz.

In cases where aided thresholds are not affected by noise then loudness of supra-threshold tones will also be unaffected. Thus a loudness balance technique can also be used to measure aid gains in those cases where a threshold shift technique is possible.

Pascoe [1; 1974] who used a threshold shift method with hearing impaired subjects to determine the response of a master hearing aid claims that the noise levels generated within his aid did not affect his measurements. This is consistent with the estimates of Appendix D for the fairly low gain settings he used and for the hearing levels of his subjects.

10.2.2 Transmission gain methods

Transmission gains were measured using the standard 2 cc coupler and the Zwislocki and IRPI XD-1053 ear simulators as loads. The tubing lengths and diameters used with these were the same as those used with real ears and earmoulds and with Kemar (3.5, 3.6.2). Direct comparisons between the couplers, ear simulators and real ears are therefore valid since each was driven in exactly the same manner.

10.2.2.1 Zwislocki and IRPI ear simulators compared

The response of an aid measured on the Zwislocki ear simulator was in all cases similar to the response on the IRPI ear simulator with the odd decibel difference at peaks and dips (Figures 8.9, 9.6, 9.8). This bears out Zuercher and Burkhard's claim [2; 1976] that a carefully designed 2-branch simulator such as the IRPI XD1053 can closely duplicate the performance of the full four branch Zwislocki ear simulator even though its construction is simpler.
10.2.2.2 Zwislocki ear simulator and 2 cc coupler compared

Figure 10.1 shows how the frequency responses of four aids differ between measurements on the Zwislocki ear simulator and the 2 cc coupler. The four curves were obtained by subtracting carefully aligned level recorder charts of the responses measured on each coupler within minutes of each other in the hearing aid test box.

Below 1 kHz there is a fairly constant difference in the measured gain of each aid between the ear simulator and coupler which mainly reflects the difference in their cavity volumes. Above 1 kHz the two couplers give divergent responses where the different cavity shapes of the two devices and the side branches of the Zwislocki make the acoustics more complex. At 4 kHz the difference between measurements on the two couplers is 10-14 dB.

These differences between the 2 cc coupler and the Zwislocki ear simulator are very similar to those found by Sachs and Burkhard [3; 1972] with insert receivers. Their results are shown in Figure 10.2. They found a roughly constant 3½ dB difference below 800 Hz followed by a linear increase to about 15½ dB at 7½ kHz, which is similar to the curve in Figure 10.1 for the BE11.

10.2.2.3 The Zwislocki ear simulator and real ears compared

A detailed comparison of sound levels at points within real earcanals occluded with earmoulds and within the Zwislocki ear simulator can be found in section 7.5.2.

To summarise, Sachs and Burkhard [3, 1972] found the agreement in absolute sound levels between a Zwislocki ear simulator and the median of 12 real ears to be ± 2 dB up to 7 kHz. The present study has shown the Zwislocki ear simulator to give sound levels between 0 dB and 4 dB higher than the mean of four real ears up to a frequency of 4 kHz (Figure 7.8). But with this small number of subjects the sample mean is not necessarily a good estimate of a population mean and a later experiment (9.3) with measurements at 1 kHz only showed sound levels to be lower in the Zwislocki ear simulator (Kemar's ear) than in the mean of six real ears, by 1.8 dB and 2.6 dB (Table 9.5 conditions (iii) and (iv)) as compared to the -2.5 dB of the four ears above.
The relative distribution in sound levels along the occluded Zwislocki ear simulator was very similar to the mean variation along a typical real ear canal (Figure 7.7), and reasonably similar to the mean of four real ear canals (Figure 7.7).

Sound levels can differ by more than 10 dB between the 2 cc coupler and the mean of real ears as shown in Figure 10.3, which is derived from Figure 7.8 and Figure 10.1, and which also shows Sachs and Burkhard's curve.

All data presented in this section regarding the Zwislocki ear simulator apply when the simulator is occluded. Further validating measurements comparing the Zwislocki simulator with real ears were the measurements on Kemar—a Zwislocki ear simulator with attached head and torso (10.2.4), in which the Zwislocki ear simulator is unoccluded.

10.2.3 Transmission gain and insertion gain compared

A transmission gain does not take account of diffraction of sound around an aid wearer's head or the transmission characteristics of the ear canal and the way these change between aided and unaided listening.

Figures 10.4-10.7 illustrate differences between insertion and transmission gains for four hearing aids. The continuous line for each aid shows the difference between its insertion gain on Kemar facing directly towards the sound source and its transmission gain on the Zwislocki ear simulator.

The insertion gains of the Widex 691, the Alto and the BEll were measured by replaying to an aided Kemar a tape recorded frequency sweep synchronised to a level recorder. This sweep was compressed when recorded using Wansdronk's method [4; 1959] to produce a flat response when replayed to Kemar's unaided ear. Thus the aid response was plotted directly with only slight correction needed for a roll off in the tape recorder's frequency response. Transmission gains of these three aids were measured immediately afterwards in the hearing aid test box. The curves for transmission and insertion gain were aligned and subtracted.

The Widex Baritone's insertion gain was measured some time ago (1975) by the cruder method of subtracting curves of sound levels in Kemar's aided and unaided ear. The transmission and insertion gain were measured on
separate occasions. The Baritone aid was no longer available for repeat measurements when measurements on the other aids were made and its curve is therefore less reliable than the curves for each of the other three aids.

Since Kemar's ear is a Zwislocki ear simulator the difference between insertion and transmission gains in these figures is determined entirely by the acoustic performance of Kemar's head, body and ear influencing the input to the aid; the output of the aid being loaded in exactly the same way for both transmission and insertion measurements.

The dotted line in each figure is a general curve for an aid with either a forward or downward facing microphone as appropriate. These general curves were measured by Burkhard [5; 1976/1978]. The squares are values predicted from real head diffraction data of Olsen and Carhart [6, 1975] and ear canals pressure transfer ratios of Wiener and Ross [7; 1946] as in Chapter 4. Again values for a typical forward facing or a typical downward facing microphone have been entered on each figure as appropriate.

As can be seen the agreement between the curves measured in the present study and Burkhard's curves is excellent for the Widex 691 and BE11 aids and very good for the Alto. The Alto curve from the present study is almost exactly the same shape as Burkhard's but is displaced by about 2 dB across the entire frequency range. With the Baritone aid agreement is excellent up to 2 kHz but only fair above. But this is to be expected since the curve for the Baritone was obtained by a method less accurate than the method used with the other aids.

Although the agreement between each curve measured here and Burkhard's curves is very good, there is nevertheless some difference between the Alto and the Widex 691 curves and between the Baritone and BE11 curves though microphone positions are similar within each pair of aids. With the downward facing microphone position this is to be expected since as Burnett and Kuhn [8; 1976; 9; 1977] point out, hearing aid microphones behind the pinna are exposed to severe sound level differences from position-to-position and from frequency to frequency.

The difference between the Widex 691 and Alto aids implies some
variation in level from position to position above the pinna also. This
variation, typically 3 or 4 dB but approaching 8 dB at some frequencies
is a little larger than would be expected from the work of Burnett and
Kuhn [8, 9] and Madaffari [10; 1974].

The curves differing most from Burkhard's, those of the Baritone and
the Alto aids, tend to be similar to the points predicted from Olsen and
Carhart's real ear data. It is possible that the positions chosen by
Olsen and Carhart to represent typical microphone positions of hearing
aids were similar to those of the Baritone and Alto, whilst Burkhard's
chosen positions were similar to those of the Widex 691 and BE11. But
whether or not this was so, some important conclusions may be drawn from
Figures 10.4-10.7.

Firstly a transmission gain measured on an ear simulator may be up
to 20 dB greater than the insertion gain of the same aid on a manikin or
presumably on a person. Secondly, the relationship between insertion
and transmission gain will vary significantly with the microphone position
of an aid. Such variation will not be confined to broad differences
between forward and downward facing positions; there will be differences
between forward facing positions and between downward facing positions.

The differences between insertion gain on Kemar and the transmission
gain on the 2 cc coupler are shown in Figure 10.8. A comparison with the
previous four figures reveals an interesting paradox. The transmission
gain measured on the 2 cc coupler is closer than the transmission gain on
the more exact ear simulator to the insertion gain. For the BE11 in
particular, the transmission gain on the 2 cc coupler is within ± 2½ dB
below 3 kHz and within ± 5 dB up to 4 kHz of the insertion gain.

The error inherent in the response of the 2 cc coupler roughly cancels
with the difference between insertion and transmission gain. This is
purely fortuitous for the 2 cc coupler, but may explain how the 2 cc
coupler has survived the introduction of head worn aids and remained
virtually unchanged for nearly 40 years from Romanow's original design
[11, 1942].

99.
10.2.4 Kemar and real subjects compared

Kemar's occluded ear is no more nor less than a Zwislocki ear simulator which has already been compared with real ears (10.2.2.3).

A comparison of Kemar with real subjects for unaided listening is given in detail in section 7.5.1. To reiterate the mean pressure transfer ratio from the canal entrance to the deepest probe microphone position of 16 mm of four real ears measured in this study was within 2 dB of the comparable value on Kemar at all test frequencies, the highest of which was 6 kHz. Pressure transfer ratios from canal entrance to points 10 mm and 12 mm into Kemar's ear canal were within 2½ dB of comparable real ear data from Djupesland and Zwislocki [12, 1972] and Wiener and Ross [7, 1946], respectively (Figures 7.9, 7.10). Kemar's unoccluded ear therefore is acoustically very similar to the mean of real ears. But these transfer ratios describe the performance of Kemar's ear simulation alone and are virtually independent of the rest of the head and body and any sound diffraction around them.

The pressure transfer ratio from free field to eardrum does depend somewhat more on the diffraction pattern around Kemar as a whole. This ratio was within 3 dB of the composite curve published by Shaw [13, 1974] for all frequencies measured (up to 7 kHz).

The comparison of an insertion gain of a hearing aid measured on Kemar to that measured on real subjects also indicates the degree to which the acoustic field around Kemar duplicates that around real persons. Such a comparison depends not only on the accuracy of the ear simulator but also on the similarity of the sound levels present at the aid microphone position on Kemar to the sound levels at the corresponding position on the real persons. In other words - does the input to the aid microphone vary between Kemar and the mean of real wearers?

Comparative measurements of insertion gains on Kemar and pressure insertion gains on real people in this study are available at frequencies up to 4 kHz for the BE11 aid and at 1 kHz only for two aids with forward facing microphones.

The gain of the BE11 on Kemar was within 3 dB of the mean pressure insertion gain on four subjects at 500 Hz, 1 kHz and 2 kHz (Figure 8.7).
Its gain at 4 kHz was $14\frac{1}{2}$ dB greater measured on Kemar than measured on the real subjects. Of this $14\frac{1}{2}$ dB, up to 4 dB may have been due to sound levels in the real occluded ear being less than in Kemar's (Figure 7.8), the remaining $10\frac{1}{2}$ dB would be attributable to differences in sound level at the aid microphone position. This discrepancy may be due to the severe sound level changes with position behind the pinna identified by Burnett and Kuhn [8, 9] coupled with the aid adopting a slightly different orientation on Kemar than on real wearers. Nevertheless, at 4 kHz the gain on Kemar is 2.4 standard deviations from the mean of the real ears. The real ear probe microphone measurements at 4 kHz were accurate and certainly not in error since acoustic reflex measurements corroborate them. Again the question is: how representative was the sample of 4 real ears of the population and how representative is Kemar?

In the above experiment the BEll's gain at 1 kHz was 3 dB higher on Kemar than the mean of 4 real ears. In two subsequent experiments (9.2 and 9.3) agreement was closer on four different BEll aids, the gain on Kemar being 0.6 dB, 1.0 dB, 0.6 dB and 2.3 dB more than the mean of six real ears. Again it is difficult with a few subjects to estimate the population mean from a very accurately known sample mean. This will be a problem with most subjective experiments of this type in which a few subjects is the norm.

The gains of the two aids with forward facing microphones, the Widex 691 and the Alto Focus, were greater on Kemar than the mean of 6 real subjects by 5.1 dB and 4.5 dB respectively at 1 kHz. In these two cases Kemar is about 1.3 standard deviations from the real ear mean.

The above measurements are presented to illustrate the differences which may be expected when a typically small sample of aided real people is compared to Kemar. No criticism of Kemar is implied since the data in this study are too few to extend any conclusions from the test sample to a population which Kemar may or may not represent.

However, comparing the aided Kemar with aided real subjects is made more difficult by the lack of any published data. Kemar was specifically designed for measuring hearing aid responses [14; 15; 16] and many papers have been written extolling the advantages of manikins for this purpose [17, Cole, 1975; 18, Knowles & Burkhard, 1975]. Where responses of
hearing aids measured on Kemar have been published these have been purely illustrative [14; 15; 16; 17; 18]. Measurements of sound fields alongside Kemar's head have been made [8; 9; 10] but there are to date no published data relating sound pressures around Kemar's head at positions other than the ear canal entrance to sound pressures around real heads. And there are to date no published data relating responses of aids as measured on Kemar to responses of the same aids on real people. The original validating comparisons of Kemar with real persons [14; 15; 16] were unaided; no mention is made of aided measurements, a peculiar omission since Kemar was designed for hearing aid measurements. Only the occluded Zwislocki ear simulation in isolation has been systematically compared in published papers with real occluded ears.

The agreement between Kemar's ear simulation and real ears is undoubtedly good. The agreement between Kemar's various pressure transfer ratios for unaided conditions with those of real persons are also very good. In those respects this study supports previously published work. But because Kemar's unaided performance is good it cannot necessarily be assumed that his aided performance is equally good. Simply because more variables are involved Kemar's aided performance may be poorer than his unaided performance. Whilst the limited data in this study cannot be used to draw conclusions about Kemar's aided performance, they inevitably lead to the conclusion that more work is necessary.

In short, Kemar has not been validated against real subjects for his intended application in hearing aid measurements. This is a serious omission.

Several studies have been or are likely to be conducted in which measurements on Kemar will be made as substitutes for real ear data. The saving in time, money and effort afforded by a manikin make this almost inevitable. The assumption that Kemar will be very similar to the median of real persons will be implied if not stated. In fact this study is not immune from criticism as the comparisons of transmission gain and insertion gain earlier in this chapter were all made with Kemar rather than real heads (10.2.3). But since manikins will be increasingly used in research and development, Kemar should be entirely validated before, not after, this happens.

102.
A validation of Kemar when aided should ideally be in two parts. Firstly a comparison of sound pressure levels point for point on a grid around the head between Kemar and real ears. Emphasis should be placed on the region around the ears. Measurements of this type have been made by Madaffari but for Kemar only - not for real heads. These comparative measurements should cover as wide a frequency range as possible. Behind the ear aids with responses up to 8 kHz and in the ear aids with responses to 16 kHz are already being designed for laboratory studies [19, Killion, 1978]. And the certain use of Kemar in applications other than hearing aid measurement may also demand a wide bandwidth.

The second part of a validation would be a comparison of the responses of several hearing aids on Kemar with the pressure insertion responses of the same aids on a sample of real persons. This would give a direct test of Kemar in his intended role with all sources of variation associated with the head and the ear for both aided and unaided conditions included at once. The aids should be chosen so that their microphone positions cover as wide a spread as possible, above and behind the pinna and in the ear. Their frequency ranges should be as wide as possible. Fit-refit repeatability measurements should be made on all individuals including Kemar.

Should Kemar be as good a representation of the median real person when aided as he is when unaided he will prove an ideal tool in hearing aid research and development.

10.3 IMPLICATIONS OF THIS STUDY FOR HEARING AID RESPONSE MEASUREMENTS

Having compared hearing aid responses measured by different methods the natural question is: which method should be used and when?

In many cases the requirement is, or should be, an aid insertion response measured on an individual rather than an average response or a typical response. For example, research into aid responses has traditionally been directed toward relating a response to some criterion of performance quality as measured by word discrimination or by a subjective assessment.

The same aid, however, can have vastly different responses on different
people. To ignore this fact and correlate *aids* with word scores or subjective assessments rather than correlate *responses* with word scores or assessments is to ignore a major source of variance which would easily conceal real differences between responses. Such a criticism applied in retrospect to the classic Harvard [20, Davis et al, 1947] and MRC [21, MRC, 1947] studies, though for their time they were very ambitious and pioneering studies. But as Pascoe [1] has demonstrated, the right response set for each individual can give significantly greater benefit.

The clinical parallel of the individual approach in research is the prescription form of dispensing.

Various methods of measuring or setting up an aid response on an individual are available. These are loudness balance, auditory threshold shift, probe microphone measurements and acoustic reflex threshold shift. The first two involve subjective judgements whilst the last two do not. There are limitations to each of these methods.

Because the circuit noise generated within an aid affects loudness judgements the loudness balance technique cannot be recommended for measuring the gain of current hearing aids with normally-hearing subjects or those with mild to moderate losses. Circuit noise will also affect auditory threshold shift measurements with the subjects.

Loudness balance and threshold shift measurements should be possible when subjects have a moderate to substantial hearing loss. In those cases where a threshold shift is not affected by noise, loudness balance will also be unaffected. Rough calculations shown in Appendix D suggest that if a hearing aid's insertion gain is artificially kept below 20 dB at 1 kHz then aided pure tone thresholds will be reliable if participating subjects have hearing levels of at least 40 dB. If subjects are permitted to choose their own gain settings to approximate realistic use-gains then their aided thresholds will only be reliable if their hearing levels exceed about 50 dB at 1 kHz.

In cases where aided thresholds are unaffected by circuit noise then either a loudness balance or a threshold shift method of measuring gain may be used.

Pascoe [1] who used a threshold shift method with hearing impaired subjects to determine the response of a master hearing aid claims that
noise generated by his aid did not affect his measured aided thresholds. This is consistent with the estimates in Appendix D for the fairly low gain settings which he used and for the hearing levels of his subjects. However it is doubtful whether Pascoe's technique could be used to optimise frequency responses of a master aid on a patient with a hearing level of less than 50 dB unless the gain of the aid were held artificially low, below a typical use-gain, and the input to the aid microphone were increased accordingly to compensate. The signal-to-noise ratio after all can be varied only by changing the acoustical signal level to the microphone, and the use of a threshold shift technique would not be possible on current production aids to measure responses at normal use-gains with these subjects.

Speech discrimination unlike loudness judgements or threshold measurements is not likely to be affected by electrical noise provided the speech-to-noise ratio is above a minimum threshold. Speech reception is therefore likely to correlate with objective response measures rather than subjective response measurements if the latter are contaminated by noise. It is therefore recommended that objective measurements such as those of pressure insertion gain or acoustic reflex threshold shift are used to measure an individual's or an average real ear response at typical use-gain, at least until loudness balance and threshold shift have been validated on partially hearing subjects.

Pressure insertion gain will be similar to acoustic reflex threshold shift but will be very much quicker, usable over the entire frequency bandwidth of an aid and will not need high sound levels. An EAR earplug with holes made for the sound tube and probe tube will provide a serviceable earmould substitute (see Figure 6.9); so unless venting is required no special mould need be made.

A probe microphone method may not be possible with narrow or sharply bending earcanals. An acoustic reflex shift may not be possible at low frequencies or when a conductive hearing loss is present.

Sometimes it may be impracticable or undesirable to measure responses of an aid on individuals. Hearing aid manufacturers, for example, cannot test every aid on every possible wearer. Also tests on real ears at every stage of the design and development of an aid would be prohibitive in time and money and then a manikin such as Kemar may be useful. Kemar
will give no idea of inter- and intra-subject variability but will give a response of an individual within the range of human subjects. Differences in insertion gain between individuals may easily exceed 20 dB even below 2 kHz with ear-level aids [22, Dalsgaard and Jensen, 1976] so Kemar measurements must be interpreted with caution. However, the lack of variability of Kemar can be turned to a useful advantage when small differences and modifications to aids are being studied. A major advantage of a manikin over a real person is the repeatability and constancy; whether a manikin is exactly similar to the median person is often of secondary importance so long as the manikin falls within the range encompassed by the majority of real persons.

A manikin is a tool for studying sound fields around heads. It can show what factors or variables are important in their effect on aid response and give an idea of the magnitudes of any effect before recourse to human subjects. A good example here is the effect of the head and body in offsetting or rotating the directivity pattern of a directional microphone.

But design criteria for aid responses are not sufficiently advanced at present, and a disadvantage of a manikin is that it may encourage approaches to aid designing which are not the best. Research into aid responses must always be directed towards the needs of the hearing impaired, research projects should not be governed by what is and what is not possible on a manikin. For example, Pascoe has shown that very real benefits may be obtained in some cases by prescriptive aid fitting for individual wearers, a procedure which has no counterpart on a manikin and which might easily have been overlooked if the manikin had already been established. The manikin must not be assigned too great an importance and its limitations must not be forgotten - this, after all, was what happened with the 2 cc coupler.

Given the above considerations, Kemar should have a substantial role in design and development work by aid manufacturers and researchers.

Where a Kemar is not available a Zwislocki or equivalent ear simulator may sometimes be corrected for typical head baffle and ear canal resonance effects using published curves [5] to give an approximate insertion gain. But in the author's experience results may deviate by up to 5 dB from the actual Kemar response at some frequencies. This method of course cannot
take the directivity of aid microphones into account and errors with
directional microphones may be greater. A 2 cc coupler response may not
easily be modified in this way.

For quality control and production testing and exchange of informa-
tion, where only aids of the same type are compared, where frequency band-
width is limited to below 4 kHz, and where insertion gains are not needed,
the 2 cc coupler is the cheapest method and is perfectly adequate for this
limited role.

10.4 THE EFFECT OF AID CIRCUIT NOISE ON GAIN SETTING CHOSEN BY
AID WEARERS

In the loudness balance experiments partial masking by low levels of
circuit noise generated within a hearing aid caused the test signals to
sound considerably quieter than they would if no noise were present. The
size of the effect was surprising. For example, an 84.6 dB SPL tone
through an aid sounded equal in loudness to the 75 dB SPL reference tone,
even though there was a 41 dB ratio of signal-to-noise measured in the
1 kHz third octave or a 33 dB ratio of signal to A-weighted noise. The
normally hearing subjects consequently chose higher than expected sound
output levels from the test aids for a given loudness.

Hearing aid wearers often express dissatisfaction with the circuit
noise of aids during clinical trials [23, Sung et al, 1976]. Since many
aid wearers can clearly hear the circuit noise the question arises: would
hearing aid wearers also choose a higher than expected output level in
normal use by increasing the gain because speech through the aid would
sound "quiet"?

It is impossible to give an absolutely definite answer to this question;
there is simply not sufficient information. However, normal use differs
from these experiments in several respects.

A hearing impaired person with an aid has no more requirement for a
particular loudness than has a normally hearing person listening to a radio
or record player. A normally-hearing person will adjust a radio to a
comfortable listening level - loud enough to hear without strain whilst
quiet enough not to be "too loud". This setting will also give good speech
reception. But the actual level chosen will vary according to ambient
noise levels or any other distractions and the personal preference and
habits of the listener.

107.
Similarly a partially hearing person who is not harrassed by prying experimenters will want optimum speech discrimination at a comfortable output level from his aid. At least, this appears to be the criterion which aid wearers most frequently use to set gain. For example, Reid et al [24, 1977] have shown that the gain of an aid adjusted by a patient in the clinic for a comfortable output from a "cold-running" speech input of 67 dB SPL (a typical conversational level) is approximately the gain chosen by the wearer in average daily use. Walden et al [25, 1977] have found similar results with a slightly higher input level of 70 dB SPL.

Shapiro [26, 1976] claims that hearing aid prescription on the basis of matching the response to the most comfortable listening levels at various frequencies is very successful and has demonstrated [27, 1978] that there is close agreement between aid output levels chosen by patients with sensori-neural hearing losses and their most comfortable listening levels for narrow bands of noise.

The wearer sets his gain for comfort rather than for best speech discrimination, since in many cases a hearing aid wearer will achieve best discrimination at a level above his chosen comfortable listening level [e.g., 28, Katakura, 1968; 29, Lawton, 1978].

Thus a comfortable sound level in the ear seems to be the most common criterion used by aid wearers to set their gain controls. The original question of whether aid circuit noise affects the perceived loudness of speech is best replaced by the question of whether aid circuit noise will affect most comfortable listening levels.

One of the factors influencing our normally hearing person's choice of how loud to have his radio was the ambient noise level. By turning up the radio he can improve the signal-to-noise ratio. The electrically generated circuit noise of a hearing aid cannot be compared with ambient noise. For a hearing aid wearer in any given acoustic the signal-to-noise ratio depends on the input sound level and does not vary with gain setting. This is because the noise in a hearing aid originates in the aid's microphone itself, [30, S.C. Ewens, 1978]. From this stage onwards both the signal and the noise are amplified to the same degree and the signal to circuit-noise ratio is constant.

Since he cannot improve his signal-to-noise ratio the aid wearer must presumably adjust his gain setting to give the most comfortable level in
the ear of a mixture of speech and noise. Presumably for comfort, though not for speech discrimination, the sound pressure level in the ear is alone important and whether the level is due to speech or to noise is immaterial. In normal face to face conversation for most of the time and for most of the frequency bandwidth speech-to-noise ratios will easily exceed 20 dB. Around 1 kHz the ratio of long-time average speech levels to circuit noise will probably be 30 dB to 40 dB in conversation. With these signal to circuit noise ratios the circuit noise will not contribute to sound pressure levels in the ear. If the aid wearer is setting gains purely for most comfortable output levels in the ear, the circuit noise is unlikely to affect use-gain settings. Furthermore in the unlikely event that noise were to affect use-gain settings, a noisy aid would be more likely to induce wearers to reduce rather than increase gains.

The above remarks suggest that hearing aid wearers set gains to give a comfortable level in the ear. Studies by Millin [31, 1965], Brooks [32, 1973], Martin [33, 1973], Boorsma and Courtoy [34, 1975], Byrne and Fifield [35, 1974] and Martin et al [36, 1976] have each shown that experienced aid wearers particularly those with sensori-neural hearing losses choose a gain setting in daily use fairly close to half their hearing loss in decibels. Gains measured in these studies were transmission gains on a 2 cc coupler but are likely to be roughly similar to corresponding real ear insertion gains for most of the aids used.

The threshold of discomfort for most persons with sensori-neural losses will be similar to that of most normally hearing persons. A gain of roughly half the hearing loss will tend to maintain peak levels in the ear approximately half-way between the threshold of hearing and the threshold of discomfort for an aid input of 60 or 70 dB, a level typical of speech peaks. Such gain values are consistent with the hypothesis that the wearer sets for the most comfortable level in his ear. Unfortunately there are no studies relating use-gain to circuit noise levels of aids.

But although an experienced aid wearer will choose a gain roughly equal to half his hearing loss a novice wearer will often choose to use very low gains, perhaps no more than a few decibels. Either the novice's criterion of comfort or the criterion by which he sets the gain is different from that of an experienced wearer. Probably the novice is unused to listening to an aid and needs time to aclimatise to the higher sound level,
different sound quality, different balance of direct to reverberant sound, etc. Circuit noise accompanying a signal might be just one of many unpleasant aspects that a novice notices on first wearing an aid, and to this extent circuit noise may influence gain settings.

10.4.1 Feasibility and desirability of low noise hearing aids

Most of the circuit noise of a modern headworn aid originates in the FET preamplifier built into the ceramic and electret miniature microphones which are the most frequently used types [30, S.C. Ewens, 1978]. These microphones are already extremely sensitive with excellent signal to noise ratios even if no allowance is made for their size. Typical noise floors of the better commercially available designs are equivalent to sound input levels of about 26-28 dB(A), see, for example, data sheets for the BT and BL series of Knowles Electronics Ltd. This figure is only 13 dB(A) worse than a typical laboratory-grade one-inch condenser microphone and until a year or two ago was considerably better than half-inch microphones. But a microphone noise floor equivalent to just 25 dB(A) sound input can easily become 55-65 dB(A) in the ear when an aid has a transmission gain of 30 or 40 dB.

The cost of making a low noise miniature microphone is higher than that of a normal production type because a lower noise higher quality FET must be used. The cost of the FET increases rapidly and may easily double for just a few decibels improvement in its noise floor [28].

However, low noise miniature microphones do exist already as experimental prototypes. Their noise floors are 6-8 dB lower than normal production units giving an electrical noise level equivalent to an input sound level of 20 dB(A) [37, Killion, 1976]. Incorporating those low noise microphones in hearing aids would reduce their circuit noise levels by about 6-8 dB, other things being equal, and improve signal to circuit noise ratios accordingly.

But using low noise microphones is not the easiest way to reduce circuit noise in an aid. More careful positioning of the microphone in relation to the head and ear in future hearing aid designs must be used to enable the microphone to take the fullest advantage of the natural gain given by head baffle and pinna effects. With a stronger direct signal to
the microphone the aid user can reduce his gain setting and thereby reduce the level of circuit noise in his ear while maintaining the same levels of speech sounds.

Using the microphone position which has the strongest signal from a frontal source will also improve the ratio of direct to indirect sound and give a less reverberant signal more immune from room acoustics and thus improve speech reception.

The optimum microphone location, provided acoustic feedback does not occur from the receiver, is that used for in-the-ear aids, i.e., at the sealed entrance to the ear canal. This position takes full advantage of head baffle and the remaining directionality of the pinna [38, Teranishi and Shaw, 1968; 39, 40, Shaw, 1974, 1975] and in addition to giving a strong reception of a frontal signal with reduced pickup of noise and reverberation from behind may also enable the aid wearer to localise sound sources, especially with a binaural fitting.

The next best location is directly in front of the ear [41, Temby, 1965], then above the ear facing forward. Behind the pinna is amongst the worst possible choices of microphone location in a head-worn aid. Temby has shown that a microphone mounted in or in front of the ear gives consistently higher outputs by about 3-4 dB from 800 Hz to 4 kHz at least than a microphone mounted anywhere behind the ear for sound reaching the wearer within an angle of ± 60° of directly ahead. In addition this position will suppress sounds from behind the wearer by between 1 and 5 dB over the same frequency range. Madaffari's measurements on Kemar show similar results [42, 1974].

Moving an aid microphone to the entrance of the blocked ear canal and using a prototype low noise microphone would together give a 10-12 dB improvement in signal-to-circuit noise ratios over the aided frequency bandwidth, compared to current aids with standard microphones behind or above the ear. Thus 10-12 dB reduction in circuit noise levels in the ear is feasible with current technology. Any further reduction is likely to be unnecessary since ambient acoustical noise would then be the limiting factor in most normal environments. But a reduction in circuit noise of 10 dB would be substantial and noticeable to many aid wearers. In many cases circuit noise would be pushed below threshold or below the ambient acoustic noise level. But what precisely would be the benefits
to aid wearers and would these be worthwhile?

Firstly there is evidence that patients prefer the aids with less circuit noise in a hearing aid evaluation, and that circuit noise does influence their choice of aid. Patients frequently "express dissatisfaction" with the circuit noise of some aids even when the gain control is "adjusted to only the half-on position" (sic) [23]. So some but not all patients are clearly bothered by circuit noise though whether those who are dissatisfied all fall into certain categories of hearing loss and what percentage of the total number of patients is affected is not clear.

Secondly, low noise aids per se would probably give no measurable improvement in speech discrimination [23]. However judicious choice of microphone locations to maximise speech input levels and thereby enable lower gains to be used with a consequent reduction of noise level in the ear would probably result in improved speech discrimination. This improvement would not be due to the reduced noise however, but to the reasons outlined above.

Finally, hearing impaired people often derive great enjoyment from listening to music. Large numbers of aids are usually in evidence at concerts. Would not a low noise aid be more acceptable to these persons?

It is impossible to quantify the benefits of low noise aids without laboratory and field trials. The first stage would be to identify that category of hearing impaired person who is dissatisfied with or can hear circuit noise. Following this a direct laboratory comparison between regular and low noise aids would be needed. These comparisons should not be confined to speech discrimination tests which can often be insensitive. Due consideration should also be given to the preferences and subjective comments of the wearers. Finally, if justified by the laboratory trials, a comparison in the field under normal and typical acoustic conditions and ambient noise levels. It would be important throughout to take advantage both of low noise microphones and of microphone placement for low noise though the study would have to enable their effects to be distinguishable. The other benefits of a carefully sited microphone could well outweigh the benefits due to reduced noise.

Low noise designs of hearing aid is a topic worthy of future research but should not be isolated from microphone placement, which may well be the more important.
10.5 IMPLICATIONS FROM THESE EXPERIMENTS FOR LOUDNESS BALANCE STUDIES

The experiments in this series have highlighted a finding which is by no means new but which was unexpected and the importance of which is not widely recognised or appreciated. This is that wide-band background noise can affect judgements of the loudness of a pure tone by several decibels even if the tone-to-noise ratio is several tens of decibels. This effect, partial masking, and its magnitude have been extensively studied and well documented in the past and Figures 9.16 and 9.17 illustrate the findings of two typical studies [43, Zwicker, 1963; 44, Hellman and Zwislocki, 1974].

Notice that wideband noise causes a greater partial masking of a tone than noise confined to the critical band centred on the tone even though the two noises have the same level within the critical band. Thus noise outside the critical band contributes to partial masking. Notice also just how far above the noise level the partial masking caused by that noise can extend.

The findings of the present experiments are fully consistent with the references cited above. The present experiments further illustrate that even noise levels which are marginally above threshold but are below the A-weighted noise floor of the half inch microphone in an ear simulator may partially mask tones of 75-85 dB SPL at 1 kHz. The extraneous noise in this case originated within the circuitry of the hearing aids and the equipment used to generate and amplify the loudness balance tone stimuli.

Electrical circuit noise will be the most common extraneous noise encountered and unless deliberately filtered is likely to be wideband. Many if not most of the spurious supra-threshold results labelled "closed ear effects" or "missing 6 dB" effects in the past can probably be attributed to circuit noise, especially in earlier studies when equipment was less sophisticated. Certainly whenever circuit noise exists in signals presented for subjective evaluation it cannot be ignored. And the implications of this spread wider than hearing aid research - any experiment involving loudness judgements may be affected. Past uses of loudness balance include calibration transfers between audiometric earphones, subjective comparisons of different transducers, determination of freefield sensitivity levels of headphones to DIN 45 619 [43] calibration of ear impedance probes, etc.
In view of the widespread disbelief that low levels of noise can affect loudness judgements to such a large extent the following comments may be opportune.

The loudness of a tone may be affected by any measurable or audible noise accompanying it, and will certainly be affected by any audible wide-band noise.

The degree to which loudness is affected can be estimated from published data [e.g., 44, 45].

Whenever extraneous noise accompanies the wanted test stimuli loudness judgements of these stimuli will be unreliable.

These observations apply at any audible frequency and are not confined to the low frequencies at which physiological noise may cause similar partial masking.

These observations should be taken into account when loudness judgement experiments are designed, in order to ensure the validity of these judgements.

10.6 SUGGESTIONS FOR FURTHER WORK

The following list contains some suggestions for further work arising directly or indirectly from this study. Many of these suggestions have already been proposed in previous discussion sections.

1. A comparison of aid measurements by loudness balance, threshold shift, acoustic reflex shift and probe microphone on hearing impaired subjects. Do the conclusions of this study hold for hearing impaired persons? To what extent will circuit noise influence threshold or loudness judgements? Are the estimates of Appendix D correct?

2. A comparison of the sound diffraction pattern around Kemar's head and ears with those of real heads (see 10.2.4).

3. Validation of Kemar against real persons for aided listening (see 10.2.4).
4. A survey of aid users to define how many persons and what category of hearing impaired person can hear aid circuit noise, and of these how many find it objectionable, disturbing or annoying and under what circumstances.

5. Should a significant number of persons find aid noise objectionable then a study of the effect of circuit noise on (1) speech discrimination, (2) use-gain settings and (3) user acceptability would be in order. What is the noise floor set by normal acoustic environments? Does acoustic noise mask the circuit noise?

6. Further mapping of sound levels in aided ears on more subjects to augment the limited data of Chapter 7.

7. A study of the sources and degree of variability in earmould impressions and earmoulds (see Appendix B). Can earmoulds be improved? What are the implications for maximum gain settings, comfort and acceptability?

8. Design of an aid for optimum microphone position, optimum directivity, minimum noise and reverberation pick-up, optimum signal to circuit noise ratio. Evaluation of such an aid.

9. The application of the probe microphone and acoustic reflex techniques in research into aid responses for individuals (cf. Pascoe's use of threshold methods [1]) with possible extension into normal clinical practice (cf. Fournier's use of threshold methods [46, 1968]).

115.
FIG. 10.1 COMPARISON BETWEEN THE RESPONSE ON THE ZWISLOCKI EAR SIMULATOR AND THE RESPONSE ON THE 2 cc COUPLER OF FOUR DIFFERENT AIDS, THIS STUDY.

FIG. 10.2 COMPARISON BETWEEN THE RESPONSE ON THE ZWISLOCKI EAR SIMULATOR AND THE RESPONSE ON THE 2 cc COUPLER OF TYPICAL INSERT RECEIVERS, AFTER SACHS & BURKHARD [3].
FIG. 10.3 APPROXIMATE AVERAGE DIFFERENCE BETWEEN SPL IN REAL AIDED EARS AND SPL IN THE 2 cc COUPLER.
FIG. 10.4 COMPARISON OF INSERTION GAIN ON KEMAR WITH TRANSMISSION GAIN ON THE ZWISLOCKI EAR SIMULATOR FOR THE OTICON BE 11 (DOWNWARD FACING MIC.)

FIG. 10.5 COMPARISON OF INSERTION GAIN ON KEMAR WITH TRANSMISSION GAIN ON THE ZWISLOCKI EAR SIMULATOR FOR THE WIDEX BARITONE (DOWNWARD FACING MIC.)
FIG. 10.6 COMPARISON OF INSERTION GAIN ON KEMAR WITH TRANSMISSION GAIN ON THE ZWISLOCKI EAR SIMULATOR FOR THE WIDEX 691. (FORWARD FACING MIC.)

FIG. 10.7 COMPARISON OF INSERTION GAIN ON KEMAR WITH TRANSMISSION GAIN ON THE ZWISLOCKI EAR SIMULATOR FOR THE ALTO FOCUS (FORWARD FACING MIC.)
FIG. 10.8 THE DIFFERENCE BETWEEN INSERTION GAIN (ON KEMAR) AND TRANSMISSION GAIN ON THE 2cc COUPLER.
CHAPTER 11

CONCLUSIONS

This study has been designed to develop physical and psycho-acoustic techniques for measuring real ear insertion responses of hearing aids and to validate such methods for use in hearing aid research.

Two real ear methods of measuring hearing aid response, an acoustic reflex threshold shift and a probe-tube microphone technique, have been developed as practical procedures and their validity established on normally-hearing subjects. The two methods gave very similar and reliable results, the average responses measured by each method being within 3 dB of the other over the frequency range tested, i.e., up to 4 kHz. Similar responses between methods were obtained for each individual subject despite intersubject variations. Both methods are suitable for research, though each has a practical disadvantage which may render it unusable with certain persons. For example, not everyone will have a detectable acoustic reflex, and a microphone probe tube will not always fit small or sharply bending ear canals. Although the two methods were not validated on hearing impaired persons it is nevertheless considered that the above conclusions will still apply.

A third real ear method, a subjective loudness balance, was also developed and evaluated with the intention of measuring the aid response as perceived by the wearer. This method was found to be liable to gross error giving gain values which could be lower than those determined by the two above methods by 17 dB or more. These low gain values were shown to be a necessary consequence of, and entirely attributable to, partial masking of the aided loudness balance stimuli by the low level electrical circuit noise generated within the hearing aid. This finding was totally unexpected and has also proved surprising to the many researchers and colleagues with whom the author has been associated. It is all the more important because of this.

116.
It was further shown that this electrical noise, which can be generated at similar low levels by any modern aid, even if only barely audible to the wearer, will partially mask aided loudness balance stimuli even when the stimuli levels are several tens of decibels above the masking noise. This degree of masking has been reconciled with previously published literature.

To emphasize the degree of masking which can occur, low levels of wideband noise of the order of 10–20 dB(A), can affect loudness judgements of tones at levels higher than 75 dB SPL when presented to normally hearing subjects. A rough rule of thumb which applied over the range of masking noise levels (up to 60 dB(A)) and tonal stimulus levels (55–85 dB SPL) in these experiments was that each 20 dB increase in wideband noise level reduced the loudness of a 1 kHz tone by 6–7 dB. It should also be noted that there was no evidence of any "Missing 6 dB" or "Closed Ear" effect in these experiments other than that attributable to partial masking.

Although it should be possible under some circumstances to measure a gain accurately by loudness balance, this would require restricted gain settings on aids and the use of moderately to severely hearing impaired persons. Furthermore, since there is no way of readily distinguishing an accurate loudness balance result from an inaccurate one, the technique would not be reliable. The loudness balance procedure, despite its being a traditional and respectable psychoacoustic technique and despite similar methods having been used before for aid measurements, was therefore rejected as unreliable and potentially subject to gross error. Since there is little prospect of circuit noise levels being significantly reduced in the future, there is little prospect of revising this conclusion.

Circuit noise from an aid will also influence measurements by any auditory threshold shift technique. Although not attempted here for this very reason, such a technique has been employed by Pascoe to measure the response of a low gain master aid on moderately hearing impaired subjects. With this combination his results appear not to be contaminated by masking. Nevertheless, the comments made above concerning the use and reliability of loudness balance will also apply to threshold shift.

117.
Although levels of electrical noise from a modern aid prevent the reliable routine use of a loudness balance or threshold shift in the laboratory or clinic, these levels are unlikely to present any problems in normal use of the aid. Whilst electrical noise may be audible in particularly quiet environments this would not often occur in normal everyday use. Levels are too low to adversely affect speech discrimination and the reduction of noise generated by aids, though desirable if possible, is not a priority for research effort.

Measurements on ear simulators and couplers showed the Zwislocki ear simulator to be a better representation than the 2 cc coupler of the average real ear. Over the frequency range tested (up to 4 kHz) sound levels generated in the occluded ear simulator were within 4 dB of the mean of real ears, whilst the 2 cc coupler differed by up to 9 dB from the mean real ear values. The Zwislocki ear simulator retains this good performance when unoccluded and a comparison of the manikin Kemar with real ears showed an excellent unaided performance. The free field to eardrum pressure transfer ratio of Kemar was within 3 dB of published real head data over the frequency range tested (up to 6 kHz), whilst Kemar’s pressure transfer ratio from ear canal entrance to a position 16 mm into the canal was within 2 dB of mean values measured over the same frequency range.

Insertion gains of hearing aids measured on Kemar were within 5 dB of the mean real ear insertion gain for all aids tested at 500 Hz, 1 kHz and 2 kHz. However, at 4 kHz the insertion gain of a BE 11 aid on Kemar differed by 14 dB from the mean real ear value. The real ear measurements were considered reliable, similar values being obtained by 2 completely different methods (probe microphone and acoustic reflex). Since there is no published comparison of the aided performance of Kemar with that of real persons, the above data are the only available. In view of this large discrepancy and the lack of any other data, this aspect requires further investigation and clarification. This need is urgent since a manikin such as Kemar is likely to be adopted in the near future as a standardised device for hearing aid measurements. As experience with the 2 cc coupler has shown, once a device has been standardised it is likely to remain in use unmodified for a very long time. In the meantime high frequency aided measurements on Kemar should be treated with caution.

118.
In summary, this study has validated the use of an acoustic reflex threshold shift and a probe microphone technique for measuring hearing aid responses, but has shown that a loudness balance method should not be used since results may be subject to gross error. Thus two methods of measuring an aid’s response are available to hearing aid researchers and may now be used, for the first time with some confidence that the results obtained are valid, whilst a formerly accepted method has been rejected. A discrepancy was noted between the aided performance of a manikin and that of real persons and although data is limited, there is an urgent need to investigate this fully before a manikin is adopted as an industry standard.
APPENDIX A

VALIDATION STUDY OF THE LOUDNESS BALANCE TECHNIQUE

A.1 INTRODUCTION

Before using a loudness balance technique as a measuring tool it is necessary to have an estimate of its inherent accuracy. Consequently this experiment was devised to evaluate the abilities of subjects in binaural loudness balancing, and to prove the equipment.

Particular aims of the experiment were:

(i) to obtain measures of mean balance errors and their deviations;
(ii) to evaluate the effects of frequency and intensity on balance accuracy;
(iii) to test the performance of the equipment;
(iv) to identify any possible sources of error inherent in the technique;
(v) to establish a pool of trained experimental subjects, to take part in subsequent experiments, who would be familiar with the equipment and technique.

A.2 PROCEDURE

A.2.1 Experimental Design

Subjects were required to adjust the loudness of a pulsed pure tone under their direct control and presented to one ear from a hearing aid receiver via an earmould, to match the loudness of a fixed level reference tone similarly presented to their other ear. For each balance the reference and variable stimuli were identical in every respect save intensity.

Loudness matches were made with two levels of reference stimulus, 60 dB and 88 dB as measured in the Zwislocki ear simulator; at six frequencies, 250 Hz, 500 Hz, 1 kHz, 2.5 kHz, 5 kHz and 10 kHz; and with each ear, left and right, alternating as the reference ear. Each subject balanced loudness twice for every possible combination of frequency, reference level and reference ear during the course of four separate sessions, each short enough to maintain the subject's interest and attention. Thus there were twelve separate loudness matches per session, each session constituting
effectively a half- replicate of the experiment. Table 1 shows the combinations of frequency and reference level balanced in each session. The order of these balances within the session was randomised to minimise presentation order effects and to spread learning effects equally between all experimental conditions.

A.2.2 Subjects

Twenty normal subjects as defined in section 3.4 took part. Thirteen were male, seven female.

A.2.3 Facilities and Equipment

The equipment used is shown in the block diagram of Figure A.1; the subject was seated in the anechoic room.

Pure tone signals from the audiometer were presented to the subject from two matched experimental wideband receivers each mounted in a behind-the-ear hearing aid shell. The acoustic signal was transmitted to the subject's ear via an earmould containing 2 mm internal diameter tube of 125 mm length. (This experiment was conducted prior to the decision to standardise on a 40 mm length of tubing.)

Each audiometer channel was calibrated at each frequency by relating the voltage registered on the voltmeter to the sound pressure level produced in the Zwislocki ear simulator, and was found to be linear over the range of the voltmeter.

Comparative frequency responses of the two matched wideband receivers together with their associated earmould tubing are shown in Figure A.2.

A logarithmic-law sliding attenuator operated by the subject gave him control over the level of the test channel. For some tests the subject's manipulation of this control was monitored on a level recorder.

A.2.4 Experimental procedure

The nature of the experiment was explained to each subject at his first session and he was given these written instructions:
INSTRUCTIONS

Your task in this experiment is to adjust the loudness of a sound in one ear to match the fixed loudness of a similar sound in your other ear.

When a yellow light comes on at the top or bottom of the control move the slider as far as it will go towards the light.

When you hear pulses of sound in each ear alternately, balance the sounds in each ear to the same loudness using the slider. A green light will indicate which ear, left or right, is under your control.

Please take as long as you like to achieve as good a match as possible between the sounds; there is no hurry.

When you are satisfied that the sounds in each ear are equally loud, press the "task complete" button and wait. The above sequence will be repeated with different sounds, or with the other ear under your control, or both.

If you have any questions or problems during the experiment please ask over the intercom which will be on continuously.

Have you any questions?

The instructions were repeated at subsequent sessions.

The subject performed twelve separate loudness balances in each session. Audiometer settings were arranged so that the true balance position on the subject's attenuator was never in the same place twice and the subject could not therefore predict the required settings. Signal lights were used to tell the subject to set his attenuator either at its maximum or minimum before each loudness balance. Thus for half the balances the variable stimulus was initially louder than the reference and for the remainder it was initially quieter. This was to remove any bias inherent in the starting position of the subject's attenuator.

When the subject was satisfied that the reference and test stimuli were matched he signalled by push button to the experimenter who noted the voltmeter reading of the test channel output.
The average time taken for each session was between 25 and 30 minutes. The fastest person took 20 minutes and only one person took longer than 40 minutes. Approximately 70 seconds was required on average for a satisfactory loudness balance.

A.3 RESULTS AND ANALYSIS

Each individual loudness balance gave a "balance error", the difference in decibels between the actual voltmeter reading of the adjustable channel output and the voltmeter reading which would have been obtained had the subject balanced perfectly. The balance error was shown in calibration to be equal to the error in decibels of sound pressure level measured in the Zwischenki ear simulator. A positive balance error indicates that the test stimulus was set to a higher sound pressure level than the reference when judged equally loud.

A.3.1 Statistical analysis of early data

A four factor analysis of variance of the balance errors of the first five subjects' first replicate is shown in Table A.2. All two factor interactions involving the "Reference Ear" are significant at the 5% or 1% level.

The "Reference Ear x Subject" interaction indicates that different subjects favour different ears and a subject may consistently set sounds in, say his left ear, to a higher or lower level than sounds in his right whilst another subject would do the reverse. This type of variation is to be expected.

The interaction between "Frequency and Reference Ear" suggests perhaps a difference in sound quality between the two channels differing at different frequencies or that all subjects are suffering from a similar difference in hearing acuity between their two ears at different frequencies. The first possibility is the more likely and many subjects spontaneously commented that the right ear channel sounded slightly "fuzzier", "breathy" or "further away" than the left, an opinion shared by the experimenter.

The "Reference Level" x "Reference Ear" interaction is significant
at the 5% level. This might indicate changes in the relative distortion levels in each channel as the level is changed though such was not measured objectively, total harmonic distortion being approximately 0.5% in each channel.

A.3.2 Analysis and results of complete experiment

At each frequency and reference level during each replication of the experiment two loudness balances were made by each subject, once with the left ear as the reference and once with the right. The errors from these two balances were averaged for each subject before computing standard deviations or an analysis of variance on the complete data. This between ear averaging eliminates differences in acuity between a subject's two ears, any variations between earmoulds, calibration errors, etc., from the variance of the balance error without affecting mean values. This makes the tests of significance of the other variables more sensitive. Typically standard deviations are reduced by a factor of three or four.

The analysis of variance of balance errors from the complete experiment is given in Table A.3.

Figure A.3 shows the mean balance errors and their variation with frequency and reference level. With the exception of the 10 kHz, 60 dB case where the error is less than 2 dB, errors are all less than ± 1½ dB and in some cases considerably less. The difference between reference levels is significant in the analysis of variance.

Figures A.4 and A.5 show the differences between the 2 replications of the experiment. These differences are small and not significant in t-tests.

Figure A.6 shows the mean results for male and female subjects, t-tests show no significant differences.

Monitoring the adjustment of the subject's control attenuator on a chart recorder did not reveal any obvious trends in the approach of each subject to the balancing task.

In the analysis of the full data after the between-ear averaging of balance errors the "Frequency x Reference Level" interaction has reached a level of significance of 1%.

This shows a change in the frequency characteristics of the balance
error between the two intensities. Inspection of Figure A.5 shows this to be most marked at the extreme frequencies.

The increase in the data between the preliminary and the full analysis has confirmed the significance of the "reference level" effect bringing it to 1% significance over the residual and also over its significant interaction. The effect here is obvious in the graphs: balance errors at the low level are positive, and at the high level are negative.

In this analysis of variance on the complete data Frequency and Subject main effects attain a 5% level of significance as might be expected but their effects are slight.

A.3.3 A note on averaging balance errors

In averaging balance errors the arithmetic means of values in decibels were calculated. Stevens [1; 1955] however suggests that for certain loudness balancing calculations, decibels should be converted to sones, averaged, then converted back to decibels.

Averaging sones rather than decibels gives a logarithmic mean rather than an arithmetic mean, and would effectively give more weight to the more positive balance errors.

This technique was rejected since

(i) the sone concept is not beyond question. Whilst "quieter", "louder" or "equally loud" are valid terms when applied to sounds differing only in loudness, the phrases "twice as loud" or "half as loud" are, to the author at least, suspect. Furthermore, although Stevens maintains that a 10 dB increase in level is equivalent to a "doubling in loudness" the variability in his original data for individual subjects is from 2 dB to 22 dB per doubling of loudness. This suggests that if "twice as loud" means anything at all it means different things to different people.

(ii) On a practical level the spread of data is small in the present experiment and sone-averaging would make little difference.

(iii) Any difference would be such that the mean balance error would be increased, i.e., from this experiment it appears that dB averaging gives a more accurate result overall (this does not necessarily hold universally).
(iv) dB-averaging is far simpler and very much quicker.

A.4 DISCUSSION

The mean balance errors in this validation study are less than ±1.5 dB with the exception of the error at 10 kHz/60 dB, which is less than 2 dB. The error and standard deviations at 10 kHz are slightly larger than at other frequencies, suggesting the task may be inherently more difficult at this extreme frequency. The error is in all cases sufficiently small to use the loudness balance technique as a reliable measuring tool.

The factor having most influence on the balance accuracy of the subjects is the level of the reference channel. The balance errors at the higher level are consistently negative, those at the lower level consistently positive. This is consistent with an effect noted by Stevens [2; 1966] and Green [3; 1970] whereby subjects err in their settings towards a comfortable listening level of around 70 dB. Thus a reference level higher than 70 dB will elicit a low setting of the variable channel and vice versa.

A supplementary experiment was carried out to determine whether there was a tendency for subjects to err towards the centre of the range on their control attenuator.

This consisted of two extra loudness balances with the reference channel set to 70 dB at 1 kHz. For one balance the true balance position was near the top of the attenuator range, for the other it was near the bottom. These settings were more extreme than those used in the main experiment. The results which are shown in Table A.4 reveal a slight tendency, not statistically significant with the small number of subjects, for subjects to err in their control setting towards the centre of the range. Thus there is a need to randomise the true balance position on the subjects' control.

Although factors other than frequency and level were shown to be significant, they were responsible for far smaller variations in balance error. The statistical analysis suggests there were slight differences in sound quality between left and right channels but that these were eliminated by the between-ear averaging technique.

The difference between subjects was less significant than the difference between reference levels.
The differences in balance errors between replications were small and not statistically significant. The technique is therefore repeatable, and since no significant differences were found between the abilities of male and female subjects further experiments may use male and female subjects indiscriminately.

The power of this experiment is greater than that expected and in further loudness balances both the number of subjects and the number of replications may be reduced. The results obtained from the first five subjects after completing one replicate are very similar to those obtained overall. In further similar experiments it is considered that the number of subjects may be reduced to 8 or 10 with one replicate per condition and still allow a factor of safety should the experiment prove more difficult and the results more variable. This would cut the time and cost but accuracy would not be unduly degraded. This estimation of the number of subjects required is consistent with recommendations for other similar techniques such as DIN 45619 [4; 1973].

A.5 CONCLUSIONS

The loudness matching ability of normally hearing subjects was such that a mean error of less than ±1.5 dB was obtained over the range 250 Hz to 5 kHz, increasing slightly but still less than ± 2 dB at the extreme frequency of 10 kHz.

The main influence on the balancing accuracy was the sound level of the stimuli though its effect was small and varied to some extent with frequency. Subjects tended to set the comparison stimulus about 0.5 dB below the 88 dB reference stimulus, and about 1 dB higher than the 60 dB reference stimulus. Such a tendency for subjects to err towards a "comfortable listening level" has been noted before in the literature, but these errors are sufficiently small not to affect the validity of using the loudness balance technique for hearing aid evaluation.

Variations between frequencies and between subjects were found but were less significant than the intensity effects. No differences were found between male and female subjects, at least in their abilities to match loudness.

No major sources of error were found though a need to scatter the settings required on the subject's control for a perfect loudness balance.
was identified. This was due to a slight tendency of the subjects to avoid extreme settings and err towards the centre of their control range. The operation of the loudness balance audiometer and other equipment was perfectly satisfactory.

The loudness balance technique is a repeatable and accurate measurement tool under the ideal conditions of this experiment in which each individual comparison was between two stimuli identical in all respects save their initial intensities.
<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>250</th>
<th>500</th>
<th>1k</th>
<th>2.5k</th>
<th>5k</th>
<th>10k</th>
</tr>
</thead>
<tbody>
<tr>
<td>Reference level 88 dB</td>
<td>2</td>
<td>1</td>
<td>2</td>
<td>1</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>60 dB</td>
<td>1</td>
<td>2</td>
<td>1</td>
<td>2</td>
<td>2</td>
<td>1</td>
</tr>
</tbody>
</table>

Second complete replicate (sessions 3 & 4)

<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>250</th>
<th>500</th>
<th>1k</th>
<th>2.5k</th>
<th>5k</th>
<th>10k</th>
</tr>
</thead>
<tbody>
<tr>
<td>Reference level 88 dB</td>
<td>3</td>
<td>4</td>
<td>4</td>
<td>3</td>
<td>3</td>
<td>4</td>
</tr>
<tr>
<td>60 dB</td>
<td>4</td>
<td>3</td>
<td>3</td>
<td>4</td>
<td>4</td>
<td>3</td>
</tr>
</tbody>
</table>

Tabulated numbers indicate in which session the frequency-intensity combinations were presented. Each combination was presented with each ear as the reference. All presentation combinations (frequency-intensity-ear as reference) were completely randomised within sessions for each subject.
### Table A.2

**FOUR FACTOR ANALYSIS OF VARIANCE OF BALANCE ERRORS: LIMITED DATA**  
*(FIRST FIVE SUBJECTS TO FINISH FIRST REPLICATE)*

<table>
<thead>
<tr>
<th></th>
<th>Sum of Squares</th>
<th>d.o.f.</th>
<th>Mean-square</th>
<th>F ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>FREQUENCY (F)</td>
<td>23.7</td>
<td>5</td>
<td>4.69</td>
<td></td>
</tr>
<tr>
<td>REFERENCE EAR (E)</td>
<td>1.57</td>
<td>1</td>
<td>1.57</td>
<td></td>
</tr>
<tr>
<td>*REFERENCE LEVEL (L)</td>
<td>37.76</td>
<td>1</td>
<td>37.76</td>
<td>5.41</td>
</tr>
<tr>
<td>SUBJECTS (S)</td>
<td>27.39</td>
<td>4</td>
<td>6.85</td>
<td></td>
</tr>
<tr>
<td><strong>F x E</strong></td>
<td>345.42</td>
<td>5</td>
<td>69.08</td>
<td>9.43</td>
</tr>
<tr>
<td>F x L</td>
<td>6.99</td>
<td>5</td>
<td>1.40</td>
<td></td>
</tr>
<tr>
<td>F x S</td>
<td>105.12</td>
<td>20</td>
<td>5.26</td>
<td></td>
</tr>
<tr>
<td>*E x L</td>
<td>43.15</td>
<td>1</td>
<td>43.15</td>
<td>5.89</td>
</tr>
<tr>
<td><strong>E x S</strong></td>
<td>205.04</td>
<td>4</td>
<td>51.26</td>
<td>7.00</td>
</tr>
<tr>
<td>L x S</td>
<td>51.08</td>
<td>4</td>
<td>12.77</td>
<td></td>
</tr>
<tr>
<td>F x E x L</td>
<td>53.64</td>
<td>5</td>
<td>10.73</td>
<td></td>
</tr>
<tr>
<td>F x E x S</td>
<td>170.94</td>
<td>20</td>
<td>8.55</td>
<td></td>
</tr>
<tr>
<td>F x L x S</td>
<td>69.94</td>
<td>20</td>
<td>3.50</td>
<td></td>
</tr>
<tr>
<td>E x L x S</td>
<td>71.36</td>
<td>4</td>
<td>17.84</td>
<td></td>
</tr>
<tr>
<td>F x E x L x S</td>
<td>139.52</td>
<td>20</td>
<td>6.98</td>
<td></td>
</tr>
<tr>
<td>3 FACTOR INTERACTIONS</td>
<td>505.40</td>
<td>69</td>
<td>7.324</td>
<td></td>
</tr>
<tr>
<td>4 FACTOR INTERACTIONS</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

* = significant at 5% level  
** = significant at 1% level
### TABLE A.3

THREE FACTOR ANALYSIS OF VARIANCE OF BALANCE ERRORS: COMPLETE EXPERIMENT

<table>
<thead>
<tr>
<th></th>
<th>Sum of squares</th>
<th>d.o.f.</th>
<th>Mean square</th>
<th>'F' ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>*FREQUENCY (F)</td>
<td>26.20</td>
<td>5</td>
<td>5.24</td>
<td>2.47</td>
</tr>
<tr>
<td>**REFERENCE LEVEL (L)</td>
<td>267.61</td>
<td>1</td>
<td>267.61</td>
<td>126.23</td>
</tr>
<tr>
<td>*SUBJECTS (S)</td>
<td>103.33</td>
<td>19</td>
<td>5.44</td>
<td>2.57</td>
</tr>
<tr>
<td>**F x L</td>
<td>38.58</td>
<td>5</td>
<td>7.72</td>
<td>3.64</td>
</tr>
<tr>
<td>F x S</td>
<td>139.30</td>
<td>95</td>
<td>1.47</td>
<td></td>
</tr>
<tr>
<td>L x S</td>
<td>38.59</td>
<td>19</td>
<td>2.03</td>
<td></td>
</tr>
<tr>
<td>F x L x S</td>
<td>247.72</td>
<td>95</td>
<td>2.61</td>
<td></td>
</tr>
<tr>
<td>RESIDUAL</td>
<td>463.15</td>
<td>240</td>
<td>1.93</td>
<td></td>
</tr>
</tbody>
</table>

* = significant at the 5% level  
** = significant at the 1% level

### TABLE A.4

INFLUENCE OF THE BALANCE POSITION ON THE SUBJECT'S ATTENUATOR ON THE BALANCE ERROR

<table>
<thead>
<tr>
<th>Subject</th>
<th>(a) With balance position 20% of range from top of scale</th>
<th>(b) With balance position 20% of range from bottom of scale</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Balance Error in dB</td>
<td>Balance Error in dB</td>
</tr>
<tr>
<td>A</td>
<td>-3.8</td>
<td>+4.5</td>
</tr>
<tr>
<td>B</td>
<td>+1.0</td>
<td>+4.4</td>
</tr>
<tr>
<td>C</td>
<td>-3.8</td>
<td>+0.3</td>
</tr>
<tr>
<td>D</td>
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<td>-2.0</td>
</tr>
<tr>
<td>Mean</td>
<td>-1.9</td>
<td>+1.8</td>
</tr>
</tbody>
</table>
FIG. A.1 EQUIPMENT USED IN LOUDNESS BALANCE VALIDATION STUDY

FIG. A.2 COMPARATIVE FREQUENCY RESPONSES OF THE TWO RECEIVERS MEASURED ON A ZWISLOCKI EAR SIMULATOR.
FIG. A.3  VARIATION IN BALANCE ERROR WITH FREQUENCY AND INTENSITY.

FIG. A.4  MEAN BALANCE ERROR: 88 dB REFERENCE LEVEL.
FIG. A.5 MEAN BALANCE ERRORS: 60 dB REFERENCE LEVEL

FIG. A.6 MALE FEMALE DIFFERENCES AND BALANCE ERRORS
APPENDIX B

SOUND ATTENUATION PROPERTIES OF HEARING AID EARMOLDS

B.1 INTRODUCTION

This appendix describes the measurement of the real ear attenuation of an external sound field by the types of earmould used in the main study.

B.1.1 Aims

To measure the real ear attenuation of normal earmoulds and those containing microphone probe tubes over the range of frequencies used in the main experiments.

To determine the change in measured attenuation when the earmould sealing is improved by a smearing of vaseline.

B.2 THE EARMOLDS

Two types of earmoulds were tested in this experiment, the normal ones (section 3.5) and the "probe-tube" types (section 7.3.2 and Figure 7.3).

One normal and one probe-tube mould were made from the same initial impression for each ear of each subject.

B.3 METHOD

B.3.1 Choice of method

Constraints imposed by electrical and acoustic background noise make it impossible to measure the subjective gain of a hearing aid on normally hearing persons using a threshold shift method. Such a method can be used, however, to measure the attenuation of earmoulds and has advantages over the alternative loudness balance method.

A threshold shift method was chosen for this experiment mainly because it would be at least four times quicker than the equivalent loudness
balance. To obtain the same information on the same number of moulds and conditions using the available equipment would require approximately 16 sessions per subject for a loudness balance against the 4 sessions per subject for the threshold method. Most of the time saving of the threshold method is due to the use of an automatic self-recording audiometer not requiring the experimenter to intervene to set up levels and frequencies.

The relative results given by a loudness balance and by a threshold method are discussed below.

B.3.2 Principle of the threshold shift measurement of attenuation

The method depends on finding the monaural free-field threshold of hearing for a number of subjects when their respective mould is worn, and the corresponding threshold when no mould is worn. The difference in decibels between these two thresholds is taken as the attenuation of the earmould. The non-test ear is occluded to give an attenuation greater than that of the earmould in the test ear. The attenuation values then measured will be those for the ear at which the attenuation is the least [1, Zwislocki, 1957] i.e., the test ear.

B.3.3 Subjects

Five otologically and audiometrically normal subjects as defined in section 3.4 took part, four male and one female. All were well practised in self-recording audiometry with pure tones and third-octave band noise and were regular participants in ISVR's routine and contract measurement of hearing protectors to various standards. Subjects were paid at the standard ISVR rate.

B.3.4 Facilities and equipment

B.3.4.1 Test room

This experiment was carried out in the same anechoic room as the main experiments. The noise floor measured in this room is sufficiently low not to mask unoccluded thresholds, being below the levels specified in BS 5108: 1974 [2], ASA Z24.22: 1957 [3] and ANSI S3.19: 1974 [4].
B.3.4.2 Equipment

The block diagram of Figure B.1 shows the experimental set up. The Bekesy audiometer produced pulses of pure tones at each test frequency for 30 seconds then switched automatically to the next frequency, starting at 250 Hz and ending at 10 kHz. The pulses were 410 ms long with 550 ms between them and had an exponential rise and fall. They were amplified and presented to the subject from a single unit loudspeaker. The intensity of the pulses was controlled by the subject - when the response button was pressed the intensity decreased at 4 dB/s, when the button was released the intensity increased at the same rate.

The subject was instructed to press a button for as long as a tone could be heard and to release it when no tone was heard. By alternately pressing and releasing the button the subject controlled the sound intensity so that it rose and fell above and below his threshold. The audiometer simultaneously plotted out the instantaneous sound level on a standard audiogram chart to give an up and down zig-zag trace as shown in Figure B.2. The centre-line through the up and down excursions was taken as the subject's threshold at each frequency.

The subject sat facing the loudspeaker, 1 m away with the centre of the loudspeaker about 64 mm below earcanal level. An average height was used - the loudspeaker was not raised and lowered for individuals.

The subject's non-test ear was occluded with an earplug and a monaural earmuff. The headband of the muff remained well clear of the subject's test ear, mould and aid. Figure B.3 shows the pure-tone attenuation of this type of earplug and earmuff measured previously at ISVR using a binaural threshold shift to ASA Z24.22-1957. The attenuation at each frequency when both plug and muff are worn will be greater than the attenuation of the better device alone, but by how much is impossible to say due to complex coupling between the devices [1, Zwislocki, 1957].

B.3.5 Calibration

Sound pressure levels generated at the centre-head position in the absence of a subject were calibrated against the audiogram chart markings. Below the noise floor of the measuring equipment an electrical calibration of the audiometer output was used instead. An essentially linear calibration was obtained and errors in individual measurements of attenuation

131.
attributable to the audiometer should not exceed $\pm 1\text{dB}$. Such errors are discounted.

**B.3.6 Experimental procedure**

Each of the 5 subjects took part in 4 sessions on 4 separate occasions within a fortnight. Throughout each visit for each subject one ear was the test ear, as defined by the experimental order of Table B.1. The nature and purpose of the experiment was explained to the subject and his non-test ear was occluded with an EAR earplug and the earmuff was placed over it.

The sequence of 5 conditions for the test ear (unoccluded and 4 conditions with earmoulds) are also given in Table B.1. The subject was fitted for the first condition and taken into the test room. His free-field threshold for this condition was measured using the self-recording equipment described above. The subject then left the room to be fitted for the second condition, returned for his threshold to be remeasured and so on till all five conditions had been tested.

Instructions to the subjects were given orally since all subjects were experienced. The instructions were:

"This is a standard Bekesy audiogram. You will hear a series of pure tone pulses - when you hear the tone press this button, when you cannot hear the tone release the button. Keep the button pressed for as long as you can hear the tone, and no matter how quiet the tone is, press the button if you can hear it. Please look directly ahead at the loudspeaker during the experiment."

A trial period of 20-30s was given at the first frequency of each audiogram to give time for the subject to settle.

Earmoulds were fitted by the experimenter and subjects were asked whether a proper fit had been obtained. The sound input tube through each earmould was connected to a BEll aid worn normally but not operating. To avoid trailing wires and since sound levels were too low for probe microphone measurements, the probe tube was sealed by push fitting the vent of a hearing aid receiver of the same dimensions as the probe microphone into it and allowing the tube and receiver to hang freely from the earmould. In the cases where a non-vaselined mould followed a vaselined mould both the subject's ear and the mould were thoroughly wiped with
paper tissues to remove traces of the vaseline.

The session ended when the subject had completed his five audiograms. Each session lasted about 40-50 minutes. The variation in subjects' thresholds over this period would be expected to be small, less than 2 dB [5, Lower and Martin, 1975] and in any case effectively cancelling between occluded and unoccluded conditions due to the balanced design.

B.4 RESULTS

The centre line through the up and down excursions of each audiogram for each frequency was taken as the threshold level. This is normal practice and correlates well with thresholds found by conventional manual audiometry [see, for example, 6, Burns and Hinchcliffe, 1957; 7, Burns and Robinson, 1970].

The unoccluded threshold levels of the ten ears in this experiment were averaged to give the mean sound pressure level in the absence of a subject at the centre-head position. This is plotted against frequency in Figure B.4 together with the standard deviations.

The mean difference between the thresholds occluded with each type of earmould and the respective unoccluded threshold was taken as the attenuation of that type of mould. Conventionally the attenuation of hearing protectors is plotted with attenuation increasing down the graph by analogy with a hearing loss. This convention is followed in Figure B.5 which shows the mean attenuation at each test frequency of each earmould treatment averaged over all 10 ears of the 5 subjects. Figures B.6 to B.9 show the same data but averaged over left and right ears separately.

The shape of these curves, falling to a maximum attenuation at around 3-4 kHz is typical of earplugs and similar devices and is presumed to be due at least in part to the elimination of the open earcanal resonance when a mould is worn.

It can be seen from these curves that the normal and probe-tube moulds give a very similar mean attenuation. Vaseline improves attenuation by 2-3 dB at all test frequencies. Left earmoulds generally show less measured attenuation by about 4½ dB than right earmoulds but this difference is reduced by the vaseline.
Table B.2 shows the analysis of variance of the individual raw measurements of attenuation. The implications of this analysis are fully discussed in the notes accompanying Tables B.2 and B.3.

B.5 DISCUSSION

B.5.1 Threshold data

The mean monaural threshold measured in this experiment and plotted in Figure B.4 compares well with the Minimum Audible Field (MAF) given in ISO R226 [8, 1961]. Although ISO gives curves for binaural listening an estimate of the monaural MAF can be obtained by adding about 2 to 3 dB to the binaural MAF [9, Licklider, 1957; 10, Davis and Silverman, 1970; 11, Killion, 1978; 12, Stream and Dirks, 1974], assuming that the sensitivities of subjects' left and right ears are reasonably similar.

The measured MAF would be expected to be at slightly higher sound levels than the ISO values since the ISO curve applies to persons aged 18-25 years while the age range of subjects in the present experiment was 24-30 years. This seems to be the case over the middle frequency range. The standard deviations measured are also reasonably similar to those quoted by Robinson and Dadson [13, 1956] upon whose data the ISO standard was largely based.

This closeness of agreement is taken as being a good check of the calibration and as showing that any background noise in the room was sufficiently low not to mask unoccluded thresholds and therefore not to affect attenuation measurements.

B.5.2 Factors affecting earmould attenuation

B.5.2.1 Type of mould - probe tubes

From Figure B.5 it can be seen that the mean attenuations of normal and probe-tube earmoulds are very similar and are similarly affected by the vaseline. The statistical analysis indicates that for some ears the probe-tube mould is slightly better than the normal while for others the reverse is true. But when averaged over several subjects the random variations cancel showing that the presence of the probe tube through the earmould does not degrade the attenuation properties.

134.
It is useful at this point to compare a similar though slightly softer type of acrylic earmould with no tubes through, which is intended for use as a personalised earplug. The attenuation of this is shown in Figure B.10. The acrylic earplugs give a slightly greater attenuation than the hearing aid moulds probably because the earplugs tested were made by an experienced medical technician, but the difference is not great and the standard deviations are similar.

The acrylic earplugs made by the experienced technician probably give a good indication of the maximum attainable attenuation for acrylic plugs and moulds.

B.5.2.2 Left ear/right ear mould differences

A major source of variation in earmould attenuation measured in this experiment is that right earmoulds give a significantly greater attenuation than left earmoulds by 6 dB on average. Also vaseline improves attenuation of left earmoulds more than the attenuation of right earmoulds, suggesting a quality of fit problem - the vaseline can improve the attenuation of an initially poorly fitting mould more than it can improve the attenuation of an already well fitting mould.

This quality variation is considered to be due to the (right handed) experimenter finding right ear impressions easier to make than left impressions. This effect will obviously vary with the technician making the initial impressions.

A similar difference in attenuation by left and right earmoulds was noted by Grover and Martin [14, 1974] using a different measuring technique. They found that left earmoulds gave greater attenuation than right earmoulds - the reverse of this experiment - by about 2 dB. Apparently some technicians will make better left ear impressions and some will make better right ear impressions, and the degree of skill and experience of the person taking the impressions can materially affect the quality of the final earmould. Results here suggest that up to 5 dB or so of attenuation can be lost due to a poor initial impression if this is the cause.
B.5.2.3 Vaseline

It has already been noted that vaseline improves earmould attenuation. The improvement is about 3.7 dB on average for the poorer (left) earmoulds and about 1.0 dB for the better (right) earmoulds. This improvement is not confined to the low frequency band where the effect of leaks around the mould might be expected to be greatest, but applies almost equally at all frequencies. In addition to preventing leakage around the earmould the vaseline may be damping any small vibrations of the mould and hindering retransmission of sound by the mould into the ear canal.

Vaseline is therefore recommended for good measure if the best possible sealing of an earmould to an ear is required for short experimental use. The unqualified use of vaseline on earmoulds by a deaf person is not recommended – firstly because vaseline is no substitute for a well fitting earmould and secondly its constant use might not be beneficial to the general health of the outer ear.

B.5.2.4 Possible variations in earmould manufacture

In addition to the probable systematic difference in quality between left and right ear initial impressions, the manufacturing process by which a finished mould is derived from an initial impression introduces random differences between two earmoulds derived from the same initial impression. In some cases the probe-tube mould was up to 5 dB or so better than the corresponding normal mould from the same impression, in some cases the normal mould was the better.

On average normal and probe-tube moulds had the same attenuation as noted above so differences are random not systematic. In some cases a casual inspection of corresponding normal and probe-tube moulds from the same impression revealed obvious physical differences, particularly in the shaping of the inner-most tip of the earmould.

B.5.3 Threshold shift and loudness balance measurements compared

Will a threshold shift measurement of earmould attenuation give the same results as a loudness balance measurement under otherwise identical conditions? This question must be answered before the sound leakage through or around an earmould as measured in this experiment can be compared with experiments using a loudness balance to measure the sound output from a hearing aid.
Hershkowitz and Levine [15, 1957] cite Shaw and Veneklasen [16, 1945] as having shown threshold shift and loudness balance measurements to be within 3 dB of each other when used to measure the attenuation of V-51R earplug. Unfortunately they do not state whether one method gave consistently higher or lower results or whether differences were random.

Hershkowitz and Levine's own work with earmuffs shows the threshold method to give a greater attenuation than the loudness balance by about 3 dB at 125 Hz and 6 dB at 2 kHz and 4 kHz. (125 Hz was not used as a test frequency in this experiment.) Their figure 7 shows little difference between threshold and loudness balance methods at 250 Hz with the threshold-measured attenuation possibly lower than the loudness balance equivalent at 500 Hz. But differences do occur at 1 kHz and above. The 3 dB greater attenuation shown by the threshold method at 125 Hz is probably due to increased low frequency physiological noise when the muff is worn, masking the test sounds at threshold. This physiological noise, which is due to blood circulating in the ear, heart beat, muscular movement, breathing, etc., would not be expected to cause much masking at 250 Hz and above [17, Anderson and Whittle, 1971].

Weinreb and Touger [18, 1959] also compared loudness balance and threshold shift measurements of earmuff attenuation. Again the threshold shift measurements gave on average a greater measured attenuation than the loudness balance by 3 to 4 dB at 125 Hz, by about 1 to 2 dB at 250 Hz and 500 Hz and by 5 to 7 dB at 1 kHz and higher.

Meij et al [19, 1974] have shown that a loudness balance technique can give a far smaller (15 dB) measured value of attenuation than a threshold shift. However, these data compare a diffuse-field, narrow-band noise loudness balance with a free-field, pure-tone threshold shift. Thus more is being compared than just the loudness balance and threshold shift procedures, and as Hershkowitz and Levine have pointed out, a diffuse sound field gives lower attenuation values than a directional sound field. Meij's comparison is therefore too uncontrolled for consideration for our purposes.

The studies which compare loudness matching and threshold shift methods under otherwise similar conditions, i.e., those cited above with the exception of Meij's, all seem to be in reasonable agreement. They show that under free-field conditions with pure tone stimuli, threshold shift and loudness balance measurements will be in close agreement at
250 Hz and 500 Hz, with the threshold shift method giving greater measured attenuation values by 6 dB or so at 1 kHz and above. The exact relationship between the two methods for any individual type of hearing protector or earmould may vary from this by, say ± 3 dB.

Thus from the results of this experiment and from the papers cited above it is possible to predict roughly the results which would have been obtained had a loudness balance method been used to measure earmould attenuation. Results from this experiment can therefore be compared with the results of experiments in which a loudness balance method is used.

B.6 CONCLUSIONS

The real ear attenuation at threshold of normal and probe-tube earmoulds with a BTE aid has been measured. The mean measured attenuation of each type of mould was at a minimum of 10 dB at the lowest test frequency of 250 Hz and at a maximum of about 32 dB at about 4 kHz. No significant difference in attenuation between normal and probe-tube earmoulds was measured at any frequency, though there were random differences between the two types for individual ears despite their having been made from the same initial impression. The presence of the probe tube does not degrade earmould attenuation.

Smearing earmoulds with vaseline improved their measured attenuation by an average of 2 or 3 dB. This improvement was independent of test frequency.

Vaseline improved attenuation figures for left earmoulds more than it did for right earmoulds. Left earmoulds were found to give significantly poorer attenuation than right earmoulds by an average of 6 dB without vaseline and 3 dB with.
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L = LEFT EAR AS TEST EAR  
R = RIGHT EAR AS TEST EAR  
A = EXPERIMENTAL MOULD WITH VASELINE  
B = NORMAL MOULD WITHOUT VASELINE  
C = EXPERIMENTAL MOULD WITHOUT VASELINE  
D = UNOCCLUDED  
E = NORMAL MOULD WITH VASELINE
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<th>Variance ratio</th>
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<td>47595.58</td>
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<td>4326.87</td>
<td>112.163</td>
<td>*** Δ</td>
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|                  |        |                |                  |              |                |              |
| EAR x MOULDS     | 1      | 207.20         | 0.16             | 207.20       | 5.371          | *            |
| EAR x VASL       | 1      | 437.40         | 0.34             | 437.40       | 11.338         | ***          |
| MOULDS x VASL    | 1      | 121.84         | 0.10             | 121.84       | 3.158          |              |
| EAR x SUBJECT    | 4      | 1155.42        | 0.90             | 288.86       | 7.488          | ***          |
| MOULD x SUBJECT  | 4      | 3127.45        | 2.44             | 781.86       | 20.268         | ***          |
| VASL x SUBJECT   | 4      | 380.35         | 0.30             | 95.09        | 2.465          | *            |
| EAR x FREQ       | 11     | 531.95         | 0.42             | 48.36        | 1.254          |              |
| MOULDS x FREQ    | 11     | 129.55         | 0.10             | 11.78        | 0.305          |              |
| VASL x FREQ      | 11     | 101.75         | 0.08             | 9.25         | 0.240          |              |
| SUBJECT x FREQ   | 44     | 7140.92        | 5.58             | 162.29       | 4.207          | ***          |

|                  |        |                |                  |              |                |              |
| EAR x MOULDS x VASL | 1     | 46.82          | 0.04             | 46.82        | 1.214          |              |
| EAR x MOULDS x SUBJECT | 4    | 2814.57        | 2.20             | 703.64       | 18.240         | ***          |
| EAR x VASL x SUBJECT | 4    | 146.66         | 0.11             | 36.67        | 0.950          |              |
| MOULDS x VASL x SUBJECT | 4   | 487.33         | 0.38             | 121.83       | 3.158          | *            |
| EAR x MOULDS x FREQ | 11   | 507.60         | 0.40             | 46.15        | 1.196          |              |
| EAR x VASL x FREQ  | 11     | 230.65         | 0.18             | 20.97        | 0.544          |              |
| MOULDS x VASL x FREQ | 11  | 92.56          | 0.07             | 8.41         | 0.218          |              |
| EAR x SUBJECT x FREQ | 44  | 3576.18        | 2.79             | 81.28        | 2.107          |              |
| MOULDS x SUBJECT x FREQ | 44  | 1219.35        | 0.95             | 27.71        | 0.718          |              |
| VASL x SUBJECT x FREQ | 44  | 484.90         | 0.38             | 11.02        | 0.286          |              |
| RESIDUAL          | 671    | 25884.87       | 20.21            | 38.58        |                |              |
| TOTAL             | 959    | 128054.93      | 100.00           | 133.53       |                |              |
| GRAND TOTAL       | 959    | 128054.93      | 100.00           | 133.53       |                |              |
**TABLE B.3**

**MAIN SOURCES OF VARIATION AFFECTING MAIN EFFECTS**

(see Notes on Tables B.2 and B.3 below)

(Values are mean attenuations in decibels.)

(a) | EAR | RIGHT | LEFT |
---|------|-------|------|
     | dB   | 25.24 | 20.83 |

(b) | MOULDS | PROBE | NORMAL |
---|--------|-------|--------|
     | dB     | 22.90 | 23.17  |

(c) | VASELINE | WITH | WITHOUT |
---|----------|------|---------|
     | dB       | 24.22 | 21.85  |

(d) | SUBJECT | 1 | 2 | 3 | 4 | 5 |
---|--------|---|---|---|---|---|
     | dB     | 22.09 | 27.86 | 22.70 | 14.09 | 28.53 |

(e) | FREQ. | Hz | 250 | 500 | .750 | 1k | .15k | 2k | 3k | 4k | 5k | 6k | 8k | 10k |
---|-------|----|-----|-----|------|---|-----|----|----|----|----|----|----|----|
     |       |    | 11.23 | 14.09 | 18.34 | 18.31 | 18.54 | 24.76 | 31.41 | 33.26 | 33.30 | 29.60 | 23.50 | 21.06 |

(f) | MOULD | PROBE | NORMAL |
---|-------|-------|--------|
     |        | dB     | 25.56  | 24.91  |
     | EAR    | DB     | 20.93  | 21.43  |

(g) | VASELINE | WITH | WITHOUT |
---|-----------|------|---------|
     | EAR RIGHT | 25.75 | 24.93  |
     | LEFT      | 22.69 | 18.97  |

(h) | SUBJECT | 1 | 2 | 3 | 4 | 5 |
---|--------|---|---|---|---|---|
     | EAR RIGHT | 25.11 | 29.73 | 23.83 | 17.98 | 29.52 |
     | LEFT | 19.06 | 25.99 | 21.57 | 10.20 | 27.33 |

continued/
### (i) SUBJECT

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</table>

### (k) FREQUENCY Hz

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<th>250</th>
<th>500</th>
<th>750</th>
<th>1k</th>
<th>1.5k</th>
<th>2k</th>
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<td>20.75</td>
<td>24.12</td>
<td>24.81</td>
<td>21.75</td>
<td>24.56</td>
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<tr>
<td>3</td>
<td>11.06</td>
<td>11.63</td>
<td>15.63</td>
<td>14.81</td>
<td>12.94</td>
<td>25.19</td>
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<td>5.19</td>
<td>7.00</td>
<td>10.88</td>
<td>10.06</td>
<td>14.69</td>
<td>18.63</td>
</tr>
<tr>
<td>5</td>
<td>17.56</td>
<td>18.50</td>
<td>22.94</td>
<td>23.00</td>
<td>22.56</td>
<td>33.31</td>
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### (k) cont/d.

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<tr>
<th></th>
<th>3k</th>
<th>4k</th>
<th>5k</th>
<th>6k</th>
<th>8k</th>
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<td>35.75</td>
<td>34.63</td>
<td>27.44</td>
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</tr>
<tr>
<td>3</td>
<td>36.00</td>
<td>35.50</td>
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<td>23.25</td>
<td>19.50</td>
<td>20.06</td>
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<td>37.69</td>
<td>32.37</td>
<td>26.31</td>
<td>33.56</td>
</tr>
</tbody>
</table>
NOTES ON THE ANALYSIS OF VARIANCE PRESENTED IN TABLES B.2 AND B.3

Table B.2 shows the analysis of variance of the individual measurements of earmould attenuation in decibels incorporating the repeat measurements.

The three-factor interactions "Ear x Mould x Subject" and "Mould x Subject x Vaseline" are both significant. None of the 2 factor interactions is significant over the three factor interactions, though several are significant over the residual. All the main factors except "mould" are significant over the residual and also over their interactions.

Therefore the factors Ear, Vaseline, Subject and Frequency are the most important variables when measuring earmould attenuation, and the mean variation of attenuation with these variables can be gauged from Table B.3. The interactions are less important second order effects. Though the "mould" factor is not itself significant its interactions are and this is discussed below.

A significant variation between subjects is usual in subjective-acoustical experiments (otherwise measurement could be confined to a single subject). The significant variation in earmould attenuation with Frequency can be seen in Figure B.5. A significant "Ear" effect is shown by Figures B.6 to B.9 to be due to left earmoulds giving generally less attenuation than right earmoulds by an average of 3 dB when vaselined and 6 dB when not as Ear and Vaseline interact.

There is also a significant Vaseline effect which is similar for both the normal and the probe-tube type of mould but which is dependent on which ear, left or right, of a subject is being tested. Vaseline improves left earmould attenuation by an average of 3.7 dB and right earmould attenuation by an average of 1.0 dB. This would be caused not by any differences inherent in left-right as such, but a reflection that left moulds give poorer attenuation than right moulds and that vaseline, in improving sealing of an earmould to an ear, would improve a poor seal more than an already good seal. The "Ear x Vaseline" interaction may therefore be regarded as a "Quality of earmould fit x Vaseline" interaction, combined with the fact that, as tested here, left earmoulds are generally a poorer fit than right earmoulds. The cause of left earmoulds being inferior to right earmoulds is probably due to the (right handed)
experimenter finding it more difficult to make left ear impressions than right ear impressions. If initial ear impressions had been made by someone else the left-right differences might have been reversed or even non-existent. A less likely explanation of the left-right differences might perhaps be conjectured as a greater difficulty in moulding, polishing or drilling of left earmolds at the earmould laboratory.

The improvement in attenuation due to vaseline is roughly the same at all frequencies and not confined to low frequencies as shown by the lack of the "Vaseline x Frequency" interaction.

On average the "Mould" main effect is not significant but its interaction with "Subject" and "Ear" are. There seems to be no systematic difference between the 2 types of mould but the significance of the interactions is probably due to random differences between mould types for the 10 ears tested. These differences between types of mould are up to 5 dB in either direction although both types of mould were made from the same impressions, so there seems to be some variation introduced in the earmould laboratory.

The systematic and random variations introduced at various stages from the ear impression to the finished mould could be a useful topic for a short research programme, though lack of direct relevance and lack of time preclude its inclusion in this particular programme.

Finally the two significant three factor interactions suggest that every individual earmould tested is likely to be different.
FIG. B.1 EQUIPMENT USED TO MEASURE THE SOUND ATTENUATION AFFORDED BY HEARING AID EARMOULDHS.

FIG. B.2 EXAMPLE OF AN AUDIOGRAM SHOWING THRESHOLD WITH AND WITHOUT AN EARMOULD WORN.
The difference between the two thresholds is the earmould’s attenuation. The 0dB-line is arbitrary.
FIG. B.3 PURE TONE ATTENUATION CHARACTERISTICS OF THE EARPLUG AND MUFF USED TO OCCLUDE THE NON-TEST EAR

FIG. B.4 MONOAURAL MINIMUM AUDIBLE FIELD - THIS EXPERIMENT COMPARED WITH ISO R 226.
FIG. B.5 MEASURED ATTENUATION OF THE EARMOULDS averaged over all ears and replications.

FIG. B.6 ATTENUATION OF NORMAL UNVASELINED MOULDS. averaged over subjects and replications.
FIG. B.7 ATTENUATION OF PROBE TUBE MOULDS WITHOUT VASELINE averaged over subjects and replications.

FIG. B.8 ATTENUATION OF NORMAL MOULDS WITH VASELINE averaged over subjects and replications.
FIG. B.9 ATTENUATION OF PROBE TUBE MOULDS WITH VASELINE averaged over subjects and replications.

FIG. B.10 ATTENUATION OF A COMMERCIALY AVAILABLE ACRYLIC EARPLUG MADE FROM AN INDIVIDUAL EAR IMPRESSION
APPENDIX C

PILOT STUDY ON THE USE OF A PROBE-TUBE MICROPHONE FOR REAL EAR MEASUREMENTS

C.1 INTRODUCTION AND METHOD

Measurements of sound pressure level were taken at various frequencies and depths into the unoccluded ear canal using the technique of Villchur and Killion [1, 1975]. Measurements were on the left ear of one male subject only and for sound incident from directly in front, from a loudspeaker 2 metres distant in an anechoic room. Measurements at the different depths were achieved using different length calibrated probe tubes attached to a Knowles XL-9073 microphone. The output of the microphone was filtered (1/3 octave) and measured on a B & K 2112 AF Spectrometer.

Object of measurements: to obtain experience in using the probe microphone; to get some idea of SPL variation in an individual as opposed to an average; to attempt to define a plane at which SPL measurements could be made which would bear some relationship to SPL at the eardrum.

C.2 RESULTS  See Figures C.1 to C.4.

The 14 mm depth was judged on otoscopic examination to be very approximately half-way between ear canal entrance and eardrum for this subject.

C.3 DISCUSSION AND CONCLUSIONS

Measurements (judging by a repeat measurement at 14 mm) seem in general to be repeatable and are consistent with published data [2, Wiener and Ross, 1946; 3, Shaw, 1966; 4, Shaw, 1974]. The odd exceptions at 1.5 kHz and 6 kHz are more likely due to poor positioning of the probe than to changes in sound pressure levels at a particular depth.
FIG. C.1 VARIATION IN SOUND PRESSURE LEVEL ALONG THE UNOCCLUDED EARCANAL OF A SINGLE SUBJECT.
FIG.C.2 PRESSURE LEVELS RELATIVE TO THE EARCANAL ENTRANCE AT DEPTHS INTO THE CANAL OF A SINGLE SUBJECT.
FIG. C.3 PRESSURE TRANSFER RATIO FROM FREE-FIELD TO CANAL ENTRANCE OF A SINGLE SUBJECT.

FIG. C.4 PRESSURE TRANSFER RATIO FROM FREE-FIELD TO CANAL MIDPOINT (14 mm) OF A SINGLE SUBJECT.
# APPENDIX D

**ESTIMATED MAXIMUM GAIN SETTINGS OF HEARING AIDS SUCH THAT CIRCUIT NOISE WOULD NOT MASK AIDED THRESHOLDS**

This table is adapted from Appendix I "Effects of background noise on audiometry" in W. Burns, "Noise and Man" [1, 1973].

<table>
<thead>
<tr>
<th>Hearing level at hearing level given in col. 1</th>
<th>Threshold SPL</th>
<th>Maximum permissible SPL to measure hearing level given in column 1</th>
<th>Insertion gains of typical BTE aids which would give noise level of column 5 in the 1 kHz 1/3 octave band</th>
<th>Estimate of typical use-gain required by patient having hearing level of column 1</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Column 1</td>
<td>Column 2</td>
<td>Columns 3 - 5</td>
<td>Column 6</td>
</tr>
<tr>
<td>db HL</td>
<td>SPL</td>
<td>SPP</td>
<td>LSL</td>
<td>LPL</td>
</tr>
<tr>
<td>-------</td>
<td>-----</td>
<td>-----</td>
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<tr>
<td>61</td>
<td>70</td>
<td>60</td>
<td>66</td>
<td>61</td>
</tr>
</tbody>
</table>

Notes: The frequency of 1 kHz represents the frequency range in which aid generated noise levels are at or near maximum.

Columns 1-4 are derived from Burns' Appendix.

Column 2 shows threshold sound pressure levels corresponding to the hearing levels of Column 1.

Column 3 is derived from Column 2 by subtracting 10 dB since noise in the critical band will not affect the pure-tone threshold if the noise level per critical band is 10 dB or more below the pure-tone threshold (references cited by Burns).

Column 4 is a conversion given by Burns from critical band levels to octave band levels, assuming a reasonably continuous and flat spectrum.

Column 5 is a further conversion to third-octave band levels.

140.
Column 6 shows the insertion gain of typical head worn aids at which they would generate circuit noise at the levels given in Column 5. (NB. The BE 11 is approximately at the centre of the range, the range is based on all the aids tested in this study.)

Column 7 is an estimate of typical use gains set by patients with sensori-neural or mixed losses with hearing levels as given in Column 1. Column 7 is derived from Column 1 by halving the values (see section 10.4 for references and justification).

Interpretation

If a hearing aid has an insertion gain less than that given in Column 6 its gain may be reliably measured on a person having the corresponding hearing level in column 1 using a threshold shift or loudness balance method. If the gain in column 6 is exceeded for the hearing level given in column 1 there is a risk of masking or partial masking of audiometric tones by aid generated circuit noise, e.g., for a person with a 31 dB hearing level an aid gain of about 16 dB, or 13 dB to be really certain, should not be exceeded or his aided threshold may be masked.

Column 7 shows typical gains that a person with the hearing level of column 1 would choose if given the chance. If he chooses a higher gain than the corresponding gains in column 6 then his aided threshold will be masked. Persons with about 50 dB hearing level and above will choose gain settings below the maximum permissible value given in column 6. Therefore persons with hearing levels exceeding 50 dB at 1 kHz may be left to choose realistic gain settings during a threshold shift or loudness balance measurement whilst for persons with less than 50 dB hearing loss aid gains would have to be kept artificially low.

In the event of lower-noise aid microphones becoming available the values in column 6 may be raised accordingly. Thus if, say, a 6 dB reduction in the microphone noise were achieved, then 6 dB could be added to all values in column 6.

A note on noise levels in audiometry rooms

The above table shows that a maximum level of 0 dB SPL at the ear is permissible in the 1 kHz third octave if a hearing level of 0 dB HL is to be measured. The maximum level allowed by the DHSS in audiometry rooms
is however 20 dB SPL in the 1 kHz third octave band [2, DHSS, 1974]. This apparent conflict is due to the fact that most audiometry is conducted with earphones which attenuate the external sound field — Burns cites the attenuation of TDH-39 earphones in MX41/AR cushions as 21 dB at 1 kHz — and this attenuation is allowed for in specifying maximum levels in audiometry rooms. The above table and the DHSS recommendations are therefore perfectly consistent.
CHAPTER 1

1. International Electrotechnical Commission, 1959; "Recommended methods for measurement of the electroacoustic characteristics of hearing aids". IEC R118.


33. M.C. Lower, 1975; "The response of a hearing aid measured subjectively, with Zwischoki and 2cc couplers, and on a manikin". CR 75/9, ISVR, University of Southampton.


46. Brüel & Kjaer, 1977; Tentative product data sheet for Artificial Ear Type 4154 (17-149).

47. F. Keller, 1974; "An improved coupler with a conical volume formation for measurements on hearing aids". Audicibel 23, 110-119.


CHAPTER 3


10. International Electrotechnical Commission, 1959; "Recommended methods for measurement of the electroacoustical characteristics of hearing aids". IEC R118.


1. C. Wansdronk, 1959; "On the influence of the diffraction of sound waves around the human head on the characteristics of hearing aids". *Journal of the Acoustical Society of America*, 31, 1609-1612.


3. W. Olsen and R. Carhart, 1975; "Head diffraction effects on ear level hearing aids". *Audiology* 14, 244-258.


8. DIN 45 619 1973; Blatt 1 "Kopfhörer, Bestimmung des Freifeld-übertragungsmasses durch Lautstärkevergleich mit einer fortschreitenden Schallwelle".


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CHAPTER 5


1. E.A.G. Shaw, 1974; "Transformation of sound pressure level from the free field to the eardrum in the horizontal plane". Journal of the Acoustical Society of America 56, 1848-1861.


17. ISVR Safety and Ethics Committee, 1976; Personal communication.
CHAPTER 7

1. M.C. Lower, 1976; Personal communication.

2. E.A.G. Shaw, 1974; "Transformation of sound pressure from the free-field to the eardrum in the horizontal plane". Journal of the Acoustical Society of America, 56, 1848-1861.


CHAPTER 8


CHAPTER 9


3. S.C. Ewens (Knowles Electronics Ltd) 1978; Personal communication.

4. R.J. Wilton (Knowles Electronics Ltd) 1978; Personal communication.


157.
CHAPTER 10


6. W. Olsen and R. Carhart, 1975; "Head diffraction effects on ear-level hearing aids". Audiology 14, 244-258.


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13. E.A.G. Shaw, 1974; "Transformation of sound pressure level from the free field to the eardrum in the horizontal plane". Journal of the Acoustical Society of America, 56, 1848-1861.


34. A. Boorsma and M. Courtoy, 1974; "Hearing evaluation with hearing impaired children". British Journal of Audiology, 8, 44-46.

35. D.J. Byrne and D. Fifield, 1974; "Evaluation of hearing aid fittings for infants". British Journal of Audiology, 8, 47-54.


42. P.L. Madaffari, 1974; "Pressure response about the ear". Presented at the 88th Meeting of the Acoustical Society of America, November 5, 1974, St. Louis, Missouri. Text supplied by Knowles Electronics Ltd.

44. E. Zwicker, 1963; "Über die Lautheit von ungedrosselten und gedrosselten Schallen". Acustica 13, 194-211.


APPENDIX A


APPENDIX B


12. R.W. Stream and D.D. Dirks, 1974; "Effect of loudspeaker position on difference between earphone and free-field thresholds (MAP and MAF)". Journal of Speech and Hearing Research 17, 549-568.


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APPENDIX C


4. E.A.G. Shaw, 1974; "Transformation of sound pressure from the free field to the eardrum in the horizontal plane". Journal of the Acoustical Society of America, 56, 1848-1861.
APPENDIX D
