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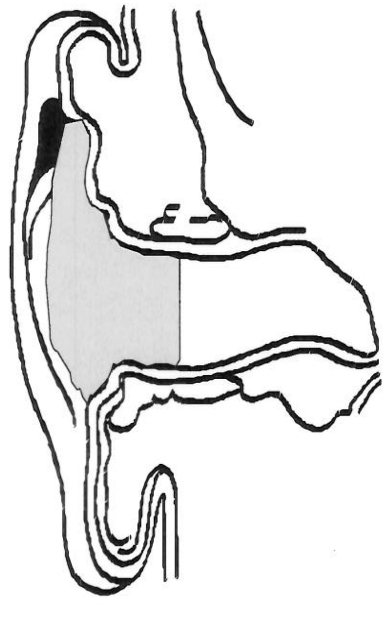
OCCLUSION EFFECTS

Part II

A STUDY OF THE OCCLUSION EFFECT MECHANISM AND THE INFLUENCE OF THE EARMOULD PROPERTIES



Ph.D.-thesis
by Mie Østergaard Hansen (EF 559)

The project was made in corporation with:

Danish Academy of Technical Sciences

Oticon A/S
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Department of Acoustic Technology
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Report No 73, 1998

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PREFACE

This report is submitted in the fulfillment of the requirements for the Danish Ph.D.-degree. The complete thesis consist of 'Occlusion effects. Part I' [Hansen, 1997] and the present report 'Occlusion effects. Part II'. The work has been carried as an industrial Ph.D.-project with Oticon AS as the industrial partner and at the Department of Acoustic Technology, Technical University of Denmark (DTU).

I would like to thank my supervisors senior engineer Peter Lundh (Oticon) and senior lecturer Torben Poulsen (DTU) for good guidance and valuable ideas as well as Niels-Ole Dalsgaard (Oticon) for good support with planning of the project. The National Research Council, Ottawa, Canada kindly hosted me to work there for 6 months, and I would like to thank Michael Stinson and Edgar Shaw for valuable discussions during my stay.

I would also like to thank everybody at the Department of Acoustic Technology , Oticon AS and the National Research Council for being helpful and providing a pleasant atmosphere.

For proofreading thanks are due to Morten Skaarup Jensen and David Caron. Finally, I would like to thank Christian Christensen, for his support.

Lyngby and Hellerup, April 1998

Mie Østergaard Hansen

SUMMARY

The sound from one's own voice is well-known to most people, but if the ear is occluded with an ear plug, the perception of own voice changes, because the sound pressure in the ear canal from own voice is greater at 1-2 kHz when the ear is occluded than in the open ear. This difference in sound pressure is called 'occlusion effect'.

The present report studies the mechanism of the occlusion effect by means of literature studies, experiments and model estimates. A mathematical model of the occlusion effect was developed. The model includes the mechanical properties of the earmould and both the airborne sound and body conducted sound from own voice. These aspects are new in the sense that previous models in the literature disregard the earmould mechanics and includes only one sound source placed in the ear canal.

Own voice is transmitted to the ear canal through vibrations in the body and via the air borne signal radiated from the mouth and nose. A literature study and a pilot experiment supported the theory, that the body conducted sound entering the ear canal is dominated by vibrations of the ear canal walls. An experiment was performed with the purpose to determine the ratio of the sound pressure created by body conducted sound and air borne sound. The ratio was found to be approximately -13 dB with a negative sloping phase at all frequencies from 100-2000 Hz.

The influence of the earmould mechanics was studied by means of pilot experiments with 1-4 subjects. Light acrylic moulds (1 g) were compared with more heavy acrylic moulds (5 g) and foam plugs (0.4 g) was compared with standard acrylic moulds (4 g). The experiments showed that the mass and the elasticity of the earmould is significant. A standard acrylic mould can for example cause up to 15 dB higher occlusion effect than a foam plug.

Finally, natural leakage between the earmould and the ear canal wall and the effect of drilling a vent in the earmould was studied.

RESUMÉ (IN DANISH)

Lyden fra egen stemme er velkendt for de fleste, men hvis en øreprop sættes ind i øret, lyder egen stemme forandret. Det skyldes, at under 1-2 kHz er lydtrykket i øregangen kraftigere, når øregangen er lukket end i det åbne øre. Denne forskel kaldes 'okklusionseffekt'.

Den nærværende rapport beskriver et studie af okklusions-mekanismen ved hjælp af et litteraturstudie, målinger og modelbetragtninger ud fra en ny matematisk model. Modellen tager højde for øreproppens mekaniske egenskaber, og den inkluderer både en luftledt lydkilde og en kropsledt lydkilde. Modellen adskiller sig på denne måde fra tidligere modeller, hvor øreproppen er betragtet som en en masseløs stiv væg. Forskellen ligger også i, at de tidligere modeller kun inkluderer en enkelt lydkilde, som er placeret inde i øregangen.

Lyden fra egen stemme bliver transmitteret til øregangen delvist gennem vibrationer i kroppen og delvist gennem luften. Et litteraturstudie og et pilotforsøg understøttede en teori om, at den kropsledte lyd fra egen stemme genererer lyd i øregangen primært ved at øregangsvæggen vibrerer. Der blev foretaget et forsøg med det formål at finde en værdi for forholdet mellem det lydtryk, som bliver skabt i øregangen af den kropsledte lyd, og det lydtryk som skabes med den luftledte lyd. Det viste sig, at den kropsledte lyd giver ca. 13 dB mindre lydtryk end den luftledte lyd i det åbne øre ved alle frekvenser. Fasen er jævnt aftagende.

Øreproppens indflydelse blev undersøgt via pilotforsøg med 1-4 forsøgspersoner. Lette (1 g) akrylpropper blev sammenlignet med tungere (5 g) akrylpropper og skumpropper (0.4 g) blev sammenlignet med standard akrylpropper (4 g). Målingerne viste, at både massen og elasticiteten af propperne har signifikant indflydelse på okklusionseffekten fra egen stemme. En akrylprop kan for eksempel give 15 dB større okklusionseffekt end en skumprop.

Ydermere er inflydelsen af naturlig lækage og vent i øreproppen blevet analyseret.

TABLE OF CONTENTS

THE MOST IMPORTANT

ABREVIATIONS AND SYMBOLS

AC	Air conduction
BC	Bone conduction
BTE	Behind The Ear hearing aid
c	sound velocity (in air: 343 m/s at 20°)
dB HL	hearing threshold level relative to the hearing threshold of the average normal hearing person
f	frequency
f_0	fundamental frequency
HZ	Hertz
IL or il	insertion depth of the earmould
ITE	In The Ear hearing aid
ITEC	In The Ear Canal hearing aid
j	imaginary notation
OE	occlusion effect (in dB)
oe	occlusion effect (dimensionless value)
OEHAoff	OE with the hearing aid turned off
OEHAon	OE with the hearing aid turned on
p	sound pressure
p_a	sound pressure due to air borne sound
p_c	sound pressure due to body conducted sound
q	volume velocity
ρ	density of air (1.204 kg/m^3 at 20°)
SPL	sound pressure level
ω	angular frequency
Z	impedance

VOCABULARY

acoustic / auditory meatus	the ear canal
air borne sound	sound traveling in the air
anterior	Toward the front
bone conduction	sound conducted in the bones in the body
bone conductor	a vibrator that is placed on the skull, commonly on the forehead or the mastoid. A bone conductor is used to test the hearing threshold of bone conducted sounds.
body conduction	sound conducted in soft cartilage, muscles, bones etc. in the body
frontal	A vertical plane cut in the right left direction
glottis	the air space between the vocal folds
mandibular	the lower jaw
mastoid	the skullbone behind the pinna
ossicles	the three bones in the middle ear
posterior	Toward the back
sagittal	A vertical plane cut in the front back direction
superior	Above
tinnitus	ringing or buzzing sound produced in the inner ear
trachea	the vocal tract
tympanic membrane	the eardrum
vertex	the center on the top of the skull

1. INTRODUCTION

The present report is concerned with the negative aspects of occluding the ear canal with an earmould. The first chapter comprises the background and the objectives of the study. Furthermore a definition and a description is given of the concept "occlusion effect". This definition is somewhat different from the usual audiological understanding of "occlusion effect".

1.1 HOW TO READ THIS REPORT

This report addresses mainly readers who have an interest in the associated problems with the use of hearing aids and ear protectors or any other use of earmoulds. It is assumed that the reader has a technical background and basic knowledge about acoustics.

A quick way of getting an idea of the aspects in this report is to read the main summary, the conclusion and/or the short summaries of each main chapter. The report is introduced with a review of the first part of the Ph.D.-thesis: 'Occlusion effects, Part I' [Hansen, 1997] which concerns the hearing aid users experience of occlusion compared to the objectively measured factors.

The report is structured with three layers of chapters: main chapter, sub-chapter and a sub-sub-chapter. This should make it easy to find a particular subject by looking it up in the table of contents.

Figure and table references are written with bold script in the main text. Table, figure and page numbers are written with the notation x.y. The first number refers to the main chapter number. References to the literature are written in square brackets, [..], or the authors name is mentioned followed by the year in [..]. Abbreviations can be found in the list of abbreviations. Special words and the most important Latin words are explained in the vocabulary. At relevant points the main text refers to appendices and other places in the main report.

1.2 BACKGROUND

Occlusion of the ear canal is good for the purpose of hearing protection and for hearing aids, but it has some negative side effects. The most prevalent side effect is that the perception of one's own voice changes and sounds produced by the user chewing on something crunchy are louder, [Hansen, 1997]. It happens because sounds transmitted through the body will be amplified in the occluded ear canal, especially at low frequencies with the consequence that one's own voice sounds more bassy and hollow, [Hansen,1997].

The occlusion effect created by bone conduction and the mechanism of bone conduction is well described, especially Huizing [1960] has written a comprehensive paper. Bone conduction has had a lot of attention because it clinically can be used to detect a dysfunction in the middle ear. By contrast, there is only little information in the literature concerning the occlusion effect mechanism due to one's own voice. Bone conductor data cannot be assumed to create the same amount of occlusion effect as one's own voice. A bone conductor excites the skull directly at one single point, whereas one's own voice excites soft tissue and bones via vibrations in the larynx and oral cavities. Hence, the mechanism of occlusion effect created by one's own voice deserves more attention.

Few literature references describe a model used to simulate the occlusion effect. Schroeter and Poesselt [1986] and Williams and Howell [1990] developed their models to simulate the occlusion effect created with a bone conductor placed on the skull. Their models did not include the occlusion effect of one's own voice. Weinrich [1986] succeeds in estimating the occlusion effect of one's own voice. However, only one sound source is included in the model, hence sound traveling into the ear canal via a leakage is not implemented. Furthermore, in all 3 models the earmould is regarded as a rigid wall in the ear canal and the mechanical properties of the earmould are not included. In conclusion, none of these previous models were not designed especially to calculate the occlusion effect by one's *own voice*.

1.3 PURPOSE

The reported work had two main purposes:

a) - to study the mechanism of occlusion, especially one's own voice. A hypothesis should be evolved from a literature survey and new experiments.

b) - to develop a computer model that can be used to predict the occlusion effect caused by one's own voice. The model serves as a tool to explain some mechanism of the occlusion effect theoretically and to be used in future work of methods to reduce the occlusion effect.

1.4 DEFINITION OF THE 'OCCLUSION EFFECT'

The original audiological understanding of the 'occlusion effect' was original related to bone conduction threshold measurements. Since then, the notation has been used in several ways to describe changes created by earmoulds. There seems to be some confusion about what the notation 'occlusion effect' exactly refers to. In part 1 of this Ph.D.-thesis [Hansen, 1997] the following definitions were used:

Occlusion	- occlusion of the ear canal
Occlusion effect in general	- objectively measurable and subjectively perceived changes between the occluded and the open ear canal situation.
Objective occlusion effect	- difference in ear canal sound pressure between occluded and open ear:. $\mathrm{OE}(f) = 20\log\left(\frac{P_{occluded\,ear}(f)}{P_{open\,ear}(f)}\right); \quad [dB]$ If nothing else is mentioned it is implied that OE stems from one's own voice.
Subjective occlusion effect	- occlusion effect detected by a psycho-acoustic measurement method
Experienced occlusion effect	- individually experienced annoyances caused by acoustical, mechanical or biological changes between occluded and open ear canal conditions.

The present report concentrates on the objective occlusion effect, OE. For the sake of convenience, the 'objective occlusion effect' will just be called the 'occlusion effect'. Also for the sake of convenience the frequency parameter will be omitted.

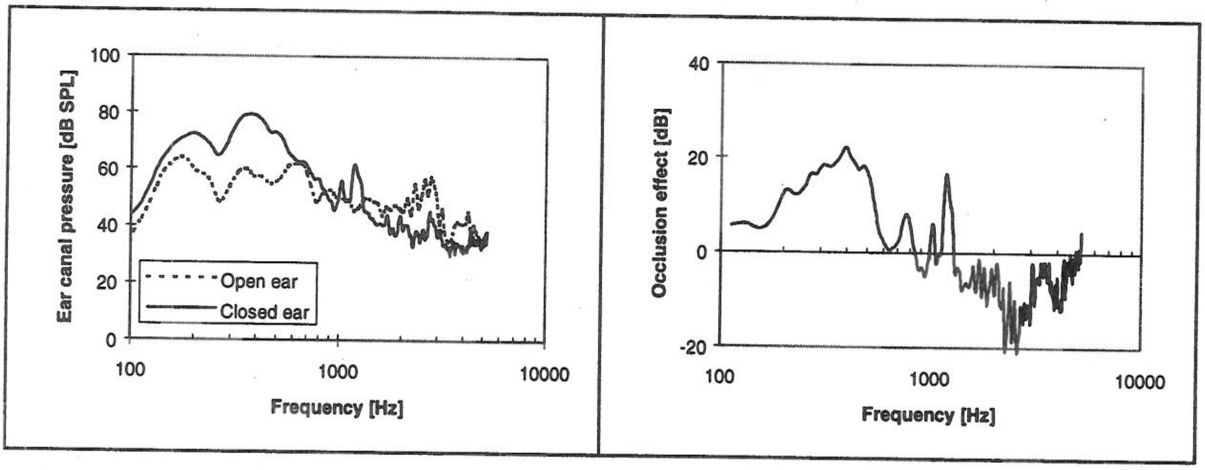


Figure 1.1. Example of the objective occlusion effect. 1 subject, subject's own voice. Note the different axis scaling. [Hansen, 1997 (subject f10)].

The objective occlusion effect varies between subjects even with the same kind of earmould and venting. An example of the occlusion effect is shown in **Figure 1.1**, where the real ear sound pressure is plotted for the open ear and occluded ear. This example is taken from Ph.D.-report Part I.

1.5 ACOUSTIC PRINCIPLE OF THE OCCLUSION EFFECT

1.5.1 One sound path way

When the occlusion effect is measured with a bone conductor placed on the skull, there is only one sound path way, namely, the bone conducted sound. The acoustic effect of occluding the ear is that the sound pressure in the occluded ear canal will be higher than in the open ear canal at frequencies below 1-2 kHz. Imagine that the ear canal is a uniform rigid tube and that the sound source is a piston placed in one end of the tube. The volume in the ear canal changes when the piston moves. If the piston moves fast enough, the volume changes creates an audible sound. The sound pressure depends upon the amplitude of the piston movements. The sound pressure in front of the piston is calculated for the open tube and for the closed tube. The calculated results are shown in **Figure 1.2**. The open ended tube is placed in an infinite baffle to symbolize that the ear canal opening is flanged by the pinna and head. The volume velocity of the piston was arbitrary chosen to $1\text{nm}^3/\text{s}$ and it creates 80 dB SPL at 100 Hz in the occluded ear canal. In conclusion, only a very small volume velocity is needed to create a very high sound pressure in the occluded ear canal.

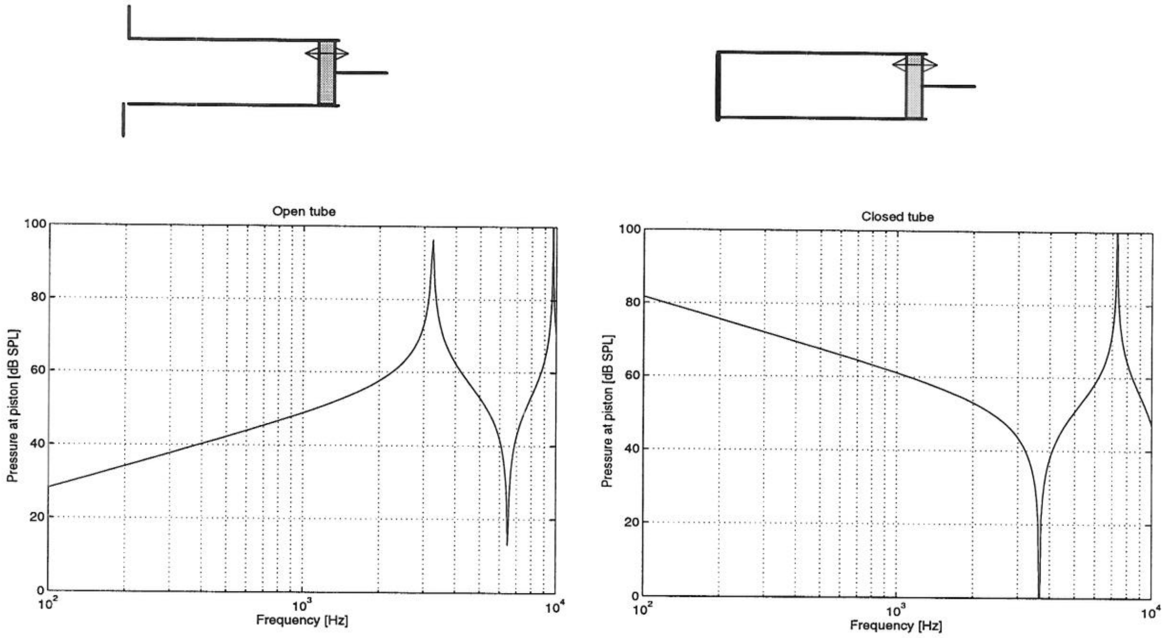


Figure 1.2. Calculated sound pressure at the piston. Left: open flanged tube. Right: rigidly closed tube (length = 23.5 mm, diameter = 7.1mm). Input: 10^{-6} mm3/s.

The occlusion effect is the difference between the closed tube and open tube sound pressure as shown in **Figure 1.3**. The calculated difference is positive below 1500 Hz, then it becomes negative. In an audiological understanding, the term 'occlusion effect' is only the positive difference at low frequencies. But in technical terms, the occlusion effect is the sound pressure difference in the whole frequency range. A negative difference is in a technical understanding also an effect of occluding the ear canal.

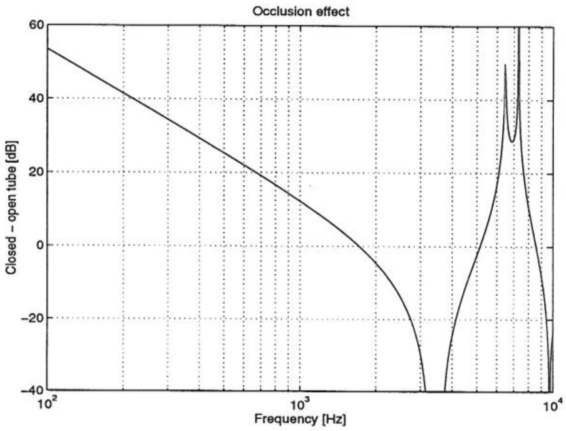


Figure 1.3. Calculated difference in sound pressure between the closed tube and open tube.

The occlusion effect in **Figure 1.3** decreases with -40 dB / decade which corresponds to the ratio of the closed tube and open tube impedance. Looking from the piston towards the other end of the tube, the acoustic impedance is very low in the open tube. In the rigidly closed tube, the acoustic impedance is high. Hence, the occlusion effect can be seen as a change in the acoustic impedance. The acoustic input impedance in an open flanged tube is given by:

$$Z_{open\ tube} \cong j\frac{\rho c}{S}\tan kl_{eff} \qquad (1.1)$$

where;
Z = complex acoustic impedance
ρ = density of air
c = sound velocity in air
k = wave number = $2\pi f/c$ where f = frequency
S = cross-sectional area of the tube
l_{eff} = effective acoustic length of the tube

The input impedance of the closed tube is:

$$Z_{closed\ tube} = \frac{Z_{end}\cos kl + (\rho c/S)\sin kl}{j(S/\rho c)Z_{end}\sin kl + \cos kl} \qquad (1.2)$$

where;
Z_{end} = the acoustic impedance terminating the tube, e.g. an earmould
l = physical length of the tube

If the mould forms an acoustical rigid surface, Z_{end} =1 and the occlusion effect in a dimensionless value, oe, becomes:

$$oe = \frac{Z_{closed\ tube}}{Z_{open\ tube}} = \frac{\cot kl}{\tan kl_{eff}} \qquad (1.3)$$

If $kl \ll 1$ equation (1.3) becomes:

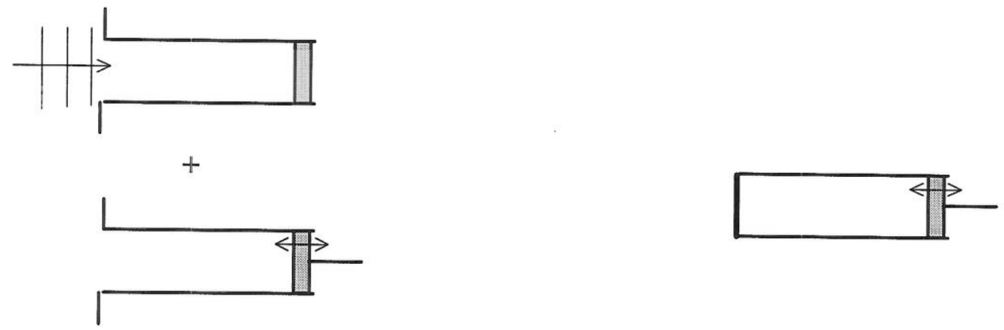
$$oe = \frac{\cot kl}{\tan kl_{eff}} \cong \frac{1/kl}{kl_{eff}} = \frac{c^2}{l \cdot l_{eff}\omega^2} \qquad (1.4)$$

where;
$\omega = 2\pi f$

The occlusion effect decreases with ω^2 which corresponds to -40 dB/decade as illustrated in **Figure 1.3.**

1.5.2 Two sound path ways

one's own voice, the sound reaches the open ear canal by two path ways; through the body and via the air radiated from the mouth and nose around the head into the ear canal. From a simple point of view, this situation corresponds to the system in **Figure 1.2** but now both the piston and an incoming sound wave is present in the open tube, see **Figure 1.4**. The sound pressure is still measured at the piston. The velocity of the piston is no longer frequency independent, but the piston creates a sound pressure in the open tube that is 10 dB lower than the sound pressure created by the incoming sound wave. One problem with a model comprising two sound sources is that the relative magnitude and phase between them has to be known. It will be shown later, why -10 dB (or more accurate -13 dB) is a reasonable estimate. The total sound pressure in the open tube, when both sound sources are present, is the sum of the two complex sound pressures. The closed tube situation is exactly the same as in **Figure 1.2** except that the piston velocity depends on the frequency.



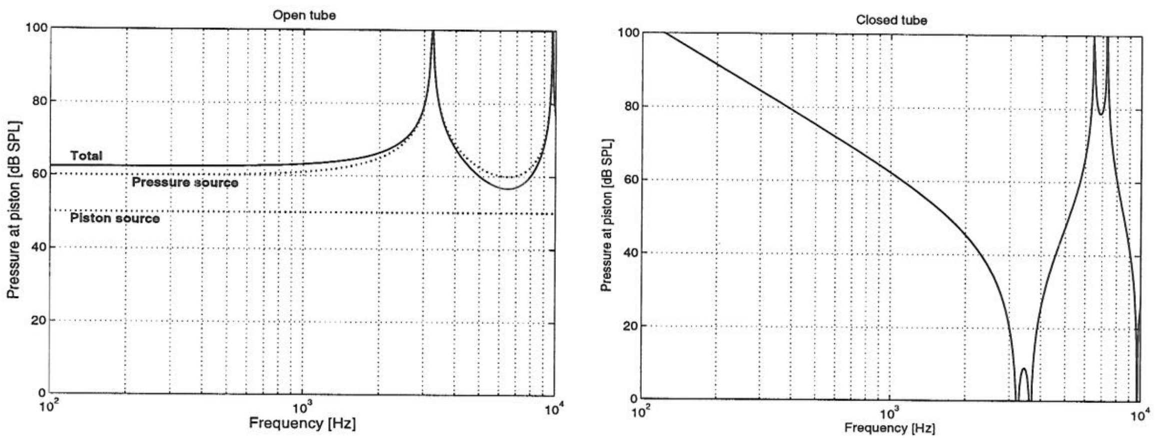


Figure 1.4 Calculated sound pressure at the piston. Rigid tube (length = 23.5 mm, diameter = 7.1mm). Input: pressure source = 60 dB SPL, piston source creates a sound pressure = 50 dB SPL.

The occlusion effect in this situation is shown in **Figure 1.5.** The slope is still -40 dB/decade.

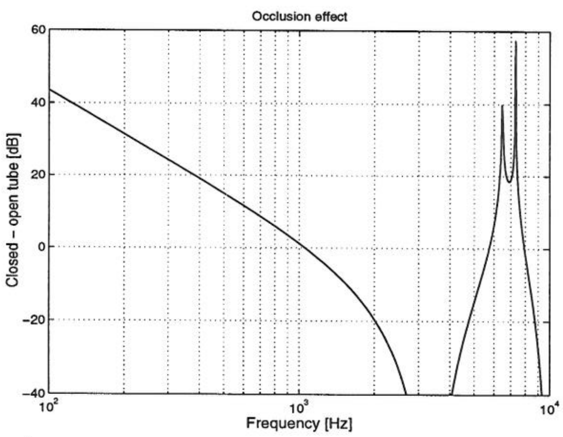


Figure 1.5 Calculated difference in sound pressure between the closed tube and open tube.

1.6 A SIMPLE OCCLUSION EFFECT MODEL

The remaining chapters will make as a first approach make use of the simple illustrated in **Figure 1.6**.

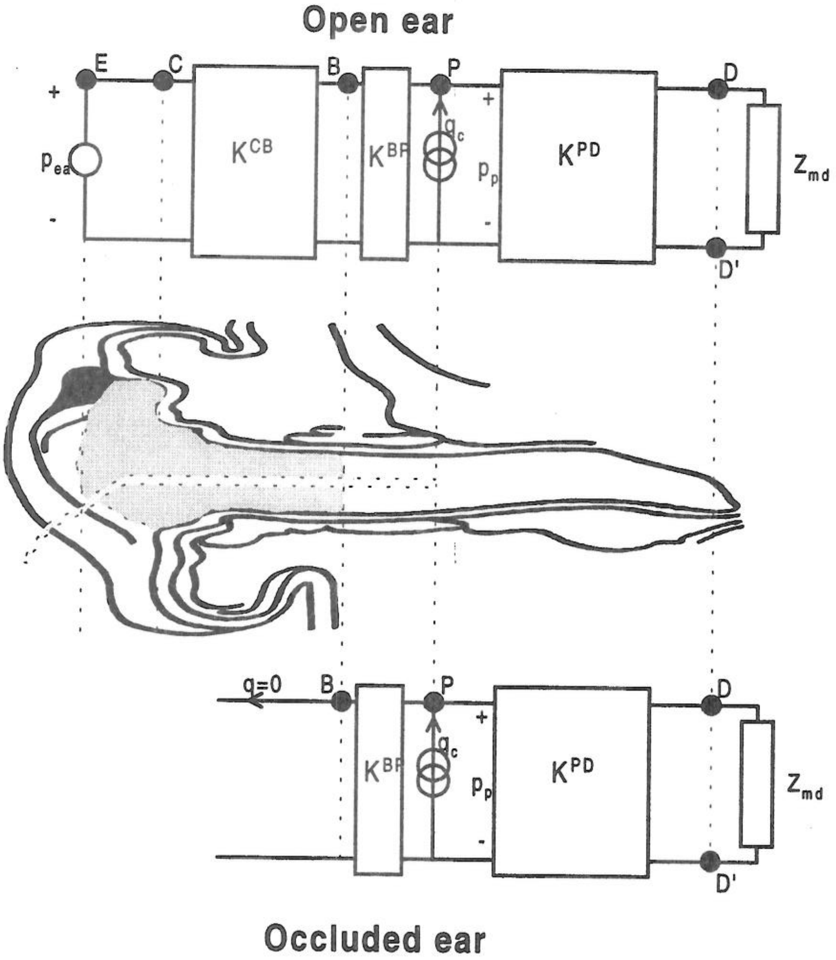


Figure 1.6 Illustration of the model. The airborne sound is P_{ea} and the body conducted sound is generated by q_c placed in point P in the ear canal. The earmould is modeled as an acoustic rigid wall, therefore q = 0. The sound pressure is measured in position P_p with a probe put through the earmould (in the occluded ear) or put into the open ear.

Default settings:

Ear canal: diameter = 7.1 mm, length (C to D) = 23.5 mm
Soft part of ear canal (cartilage) = ½ of ear canal length = 11.8 mm
Earmould : insertion depth (C to B) = 7.8 mm
Probe microphone additional depth = 3 mm (B to P)
Earmould length: BTE = 23.3 mm, ITE = 21.6 mm, ITEC = 20.2 mm
Z_{md} = modified Zwislocki network
$|p_c/p_a|$ = -10 dB
where;
$p_a = p_p$ (open ear) created by p_{ea}
$p_c = p_p$ (open ear) created by q_c

2. REVIEW OF REPORT 1

This is a review of the Ph.D.-report Part I. Detailed information can be found in: 'Occlusion effects, Part I, Hearing aid users experiences of the occlusion effect compared to the real ear level'. Department of Acoustic Technology Report No. 71, Technical University of Denmark, 1997, ISSN 1397-0542.

2.1 PURPOSE

The objective occlusion effect (see chapter 1), has been well proved in the literature for example in: Dirks (1994), Killion et al. (1988) and Wimmer (1986). But there are not many reported data on how the hearing aid users perceive the occlusion effect and other side effects, such as moisture and itch.

Only a few data on how the users perceive the occlusion effect have been determined in laboratory experiments and in most cases with normal hearing persons i.e. persons who are not used to wear earmoulds. These experiments were made with the aim to evaluate the design and vent size of the earmould, for example as in Kuk (1991). Experiments with normal hearing subjects are a simplification of the real situation, mainly because hearing aid users perceive the level and frequency components of a sound differently from persons with no hearing loss. Another factor is that the hearing aid user is exposed to natural sound as well as electronically amplified sound.

At the time where Part I was written, only one survey, made at Bispebjerg Hospital in Denmark, [Biering-Sørensen et al., 1994], reported about hearing aid users experiences with occlusion in their everyday life. The aim of Biering-Sørensen's survey was to compare hearing aid users experience of occlusion with the sound pressure level in the ear canal. In contrary to experiments with normal hearing subjects, the survey could not conclude for sure that there actually is a relationship between the measured and the perceived occlusion effect. This may be due to the fact that the individual hearing losses were not included in the data analyses.

It therefore seemed necessary to investigate the relationship between experienced and objectively measured occlusion effect in more detail. Ph.D.-report Part I had served two purposes:

> a) - to categorize the problems that the hearing aid users experiences in their daily life and to derive the most dominating user problems caused by occlusion of the ear canal.
> b) - to find the most significant relations between the users problems and objectively measurable factors, such as perceptive hearing loss, vent size and real ear measurements.

2.2 METHOD

2.2.1 Sample

The survey was made in corporation with Bispebjerg hospital, who helped to find a proper sample of hearing aid users, counting 48 hearing aid users in total. It was required that the subjects had undamaged eardrum, normal middle ear function, no inflammation in the ear canal or middle ear and were able to read and write without problems. All subjects should have perceptive hearing loss with a difference between right and left ear of maximum 20 dB HL.

The sample consisted of 24 women and 24 men between 24-80 years of age. The hearing loss at 250 Hz ranged from 0-60 dB with a mean of 30 dB sloping to 62 dB at 4000 Hz. All subjects used their own hearing aid which could be an ITEC, ITE or BTE aid. The subject's hearing aid experience ranged from 3 months up to many years.

2.2.2 Measurements

A questionnaire was sent to all subjects and they had to fill out the questionnaire at home. The questionnaire included questions not only about acoustical effects but also about mechanical and biological effects. A few weeks after the subjects had received the questionnaire, they visited Oticon's audiological clinic where the individual objective occlusion effects were measured in an sound box.

The objective occlusion effect was found as the difference in real ear sound pressure level with and without the hearing aid inserted. The real ear sound pressure level was measured with a thin (diameter = 1.1 mm) soft probe tube attached to a microphone. The probe tube was inserted into the open ear canal or placed between the earmould and the ear canal wall reaching at least 3 mm beyond the tip of the earmould[1]. The occlusion effect was measured both with the hearing aid turned off, OEHAoff, and turned on, OEHAon. OEHAon is by some authors called 'ampclusion', because it is an effect of electrical *amp*lification of the airborne sound and oc*clusion* of the ear canal.

The subjects perceive their own voice differently when the ear is occluded and also when the hearing aid is turned on compared to the open ear situation. It may be anticipated that the subjects would not speak with the same level in the three situations: open ear, occluded ear and hearing aid turned on. In order to compensate for this, the speech level was measured with a microphone placed in front of the subjects mouth.

In the pilot study, four stimuli were used: vowels, 60 seconds continuos speech, crunchy chewing and a bone conductor placed on the skull. Continuos speech gave the best retest values and speech was therefore used in the main investigation.

2.3 RESULTS FROM THE QUESTIONNAIRE

The answers from the questionnaire clearly show that occlusion of the ear has some acoustical, mechanical and biological related effects in the every day life for a hearing aid user. Speaking and crunchy chewing were the most prevalent problems.

It was found that 73% of the subjects experienced a change in their own voice when they used their hearing aids and of these, 74% were annoyed by the change. Around 30-45% could also hear louder sounds from heart beats, chewing, drinking and turning the head.

[1] This method is often used in hearing aid clinics to evaluate the gain of the hearing aid inserted in the ear (insertion gain).

About half of the subjects had problems with moist and itchy ear canals and felt that the ventilation of air in the ear canal was insufficient. Eight subjects noticed some of the more extreme biological effects such as headache, eczema and running nose.

The result of the survey does not support the common belief that the hearing aid users will get used to the change in their own voice.

2.4 RESULTS OF THE REAL EAR MEASUREMENTS

The objective occlusion effects are very different from person to person. The peak levels of the occlusion effect varies up to at least 25 dB. In most cases the occlusion effect below 1 kHz is the same with the hearing aid turned off and on, which shows that the sound pressure level produced from one's own voice is greater than the hearing aid amplified sound.

In theory a leakage in the earmould reduces the sound pressure level in the enclosure and the reduction is simply related to the length and diameter of the leakage. In practice there will be a natural leakage between the ear canal wall and the earmould and an intended leakage, a vent, drilled as a cylindrical tube through the mould. Due to the fact, that the occlusion effect varies a lot between subjects, the results from the study could not show a significant relation between the individual occlusion effect and the individual vent size. There is though the tendency, that the occlusion effect variation between subjects decreases when the vent gets larger. The reason is that the individual natural leakage influence the occlusion effect when the vent is small. When the vent is larger, the natural leakage becomes ineffective. This is illustrated later in chapter 9.

Occluding the ear seemed to have another side effect. The recordings from the microphone in front of the speakers mouth indicated that the speech level between 200-2000 Hz is significantly 2-3 dB lower when the subjects had the ears occluded than with open ears. Furthermore, the reduction of speech level was significantly correlated with the annoyance of one's own voice when wearing a hearing aid.

2.4.1 Hearing aid turned on versus hearing aid turned off

For the purpose of the present report it is important to notice that the occlusion effect when the hearing aid turned on (OEHAon) is nearly the same with the hearing aid turned off (OEHAoff). This is not emphasized clearly in Part I of the thesis.

In **Figure 2.1** OEHAon is plotted against OEHAoff. If the measured data for OEHAon is exactly the same as the data for OEHAoff, the data points should lie on the dashed line. The trend line for the measured data has however a smaller slope, which means that OEHAoff is a bit smaller than OEHAon, which is to be expected. If OEHAoff is 25 dB, then according to the trend line, OEHAon will only be 4 dB higher.

If the occlusion effect is small, the difference between OEHAon and OEHAoff will be smaller. The reason is that the earmoulds have such an effective leakage (vent), that the leakage decides the amount of occlusion. When the leakage becomes dominant over the hearing aid amplification, then OEHAon equals OEHAoff.

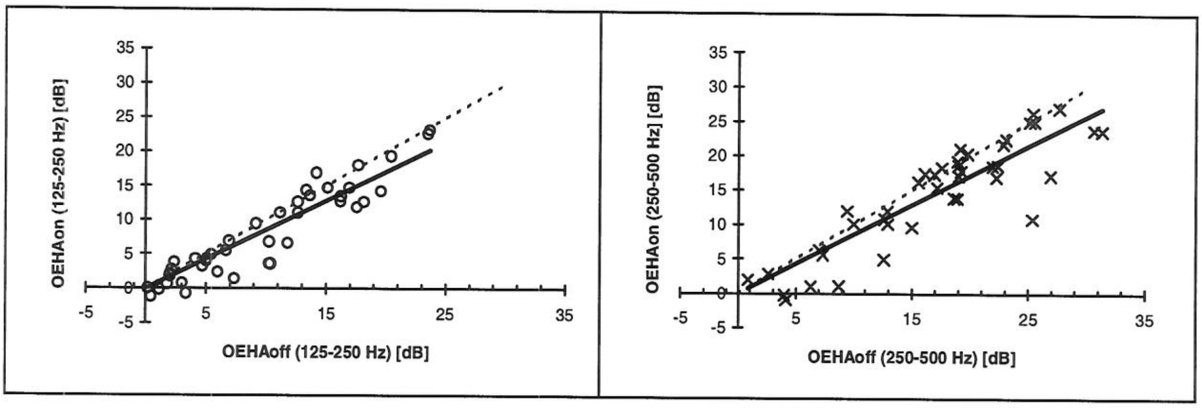


Figure 2.1 'Occlusion effect' with the hearing aid turned off (OEHAon) versus the hearing aid turned on (OEHAoff). O and X: real ear measurements. Thick solid line: trend line. Dashed line: OEHAon = OEHAoff. Left: The occlusion effect as an average from 125-250 Hz. Right: The occlusion effect as an average from 250-500 Hz.

The relation between OEHAoff and OEHAon depends of course on the electrical gain in the hearing aid. The gain is adjusted to the hearing loss, so the OEHAon should depend on the hearing loss, if the hearing aid gain was the dominant factor. But as it is illustrated in **Figure 2.2** there is no significant relation between the hearing loss and (OEHAon minus OEHAoff). There is though a tendency to that (OEHAon minus OEHAoff) becomes smaller when the hearing loss is small. This explanation agrees well with the observations in **Figure 2.1**, where OEHAon equals OEHAoff, when the occlusion effect is small, i.e. has a large leakage.

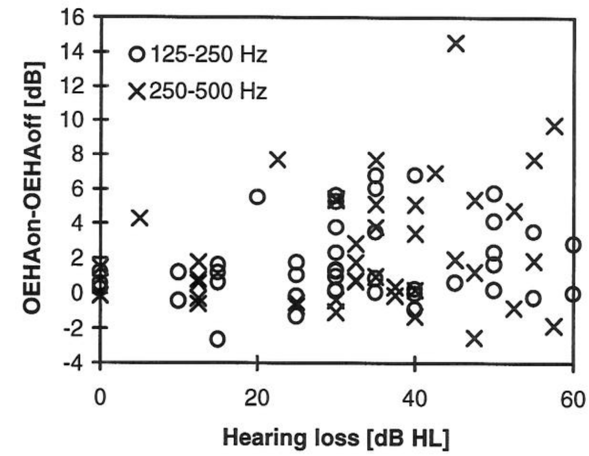


Figure 2.2 Difference between the occlusion effect with the hearing aid turned on and off versus the hearing loss. 125-250 Hz: hearing loss at 250 Hz. 250-500 Hz: hearing loss is an average at 250 and 500 Hz.

2.5 COMPARISION OF OBJECTIVE OCCLUSION EFFECT AND PERCEPTION

The questionnaire results stated that tinnitus subjects experience the occlusion effect significantly more than non-tinnitus subjects. Subjects without tinnitus got significantly more annoyed with increasing sensation level (real ear sound pressure level minus hearing loss) increased. Also, the objective occlusion effect was significantly correlated to the annoyance of their own voice. The relation between experienced occlusion effect, objective occlusion effect and hearing loss is pictured in **Figure 2.3**.

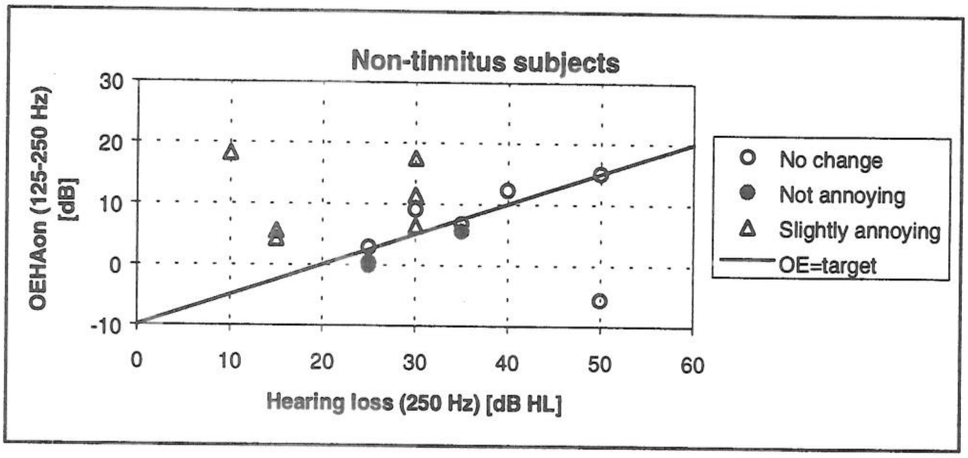


Figure 2.3. Plot of objective occlusion effect (hearing aid turned on) versus hearing loss at 250 Hz with the annoyance from the subject's own voice with occluded ear as parameter. The change in speech level which was recorded with the reference microphone in front of the mouth has not been taken into account. Solid line: OEHAon equals the theoretical target for hearing aid amplification. Each data point counts for 1 subject.

The plot in **Figure 2.3** leads one to make a rule of thumb such that the hearing aid user will experience (and be annoyed) from one's own voice when OEHAon is at least 5 dB larger than the theoretical target amplification. When the OEHAon is at least 5 dB lower than the target amplification, the hearing aid user will not experience occlusion. There is 50% chance to experience occlusion if OEHAon equals the target amplification.

2.6 FURTHER ASPECTS

The user investigation answered some questions but also posed the following questions:

1. The experienced occlusion effect is significantly related to the objective occlusion effect and hearing loss with non-tinnitus subjects. Therefore it makes sense to measure the objective occlusion effect in the hearing aid clinic in order to predict whether or not the hearing aid user will be annoyed by occlusion. It requires additional equipment than is used today. For example, equipment that has the possibility to average the input signal in 60 seconds or a 2-channel analyser, such that two probemicrophones can be used to measure the sound pressure in the right and left ear simultaneously.

2. The survey did not comprise enough subjects to evaluate the effect of monaural contra binaural fitting. It seems logical that binaural fitting would give a more intense feeling of occlusion but a pilot study with normal hearing subjects showed that it might also be the opposite because one's own voice sounds unbalanced in the left and right ear. In order to evaluate the effect of the fitting, it will be necessary to carry out a survey including two test groups: monaural and binaural fittings.

3. The great inter-subject variation in occlusion effect can have several explanations, for example that the energy of the body conducted sound varies a lot and/or that the leakage between the earmould and the ear canal wall is different from person to person. The inter-subject variations cannot be explained without a greater understanding of the mechanism of occlusion

3. LITERATURE REVIEW

Several acoustic models of the ear canal exist in the literature, but only very few try to model the occlusion effect. Three models are reviewed here. All three models assumes that the skull is stimulated directly with a bone conductor. The conclusion is that none of the models is designed especially to predict the occlusion effect created by one's own voice (speech).

3.1 SCHROETER AND POESSELT'S MODEL

The most comprehensive work about occlusion was done by Schroeter and Poesselt [1986]. Their model served the purpose of evaluating the attenuation for ear muffs and ear plugs and it concentrated on the occlusion effect created with a bone conductor. It includes physiological noise, but not one's own voice. **Figure 3.1** illustrates the model.

Model of the external ear
One dimensional waves propagate in the central part of the ear canal. The region in front of the eardrum was ignored. The ear canal was modeled as a cylindrical tube with a diameter of 7.5 mm and length 18 mm. It was found that the best results were obtained when the soft active part of the ear canal was 15 mm long. The radiation effect from the pinna could be simulated by an extension of the transmission line with 10.5 mm.

Model of the middle ear
Schroeter and Poesselt found that in order to estimate the correct insertion loss of earplugs it was necessary to reproduce the eardrum and middle ear impedance rather accurately. They tried several middle ear models and found that a modified Zwislocki model was the best one. The Zwislocki model was modified by Shaw and Stinson [1981].

Occlusion effect
The pressure in the occluded ear was assumed to be generated by two sources: one placed in the middle ear and one placed in the cartilaginous part of the ear canal.

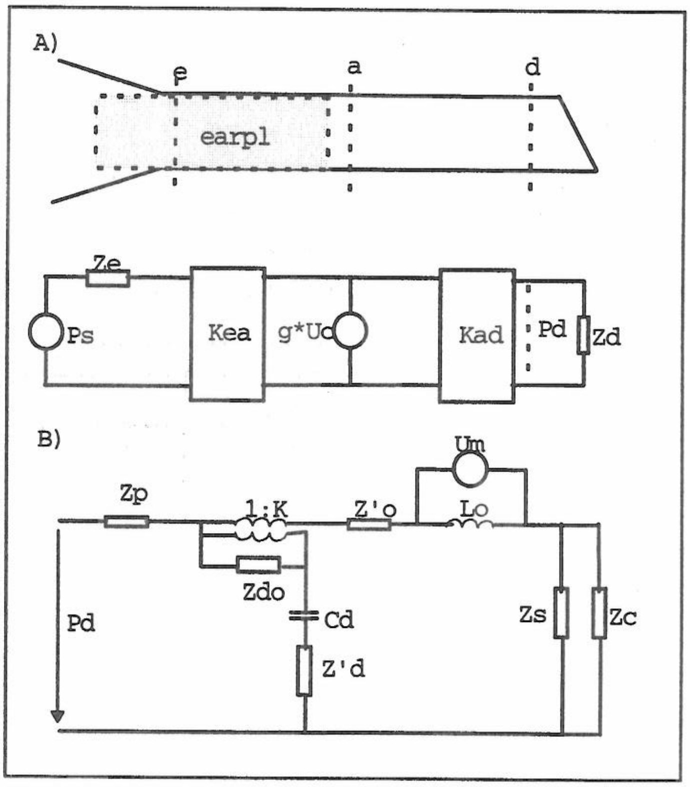


Figure 3.1. An illustration of Schroeter and Poesselt's model. A) Outer ear model. Kea, Kad = chain matrices for ear canal; Zer = impedance of sound source outside the head; Zd = acoustic impedance of eardrum; Ps = external sound source; Uc = volume velocity source; Pd = sound pressure at reference place 'd'. B) Middle ear model. Zp = middle ear cavities; Zdo = flaccid part of eardrum; Cd, Z'dr = part of eardrum coupled to malleus; Z'o, Lo = Incudomallearis; Zs = Incudostapedia; Zc = cochlea, Um = volume velocity source. [Schroeter and Poesselt, 1986].

In the middle ear, a volume velocity generator was inserted to simulate that the vibrations in middle ear cavity walls excite the inertia (the mass) of the malleus and incus. The movements of the ossicles generates a volume velocity into the inner ear but is also excites the eardrum from behind. The other sound source was assumed to be vibrations of the cartilaginous part of the ear canal. This source was modeled as a point volume velocity generator in one point in the ear canal. A gain factor (in reality an attenuation factor) was applied to compensate for the part of the ear canal wall that was covered by the earplug.

The volume velocity of the middle ear source was calculated based on measurements of the sound pressures in front of the eardrum at air conduction, Pdt, and bone conduction, Pdc, threshold respectively. The cartilaginous volume velocity is then estimated from the calculated middle ear volume velocity and Pdt. The hereby modeled occlusion effect agrees well with experimental data from Berger and Kerivan [1983] at shallow and normal insertion depths.

3.2 WEINRICH'S MODEL

Weinrich [1986] has also made a model consisting of coupled cylindrical tubes. The open ear canal is modeled as a rigid uniform 25 mm long tube with a diameter of 8 mm. The concha is a 10 mm long tube and diameter 15 mm and the radiation is a pure resistance, 15 Pa·s/m. The reference does not mention how the eardrum and the middle ear impedance is modeled. The closed ear model is drawn in **Figure 3.2.**

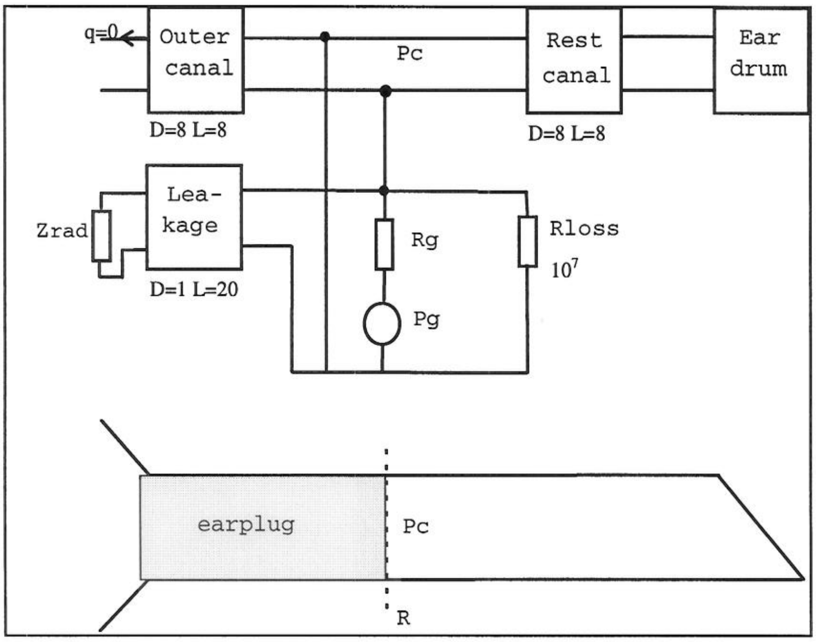


Figure 3.2. Model of the closed ear with leakage. Rg, Pg = vibrations of ear canal walls. D: diameter, L: length. Z_{rad} = radiation impedance. Acoustical units. [Weinrich, 1986].

The earmould is assumed to be a standard earmould and the residual occluded ear canal is 16 mm long. The model accounts for leakage between earmould and ear canal walls and friction between earmould and ear canal walls, which arises when the earmould moves relatively to the ear canal walls in the direction of the canal axis. The leakage is modeled as a 20 mm long cylindrical tube of diameter 1 mm. Frictional loss between the earmould and the ear canal wall is included as a pure resistance. The reference does not state how the value of the resistance was determined. The internal source here is a pressure source placed in one single point in the ear canal.

Calculations of the occlusion effect was in good agreement with real ear measurements made by Lundh [documented by Wimmer, 1986]. The largest mean deviation was 5dB at frequencies below 2 kHz. Apparently, the model seems to work well, but Lundh's measurements were done with very tight moulds made of impression material. Lundh's occlusion effect by one's own voice at 100 Hz is about 10-15 dB greater than occlusion effects measured with standard acrylic moulds determined in investigations reported in the literature. With a less tightly fitted mould, external sound will enter the occluded ear canal. External sound is not included in Wienrich's model, so the model can only be used on one's own voice if the earmould is fitted very tightly.

The difference between Weinrich's model and others is that Weinrich includes the frictional loss between the earmould and the ear canal wall. In order to see the affect of this loss, the occlusion effect is calculated with Weinrich's model with and without the loss resistance. Due to uncertainty about how Weinrich modeled the eardrum and the middle ear, this was modeled with a modified Zwislocki network, see appendix H. Apparently, the frictional loss in Weinrich's model reduces the occlusion effect a lot, see **Figure 3.3**. The resonance effect of the leakage at 280 Hz disappears because of the friction between the earmould and the ear canal wall.

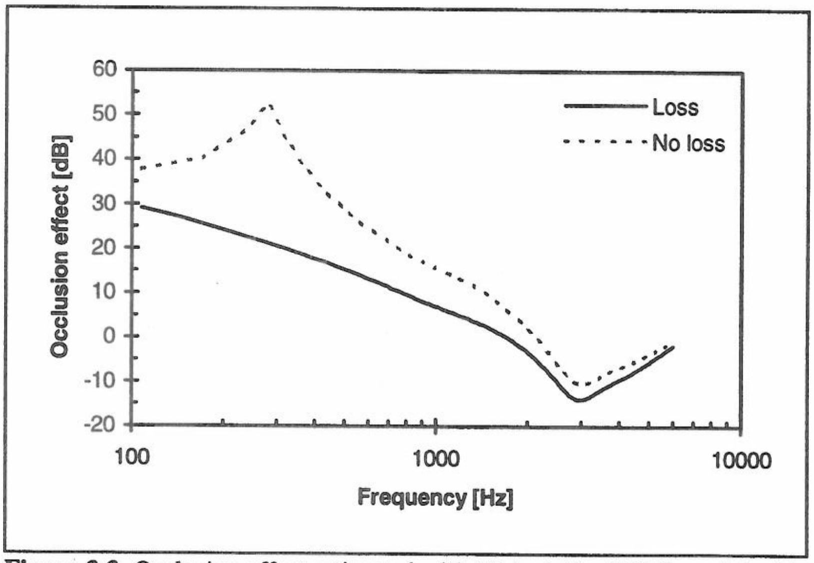


Figure 3.3. Occlusion effect estimated with Weinrich's [1986] model with and without frictional loss.

3.3 WILLIAMS AND HOWELL'S MODEL

Williams and Howell [1990] introduce a comprehensive model of the ear canal. Their model was not meant to estimate the occlusion effect, but it is still a model to predict the sound pressure in the closed ear canal when the skull was stimulated in the vertical direction. Their aim was to compare the pressure in normal ears and pathological ears, but the model for normal ears is only reviewed here.

The model is applied as a lumped element electrical network, see **Figure 3.4**. The model of the middle ear is based on Zwislocki's middle ear model. If the model is compared to Schroeter and Poesselt's model, then the outer ear is applied at point 'd' in Schroeter and Poesselt's model. The vibration of the skull was simulated by adding three sources to the middle ear; a force that directly drives the independent part of the eardrum (Vd), a force driving the incudo-stapedial joint (Vo) and a force applied to cochlea (Vo+Vs). No force was applied in the outer ear.

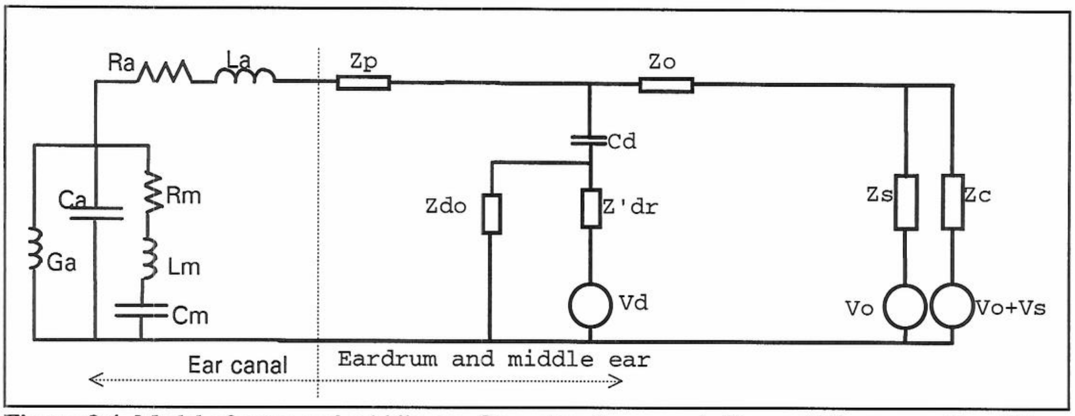


Figure 3.4. Model of outer and middle ear. Za = closed ear canal; Zp = middle ear cavities; Zdo = flaccid part of eardrum; Cd, Z'dr = part of eardrum coupled to malleus; Zo = Incudomallearis complex; Zs = Incudostapedia; Zc = cochlea, Vd, Vo,Vs = volume velocity source. [Williams and Howell, 1990].

The ear canal is applied with lumped element components. The enclosed cavity, C_a, viscous flow loss, R_a, and the air, L_a. A loss is applied across C_a due to thermal conduction, G_a, and due to flexure of the ear canal wall, L_m, R_m, C_m.

3.4 DISCUSSION

Three models of the occluded ear canal were reviewed. All three models are developed to estimate the sound pressure in the ear canal when the skull is excited. Weinrich's model and Shroeter's and Poesselt's models seem to give good results. Williams and Howell's calculated results are not compared graphically with measured data.

Frictional loss between the earmould and the ear canal wall is important in Weinrich's model. Schroeter and Poesselt did not apply a skin impedance for earplugs, only in their model for earmuffs (not shown here).

Most important is that it must be concluded that none of the models account for air transmitted and body conducted sound at the same time. Hence, a new model must be developed in order to estimate the occlusion effect by one's own voice. It seems reasonable to base a model on Schroeter and Poesselt's model because they made many detailed analyses of their model.

4. INTRODUCTION TO SPEECH PRODUCTION

This chapter is a short introduction to the production of speech. The difference between voiced (vowels and voiced consonants) and unvoiced (consonants) are explained. The important speech organs and muscle connections to the ear are illustrated.

4.1 ANATOMY OF THE SPEECH ORGANS

4.1.1 The airway

The production of speech can simply be explained as air flowing through a system of tubes and cavities that continuously changes in geometry. An air flow is for example produced by breathing. Normal breathing cannot be heard, but heavy breathing is audible. Air from the lungs are pressed up through the vocal tract (trachea) and up to the oral and nasal cavities when exhaling, see **Figure 4.1**. The air flows out of the nose and the opening of the mouth through the passway formed by the lips. The air pressure just below the air space between the vocal folds (glottis), see **Figure 4.2** is called the subglottal pressure and it must be higher than the pressure above the glottis in order to create an airflow.

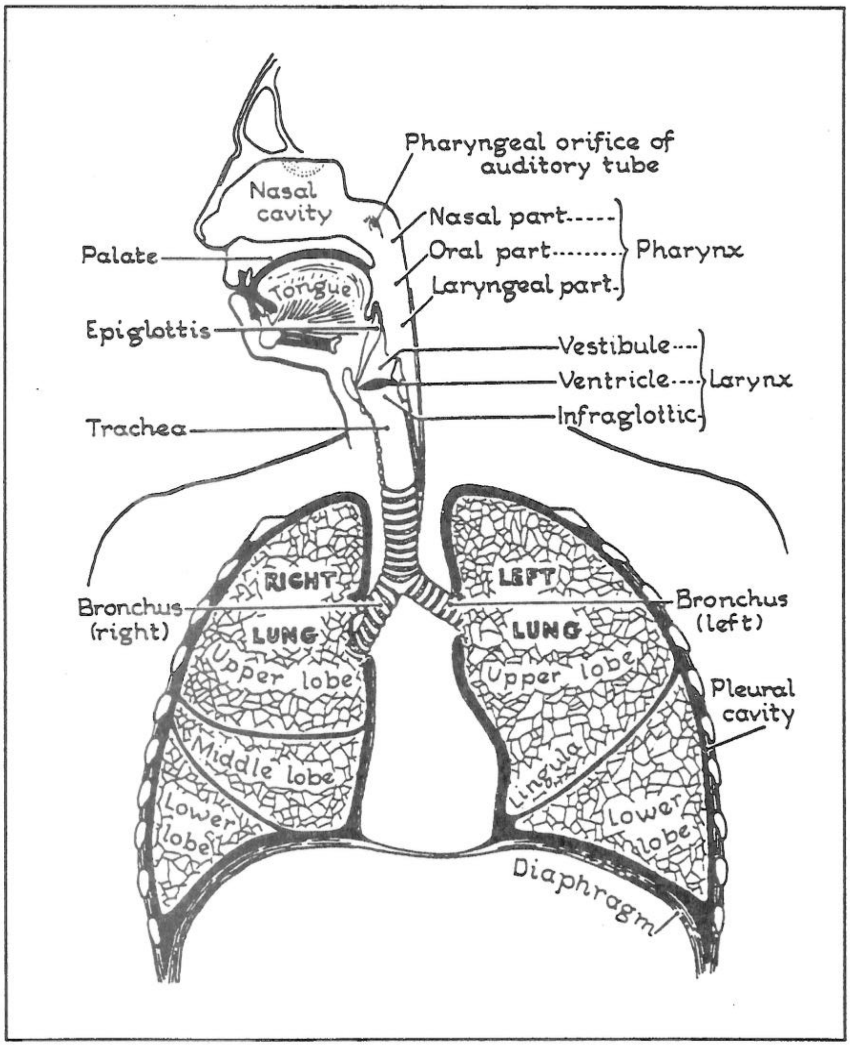


Figure 4.1 The human system of air filled tubes and cavities. Trachea = vocal tract. From: [Borden and Harris, 1980].

Respiration is normally periodic, but when we speak, the respiration becomes aperiodic because the rate of expiration depends upon the contents of the speech. The intensity of the voice is controlled by the subglottal pressure. A doubling of the subglottal pressure increases the sound intensity by 9-12 dB, [Borden and Harris, 1980].

4.1.2 Larynx

The larynx, see **Figure 4.2**, is located in the frontal part of the neck. The angle between the right and left thyorid cartilage is known as the Adam apple.

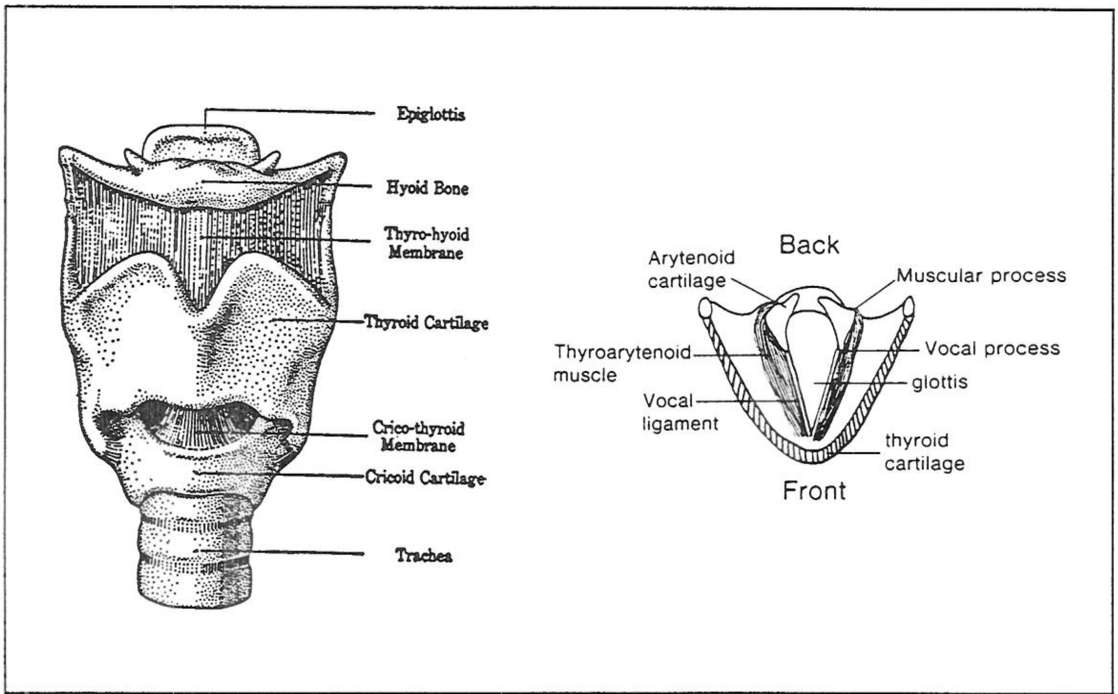


Figure 4.2. Left: Sagittal view of the framework of larynx. Right: Superior view of larynx. From: [Borden and Harris, 1980].

The vocal folds lie vertically in the anterior-posterior direction just behind the thyorid cartilage, see **Figure 4.2**. This cartilage is attached to the thyroid muscle, which runs up to the hyoid bone. Muscles from the hyoid bone are directly connected to the bone around the outer ear, the temporal bone, see **Figure 4.3**. The stylohyoid muscle runs from the hyoid bone to the styloid process on the temporal bone. It elevates and retracts the hyoid bone and thereby the larynx. The digastric muscles run from the hyoid bone to the mastoid and the jaw. They elevate the hoyid bone and assist in opening the jaw. The temporal bone makes the bony ear canal wall. Hence, the pathway from the vocal folds to the outer ear goes via muscles and ligaments.

4.1.3 Vocal folds

The outer most part of the vocal folds are covered with a 0.05-0.10 mm thick skin layer. The superficial layer contains elastin fibers that run in the anterior-posterior direction, see **Figure 4.2**. These fibers are also found in the cartilage of the soft ear canal. Because the fibers are elastic the folds can be streched 3-4 mm.

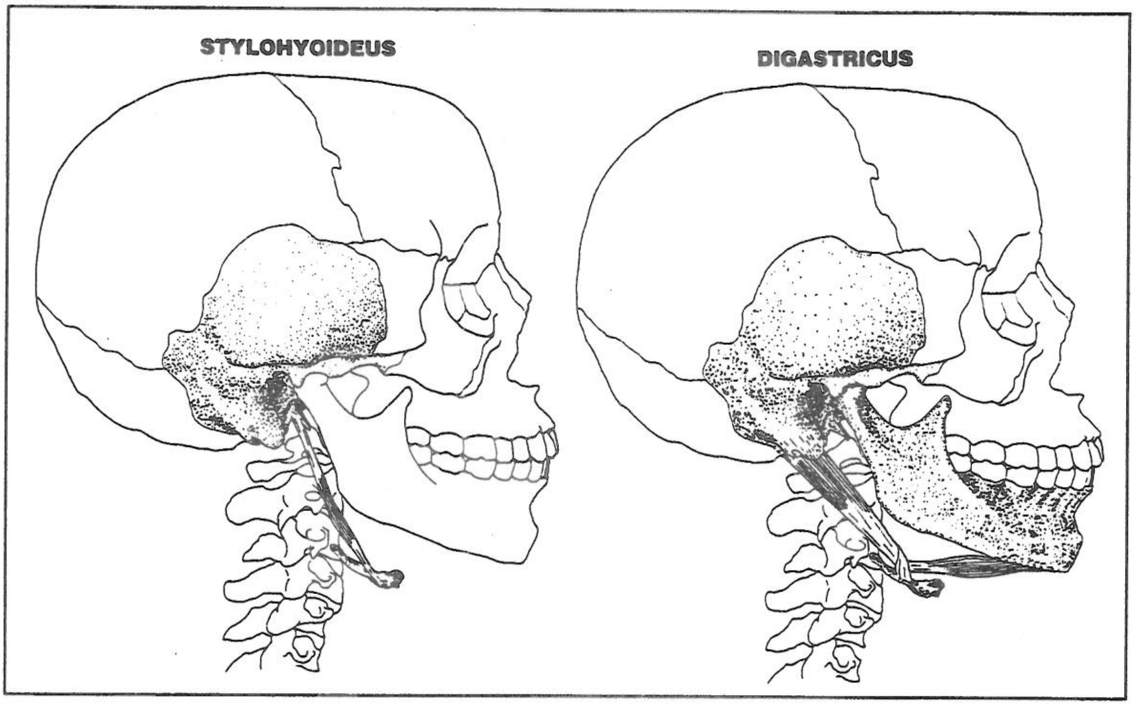


Figure 4.3. Left: Stylohyoid muscle. Right: Digastric muscles. From: [Stone and Stone, 1990]

Air pressure from the lungs makes the vocal folds open. When the folds are apart, then air will flow upwards through the opening because the subglottal air pressure is larger than the pressure above the glottis. When the air has to pass through the narrow passage between the vocal folds, the air flow velocity increases and causes the pressure against the vocal folds to decrease. The pressure drop makes the vocal folds close again. This is called the Bernoulli effect. The Bernoulli effect makes the vocal folds vibrate. The vocal folds will in this way open and close periodically and can be controlled.

Opening and closing of the vocal folds acts like a saw tooth generator. The time function of the velocity of air volume passing through the vocal folds and the open glottal area between the vocal folds is a train of triangle shaped pulses. Transformed into the frequency domain these pulses produce a line spectrum that decreases with 12 dB per octave, see **Figure 4.4**. This spectrum is named the glottal source spectrum.

The opening and closing is controlled by the cricothyroid muscles that stretch the folds, longitudinal tension. The fundamental frequency of a voice depends on the length, longitudinal stress and tissue density of the vocal folds. Typical lengths for the vocal folds are 17-24 mm for men and 13-17 mm for women, [Borden and Harris, 1980]. Consequently, a male voice has a lower fundamental frequency (f_0 = 125 Hz) than a female voice (f_0 = 200 Hz). The fundamental frequency is controlled by the muscles which we use especially for singing and it equals the number of vocal folds openings and closings per second. Changing of the fundamental frequency during for example speech is called intonation, [Stevens, 1997].

Besides that the vocal folds open and close, the edges of the folds vibrate too. The vibrations modes is heard as what one could call, the 'quality' of a voice. For example, if the folds are swollen because of a cold, the voice sounds hoarse.

4.2 PERIODIC AND APERIODIC SOUNDS

Sounds can either be periodic, aperiodic or mixed. Periodic sound are created by vocal fold vibrations as described above. Aperidoc sounds are created in the vocal tract and mixed periodic and aperiodic sounds are a mix of the two.

4.2.1 Periodic voiced sounds

The fundamental frequency of a periodic sound is created in the glottis. The sound produced in the glottis is modified in the vocal tract and in the oral cavity. The radiated pressure from the mouth is the result of the glottal source and the filtering in the vocal tract and the mouth cavity. A very simple model of vowel shaping is illustrated in **Figure 4.4.** The vocal tract and oral cavity is modeled as a uniform tube and the radiation from the mouth as a piston in a sphere.

The first line in the spectrum represents the fundamental frequency, F_G, the next lines are the harmonics and the peaks are the formants,. F1-F4, of the vowel. The fundamental frequency has most energy in the source spectrum but not necessarily in the radiated sound.

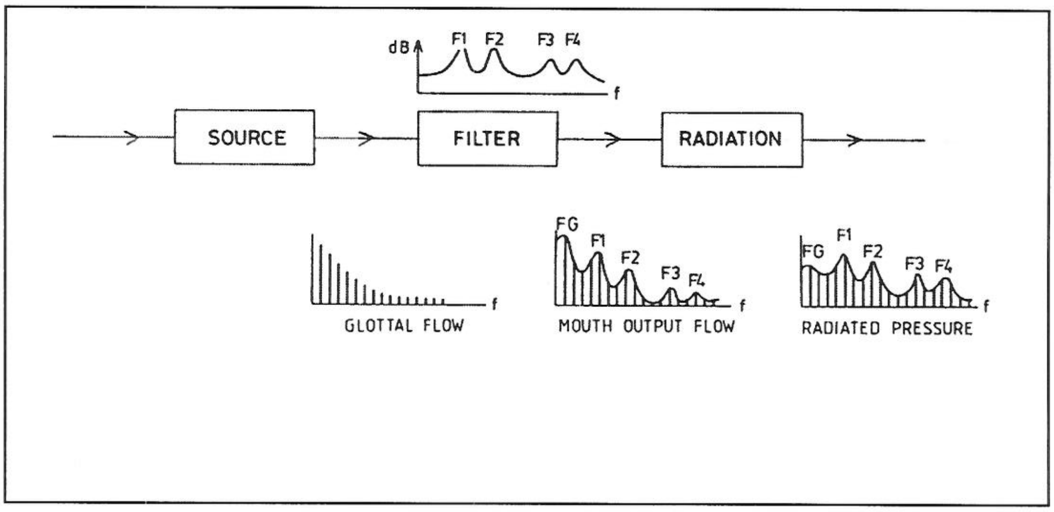


Figure 4.4. Voiced sounds. Source = airflow through glottis. Filter = vocal tract and mouth cavities. Radiation = radiation from mouth in a sphere (6 dB / octave). From [Fant, 1983].

The fundamental frequency of a vowel is created by the vocal cords and the formant frequencies occur because of resonance in the vocal tract and the oral cavities. Anatomically the vocal tract and oral cavity are not tube shaped. The shape of the oral cavity can be changed by tongue and jaw movements and the radiated spectra depends on the shape of the oral cavity and the lips.. The three vowels where the oral cavity, tongue and lips are in the most extreme positions, are called the cardinal vowels and they are: /i/, /a/ and /u/. The fundamental frequencies of the cardinal vowels and the first formant determined from 76 speakers are outlined in **Table 4.1**. The duration of one period in for example the /a/ is for men = 1/124 Hz = 8 ms. In continuos speech a periodic sound is sustained for several periods. The cardinal vowels are often used to measure the occlusion effect from one's own voice.

/i/ like in (beet)	/a/ like in (hat)	/u/ like in (boot)
Fundamental: Men = 136 Hz Women = 235 Hz	Fundamental: Men = 124 Hz Women = 212 Hz	Fundamental: Men = 141 Hz Women = 231 Hz
1. formant : Men = 270 Hz Women = 310 Hz	1. formant : Men = 730 Hz Women = 850 Hz	1. formant : Men = 300 Hz Women = 370 Hz
The tongue moves up and forward creating a small cavity. Lips shaped like a smile.	The cavity is made larger and the pharyngeal cavity is narrowed by lowering the tongue backwards and opening the jaw. The mouth is wide open.	The vocal tract is prolonged by protruding the lips or lowering larynx. At the same time the tongue goes backwards.

Table 4.1. The cardinal vowels. From [Borden and Harris, 1980].

4.2.2 Aperiodic sounds

Consonants are aperiodic sounds which are created in the vocal tract and oral cavity. In contrary to periodic sounds the vocal folds do not vibrate. An aperiodic sound is a stop, a fricative or an affricate and the time duration of the sound is much shorter than the duration of a vowel, [Borden and Harris, 1980]. An aperiodic sound has more energy between 2-4 kHz than a vowel

Stops and plosives are made by closing the oral cavity. When the air pressure increases rapidly and the air suddenly is released because the closing is relaxed, a transient sound is produced e.g. /p,b,t,d,k,g/ as in 'pie', 'buy', 'two', 'do', 'cow' and 'go'.

Fricatives is the sound of noisy vibrations in the air stream in the vocal tract. Constrictions are formed in the vocal tract and when an air stream passes through the friction, it creates vibrations and the air is pressed out between the lips. An /s/ is a fricative.

An affricate is a stop with fricative release where the lips are rounded and the tongue is retracted. In English there are two affricates as in 'chair' and 'jar'.

4.2.3 Continuos speech

The speech spectrum is often calculated from running speech during for example 60 seconds. The radiated sound pressure 1 m in front of the speaker is 65 dB SPL for men and 62 dB SPL for women but the dynamic range over a short time (less than a few hundred ms) is 30 dB, [Elberling and Nielsen, 1993]. The radiated spectrum is language independent (for most languages), [Byrne et al., 1994]. A typical speech spectrum is shown in **Figure 4.5**.

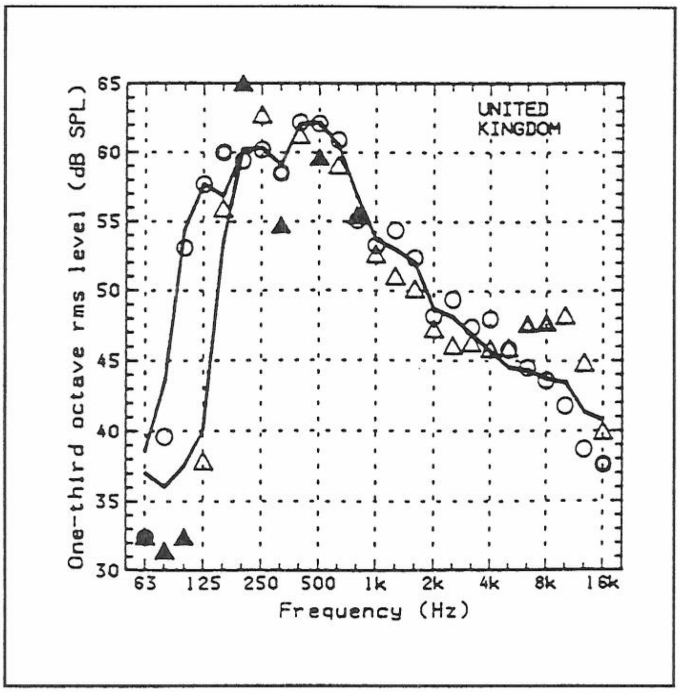


Figure 4.5. Typical speech spectrum 1/3 octave filtered. Solid lines: average for men and women respectively. [Byrne et al., 1994].

5. ANATOMY OF THE EAR

This section is not meant to be a complete or even a fully detailed description of the anatomical ear. It comprises only the general anatomy of the ear and the details worth knowing in relation to the occlusion effect. The perception of airborne sound is roughly speaking a three step procedure: first an acoustical transmission, then a mechanical transmission, and at last an electrical transmission. The transmissions take place in the outer ear, the middle ear and the inner ear as sketched in **Figure 5.1**.

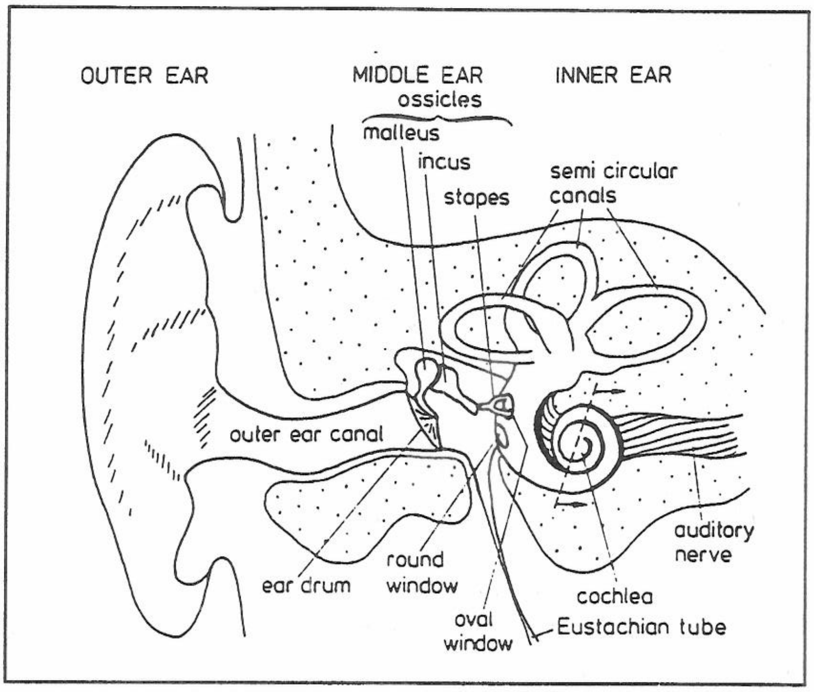


Figure 5.1. Sketch of the ear. From [Zwicker and Fastl, 1990].

5.1 THE OUTER EAR

5.1.1 Pinna

The outer ear is visible by eye and by otoscope. The most outer part is the pinna, which is composed of skin-covered elastic cartilage containing some muscles. **Figure 5.2** left illustrates the pinna and the important landmarks.

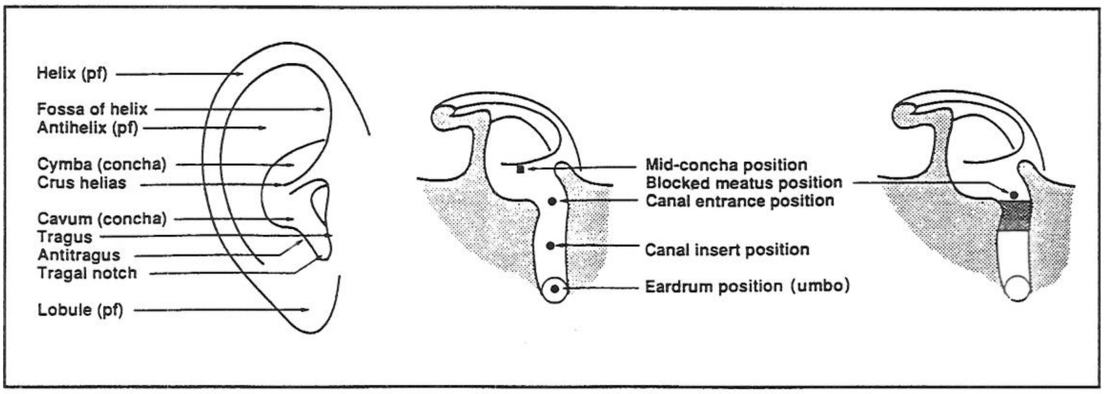


Figure 5.2. Left: Illustration of pinna. Right: The ear canal entrance position From: [Shaw, 1997].

5.1.2 The ear canal

5.1.2.1 Ear canal size and shape

The concha expands into the ear canal (acoustic meatus) or rather the ear canal begins where the concha ends but on a real ear it can be difficult to determine. The author finds that the best way of defining the ear canal opening is to look at the posterior ear canal wall where it makes a sharp edge with the concha, as illustrated in **Figure 5.2**, right. The dimension and shape of the ear canal is age related and varies significantly between individuals as shown in **Figure 5.3**. Adult shape and size of the ear canal and pinna is present already in the age of 7-9 years. The tissue is constantly in change and this is why a hearing aid might feel loose after some years of use, [Ballachanda, 1995].

Some ear canals are nearly straight whereas others bend sharply. The ear canal has two bends when seen from the upside. The diameter along the length axis varies within the single ear canal with the largest cross section at the opening. The lengths and diameters also vary between individuals.

Several authors have measured the dimensions of ear canals on adults but only some shall be mentioned here. A more detailed literature view can be found in Ballachanda [1995]. It is difficult to define the size and shape of the ear canal because the canal diameter varies over length and the end at the eardrum is wedge shaped. Anatomically seen, the anteroinferior wall is 6 mm longer than the posterosuperior wall but average data used for acoustical purposes neglect this and concentrate on the average ear canal length [Ballachanda, 1995]. Zwislocki [1980] writes that the average length is 23 mm and the average diameter is 7.48 mm.

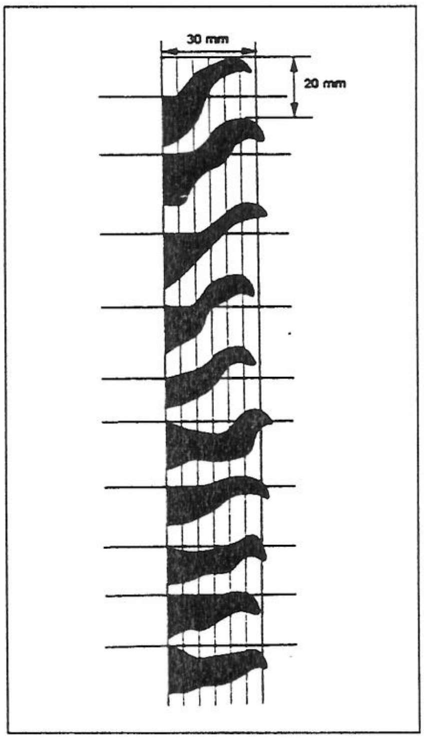


Figure 5.3. Examples of ear canal shapes and dimensions.Frontal view [Salvenilli et al., 1991].

Stinson and Lawton [1989] specified the geometry very carefully along a curved axis see **Figure 5.4** The accuracy of the measurements were 0.03 mm. Fourten subjects were measured, ages 44-87, and the average length of the ear canal following the curve of the canal was found to be 30.4 mm on a curved axis, which is a 3-4 mm greater than on a straight axis in the same ear canal. Their results agree well with measurements made by Hudde [1980] who looked at the 18 mm in the middle of the ear canal only. **Figure 5.4** right also illustrates that the ear canal size varies a lot between individuals.

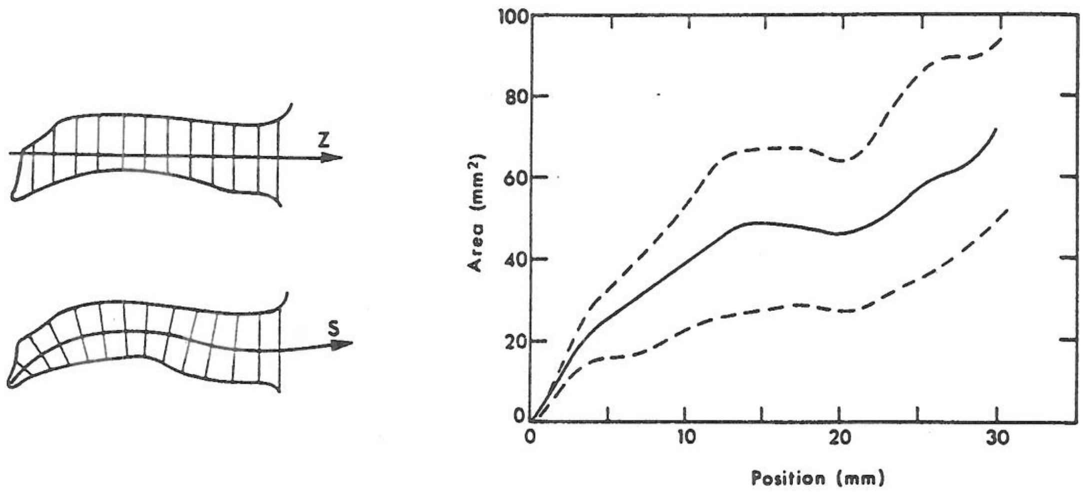


Figure 5.4. Left: Curved and straight length axis. Right: Ear canal cross sections measured on a curved axis. Solid line: average. Dashed lines: individual variations from the average. Eardrum located at position = 0 mm. [Stinson & Lawton, 1989].

Johansen [1975] measured the volume along a straight axis from the point of junction between the eardrum and the inferior wall. He made impressions on 10 cadavers (6 men and 4 women, mean age = 63 years) and found an average length of 25.7 mm and a volume of 1014 mm^3, which gives a diameter of 7.1 mm if the ear canal is a uniform tube. The axis is also illustrated in **Figure 5.4**. Johansen measured an average length of 25.7 mm, and correcting the length to the umbo (the point where the middle ear ossicles are connected to the eardrum) on the eardrum by subtracting 2 mm, it reveals to 23.7 mm, which agrees with Salvenilli et al.'s [1991] data. But Johansen's ear canal volume is greater.

Men's ear canals are in general larger than women's. According to Salvenilli et al. [1991] the average length is 23.5 mm and the average diameter 7.1 mm, see **Table 5.1**. They measured the length and diameter on 140 cadavers (making 280 ears) from 18 to 60 years old. The ear canal length was defined as the length from the bottom of concha to the umbo located at the center of the eardrum.

	Men	Women	Average
Length	25.2±2.6 mm	22.4±2.3 mm	23.5±2.5 mm
Largest diameter	9.7±1.5 mm	8.5±0.7 mm	9.4±1.5 mm
Smallest diameter	5.1±0.7 mm	4.4±0.3 mm	4.8±0.5 mm

Table 5.1. Average ear canal dimensions measured on 280 ears. [Salvenilli et al., 1991].

The different results obtained in the studies are caused by several factors: the individual ear canal size, the subject's age, the number of men and women (men's ear canals are larger than women's), the measurement method and the definition of the ear canal opening and the end of the ear canal. Johansen's data are greater than Salvenilli et al. most likely because Johansen used older subjects. Stinson and Lawton [1989] found an average ear canal length, that is longer than any other results in the literature. The first reason is that they measured the length of the curved axis, whereas nearly all other studies has measured the length of the straight axis.

5.1.2.2 Ear canal tissue

The outer part of the ear canal is cartilaginous and the inner part is surrounded by the temporal bone, see **Figure 5.5**. The bone is attached to the lower jaw (mandible) at the temporomandibular joint. Both the superior and the inferior wall are closely connected to the parotid gland, which consists of clusters of cells and systems of tubes that lead into the oral cavity.

The presence of elastic cartilage in the ear canal is important in relation to occlusion of the ear canal because it is believed that the sound produced by the person's own voice is transmitted to the ear canal mainly via the elastic cartilage.

The cartilage is directly connected with the cartilage of the pinna. This cartilage contains elastic fibers which makes it very flexible. The cartilage is covered by a 0.5-1.0 mm thick skin layer that holds hair follicles and ceruminous glands, which produce cerumen (ear wax).

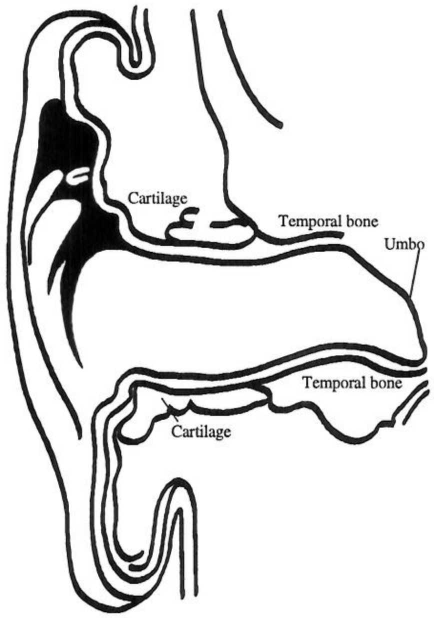


Figure 5.5. The ear canal (acoustic meatus).

In the bony part of the ear canal the skin layer is thinner approximately 0.2 mm and this skin is continuos with the skin covering the eardrum (tympanic membrane). Because of the thin skin layer in the bony part of the ear canal, the sensitivity to mechanical touch is high. Hair and cerumen are not produced in the bony part.

The respective lengths of the cartilaginous and bony part are different from person to person. Some authors are of the opinion that for an average person the cartilaginous part reaches 1/3 of the way into the ear canal [Gelfand, 1990] and [Miyamoto and Miyamoto, 1995]. Others say that the bony part only covers the inner 1/3 part leaving 2/3 of the ear canal to be cartilaginous [Garcia, 1994]. Zemlin [1968] and Oliveira [1995] claim that the cartilaginous part in the average ear reaches 1/3-1/2 of the way into the ear canal.

Oliveira [1991] scanned the ear canal of a number of subjects using a magnetic resonance technique. The ear was scanned from the eardrum and out moving in steps of 1.1mm. As mentioned before, the ear canal has two bends when seen from the upside and the scanning showed that the second bend of the ear canal is only 50% surrounded by bone. The bony surface first appears at the posterior aspect and expands to the superior. In the present report it will be assumed that the cartilage reaches ½ way into the ear canal.

The cartilaginous part is slightly concave towards the nose, the medial part bends forward and downward and the inner most part is slightly concave towards the back of the head. Seen from above, the ear canal is S-shaped. The ear canal terminates with the eardrum, that forms an angle with the inferior wall. A standard hearing aid earmould ends between the first and second bend.

5.2 THE MIDDLE EAR

The eardrum is the connecting membrane between the ear canal and the middle ear. The middle ear is an air-filled cavity in which the middle ear bones (the ossicular chain) are placed. The components of the middle ear are illustrated in the sketch **Figure 5.6.**

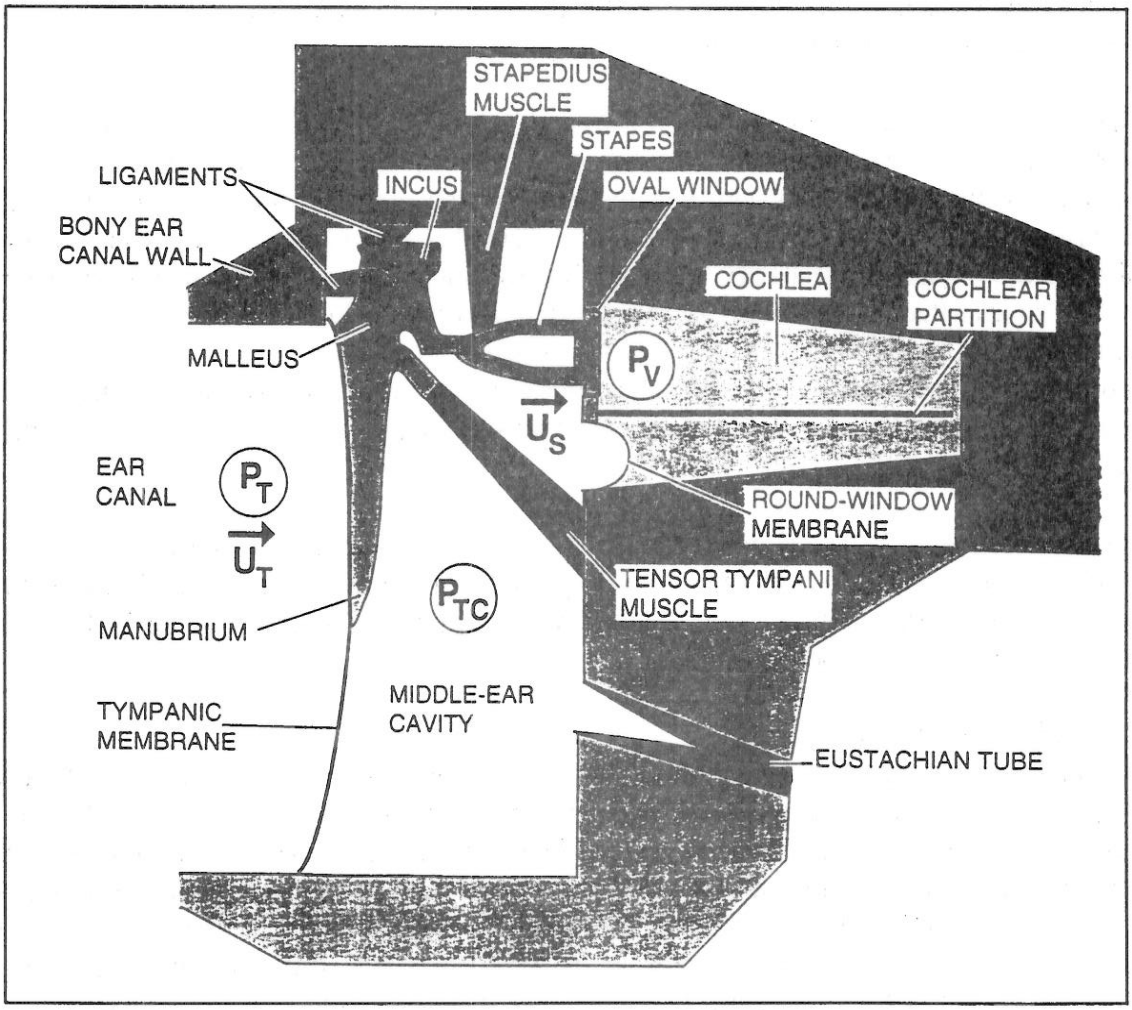


Figure 5.6. Sketch of the middle ear cavity with its bones, ligaments, muscles and the eustachian tube. [Peake and Rosowski, 1997].

The function of the middle ear ossicles is to transmit the sound in the ear canal to the inner ear, or in other words, the middle ear ossicles provide an impedance transformation between the sound transmission in air in the ear canal to sound transmission in the lymph in the cochlea. Pressure variations in the ear canal makes the eardrum vibrate and the vibrations are transferred to the middle ear by the malleus that is attached directly to the inner side of the eardrum. Vibrations of the malleus is transferred to the incus and the stapes and further to the cochlea in the inner ear via the footplate of the stapes. At $f < 5$ kHz the sound pressure is nearly the same over the eardrum surface and the footplate of the stapes makes piston-like movements. By this transmission, the sound pressure in the inner vestibule just behind the stapes becomes 20-30 dB greater than the sound pressure in the ear canal just in front of the eardrum. The gain depends on frequency. For example in a cat, the gain grows from 0 dB up to 25 dB from 100-600 Hz and is nearly constant up to 8 kHz, where the gain decreases again, [Peake and Rosowski, 1997].

Apart from transmitting sound from the ear canal to the inner ear, the middle ear protects the inner ear against loud sounds below 2 kHz. The protection gets more efficient as the frequency gets lower. The way it works is, that when the sound pressure exceeds 70 dB SPL, the stapedius muscles contract and tilt the footplate of stapes so the vibrations of the oval window are less efficient. This reflex needs 25-125 ms to go into action, [Poulsen, 1993]. The reflex adapts with time to the incoming sound and it is present for the duration of high level sounds at low frequencies, so in this sense, if the sound pressure from one's own voice is high, the stapedius muscle protects against it.

The tensor tympanic muscle goes into action at yawning and swallowing. It contracts and opens up to the eustachian tube that has it's outlet through the nose. Hereby the static pressure in the middle ear cavity is equalized with the pressure on the other side of the eardrum. Equalizing of the pressure is a necessity because blood in the muscles slowly absorbs the oxygen in the cavity. After a while the static pressure in the middle ear cavity will be smaller than at the outside of the eardrum, and the eardrum will bend inwards trying to make a smaller cavity which increases the pressure. In this situation the eardrum is more tensed and cannot vibrate so easily, hence, the vibration sensitivity decreases which results in a 'hearing loss', because the sound pressure must be higher than normal in order to make the eardrum move.

If the ossicles does not work properly, the transmission of sound from the ear canal to the inner ear is reduced and a conductive hearing loss is introduced. Malfunction in the middle ear affects the perception of sound. For example, otosclerosis reduces the mobility of the ossicles and the hearing becomes impaired.

Clinical occlusion test

Malfunction in the middle ear is commonly indicated with a clinical 'occlusion test'. One way to do it, is to place a bone conductor on the mastoid to the test ear. The bone conductor generates a pure tone at a given frequency and given level. The other ear, the non-test ear, is acoustically masked with noise using a head phone. The hearing threshold is first detected when the test ear is open, then the test ear is covered with an ear muff and the threshold is tested again. The ear canal sound pressure generated by the bone conductor is higher when the test ear is occluded and therefore if the threshold gets better when the test ear is covered with an ear muff then it indicates that the middle ear is intact. If the threshold is the same with the open and occluded test ear, then it indicates that the middle ear does not function normally.

5.3 THE INNER EAR

The inner ear is of less interest than the ear canal and middle ear when talking about the objective occlusion effect. The inner ear will only be mentioned very briefly here, though the inner ear is a much more complicated structure than the ear canal and the middle ear.

The footplate of the stapes in the middle ear (see **Figure 5.6**) is attached to the oval window of the cochlea. The cochlea is a tube coiled up like a snails shell with 2¾ windings. Looking at a cross-section of the cochlea, there are three canals: the scala vestibuli, the scala tympani and the scala media. The scala vestibuli is connected with the oval window and the scala tympani with the round window, which also makes a boundary to the middle ear. The cochlea is filled with lymph and when the stapes excites the oval window the lymph is set into motions. The scala vestibuli and the scala tympani are connected with a small hole (helicotrema) and the pressure is equalized via this hole and the round window.

In the middle of the cochlea is the scala media where the cortic organ is placed. The cortic organ contains outer and inner haircells that help to transmit the information to the brain through a network of nerve fibers as electrical impulses.

Malfunction of the inner ear is associated with a perceptive (sensorineural) hearing loss. Malfunctions in the hair cells or the nerve fibers do not affect the sound pressure in the occluded ear canal.

6. MECHANISMS OF BODY CONDUCTION

The occlusion effect can be experienced just by closing the ears with the fingers and saying /eee/. The /eee/ sound becomes louder than normal. The ear canal pressure increase at low frequencies. Khanna et al. [1976] showed that the auditory sensation depends primarily on the sound pressure level in the ear canal. This corresponds well with the conclusion from Part I of the Ph.D-thesis [Hansen, 1997] that the perception of occlusion and the real ear sound pressure level are significantly correlated, provided that the subjects do not suffer from tinnitus. Tonndorf [1972] showed with experiments on cats, that the sound pressure in the ear canal was the greatest contributor to the total bone conduction response. Hence, it is relevant to study the mechanism of body conducted sound in the ear canal.

Sound in the ear canal is generated either from sound transmitted in the air or an internal sound source in the body. Air transmitted sound enters the open ear canal via the ear canal opening or it enters the occluded ear canal via some kind of leakage in the earmould. Internal sound originates from vibrations in the body, which are transmitted to the ear canal. The mechanism of the occlusion effect is not fully understood and it is necessary to go some years back to find studies of the mechanism. The most comprehensive studies are made by Békésy [e.g. 1939, 1949], Zwislocki [1953] , Huizing [1960], Tonndorf [1972], Khanna et al. [1976] and the newest one by Schroeter and Poesselt [1986]. Békésy [1949] is the reference that addresses one's own voice in greatest detail. These references are all nearly exclusively concerned with the occlusion effect created by a bone conductor placed on the skull. Thus, the following analysis takes its starting point in the hypotheses of the bone conduction mechanism and attempts to apply this to the mechanism of speech production.

6.1 BONE STIMULATION VERSUS ONE'S OWN VOICE

The internal sound is in the literature most often designated as *bone conducted* sound, probably because the occlusion effect has been used to test the middle ear function by placing a bone conductor on the skull. A bone conductor excites the skull bone directly. When speaking, the sound originates at the glottis and travels not only in bones but also in soft cartilage and muscles. Therefore, the term *body conducted* sound seems more suitable.

Because of the different nature of one's own voice and a bone conductor, occlusion effects created by a bone conductor are not exactly the same as the occlusion effect created with one's own voice.

Revit [1992] suggested two methods for clinical evaluation of the occlusion effect. He claims that the magnitude of the occlusion effect obtained with the subjects own voice and a bone conductor placed on the mastoid is nearly the same. However, this result should be utilized with care. Revit compared a sustained /eee/ sound with a bone conductor tone at 250 Hz. It cannot be denied that these two signals will create about the same occlusion effect, but it cannot be concluded that continuos speech and tone signals at other frequencies will create the same occlusion effect. As he writes, the maximum occlusion effect does not always occur at 250 Hz but sometimes at 500 Hz. In fact, bone conductor stimulation seems to create about 5 dB more occlusion effect than one's own voice, see **Figure 6.1**. The data from Wimmer shows data for bone conduction and voice measured in the same ears.

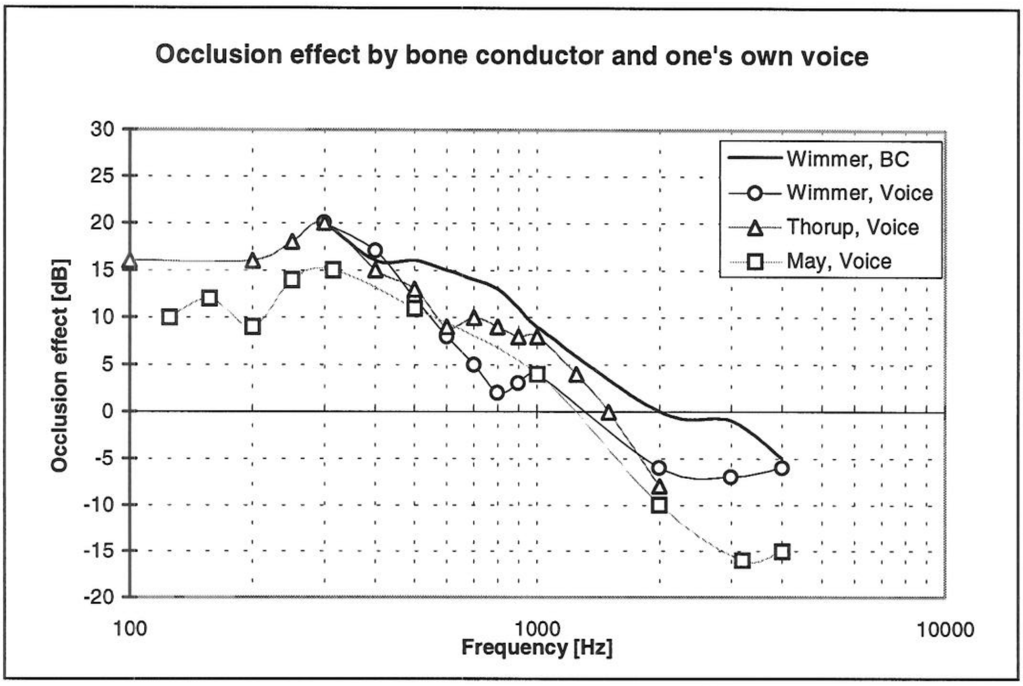


Figure 6.1 Occlusion effects from various studies with one's own voice or a bone conductor, BC, placed on the mastoid. Data from: [Wimmer, 1986]: 4 subjects, ear impression material full concha earmould, [Thorup, 1996]: 16 subjects, acrylic full concha earmould and [May, 1992]: 10 subjects, skeleton earmould.

6.2 THREE COMPONENTS IN BONE CONDUCTION

Hearing by bone conduction when the ear is occluded has been addressed several times in the literature. Wheatstone [1827] was the first to point out that it was easier to hear the tone of a tuning fork placed on the head when the hands were placed over the ears. This phenomenon has been and still is used to detect disorders in the middle ear. Thus, there is plenty of evidence that the transmission of sound to the inner ear is better when the ear is occluded. The bone-conducted sound reaches the inner ear via three pathways due to skull vibrations (a more detailed description can be found in Tonndorf [1972]):

Inner ear component

- Several effects can take place but the idea in general is that sound is generated directly in the inner ear by vibrations in the cochlea (fluid displacement, oval and round window vibrations, blood vessels)

Middle ear component

- Vibrations of the skull stimulate the middle ear bones. The footplate of stapes stimulates the cochlea and hence the inner hair cells.
- Vibrations of the skull create a volume change in the middle ear cavity, which sets the ossicles into motion.

Outer ear component

- Radiation from the ear canal walls create a sound pressure in the ear canal. The sound reaches the inner ear by the normal pathway.

The three major components are illustrated in **Figure 6.2.** Tonndorf [1972] made experiments on cats. The total response due to bone conduction was measured with implanted sensors in the cochlea. Several surgeries were made in order to study the effect of the bone conduction components. The result of the total outer ear, middle ear and inner ear component is shown in **Figure 6.3**. It is clear that the outer ear and middle ear components are dominating. The inner ear component becomes comparable at frequencies above the ear canal resonance (1,000 Hz in cats). In humans the ear canal resonance is about 2.6 kHz. The occlusion effect is only effective below 2 kHz, and the inner ear component is therefore very small at the frequencies where the occlusion effect is interesting. It is therefore adequate to analyze only the contributions of the ear canal and the middle ear in view of the occlusion effect.

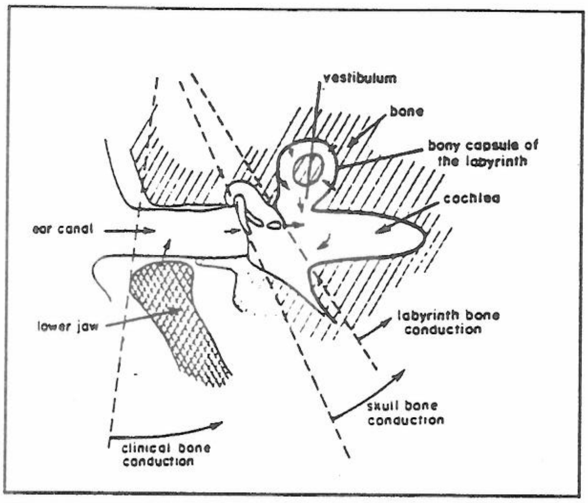


Figure 6.2. Illustration of the inner ear, the middle ear and the outer ear component. Movement of the jaw is also included. From: [Békésy, 1954].

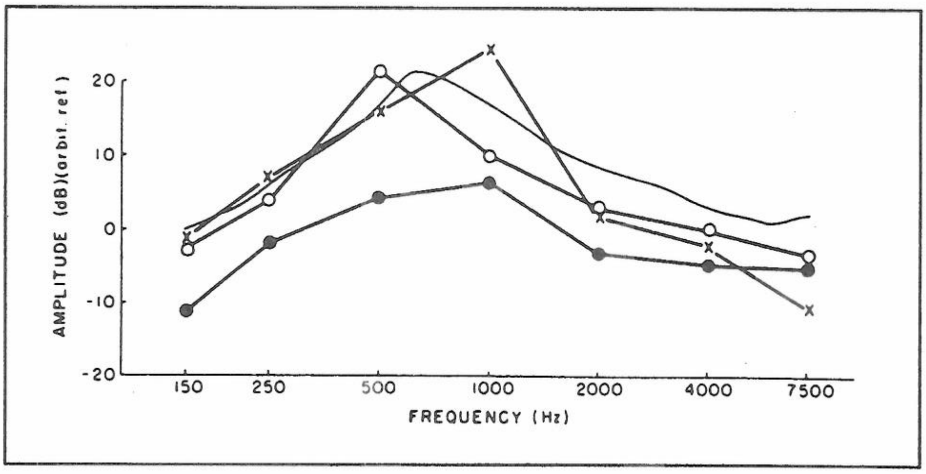


Figure 6.3. Effect of the outer ear, the middle ear and the inner component on the total bone conduction response in cochlea. From: [Tonndorf, 1972]. Curves: x = ear canal, O = middle ear, ● = inner ear, - = total.

6.3 SKULL VIBRATIONS

Studies of the vibration patterns of the skull have been carried out by placing a vibrator directly on the skull and detected the vibrations with some kind of an accelerometer.

Békésy showed in 1932 [in Hood, 1962] that the skull vibrates only as a perfect rigid body below 200 Hz. At 800 Hz the first mode occurs and the forehead vibrates 180° out of phase with the backhead. Up to about 800 Hz, the head moves approximately like a rigid body, but then moves more like a bell at higher frequencies, see **Figure 6.4**. At frequencies between the mode frequencies, flexural waves are transmitted along the flat skull bones.

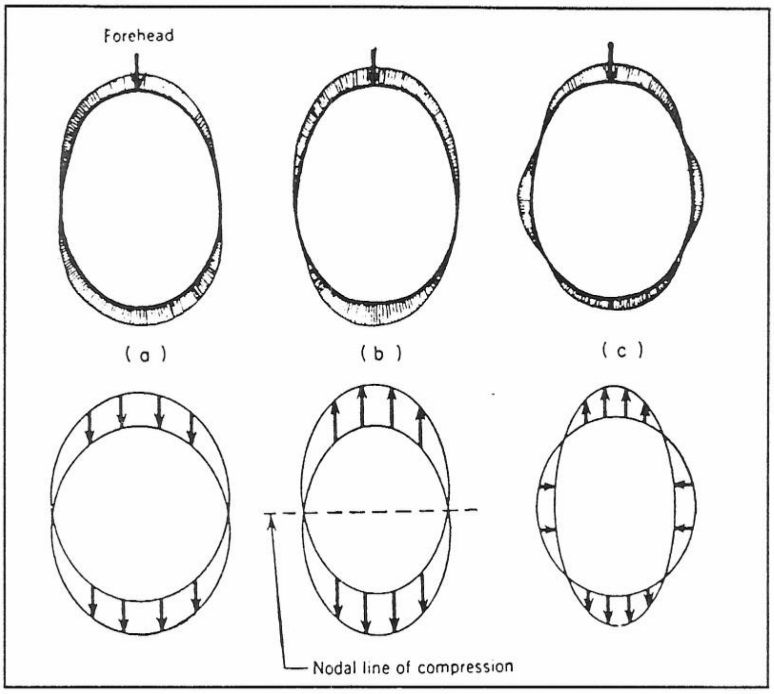


Figure 6.4. Skull vibrations in the three lowest modes when a bone conductor is placed on the forehead. a) 200 Hz, b) 800 Hz, c) 1,600 Hz. From [Tonndorf, 1972].

The findings of Békésy were to a certain degree confirmed by Kirikae [Hood, 1962]. Kirikae found like Békésy that the skull moves as a rigid body at low frequencies (f < 250 Hz) and that the first resonance occurs at 1,800 Hz. Kirikae detected also skull resonances at higher frequencies from 2,000 Hz and up.

In addition Kirikae, [Hood, 1962] measured the vibrations on a skull with and without contents and concluded that the contents of the skull, i.e. the brain, did not alternate the vibrations of the skull. This was later confirmed by Békésy himself.

6.3.1 Skull vibrations excited by one's own voice

According to Békésy [1939, 1949] the pressure changes in the mouth and the vibrations of the edges of the vocal cords sets the whole body into vibrations. The body vibrations were measured with a stethoscope while the test person vocalized. At 100 Hz the amplitudes of the vibrations just under the lower jaw was 14% of the amplitude of the vocal cords' vibrations and 5% under the nose, at the chin and the at the upper part of the mandibular. Békésy argued that the airstream passing through the vocal folds will bring the edges of the vocal cords to vibrate vertically and together with the sound pressure in the mouth cavity the skull will primarily oscillate vertically. This argumentation agrees with the later research of vocal cord vibration, see chapter 4.

Békésy measured the vibration amplitude on the skull while the subject vocalized an /ooo/. The vibration amplitude was much greater in the vertical direction than in the horizontal directions. There was almost no vibration in the right-left direction of the temporal bone. These results are interesting because the temporal bone forms a wall of the bony part of the ear canal and because Tonndorf [1972] concludes that the sound pressure in the occluded ear canal originates both from the cartilage and the temporal bone.

6.4 THE MIDDLE EAR COMPONENT

Producing a voiced or unvoiced sound alters the air pressure in the mouth and nose cavities. The alternating pressure excites the walls in the cavities that are a part of the skull. The skull is, via ligaments, in contact with bones in the middle ear, the ossicles, see chapter 5. The vibrations of the ossicles will probably lag behind the skull oscillations because the ossicles are not rigidly coupled to the skull. The skull vibrations will be transmitted to the ossicles and from the manubrium of the malleus to the eardrum, see **Figure 5.5**. The hypothesis is that when the eardrum moves, the enclosed air volume changes and generates sound.

The middle ear component has not been regarded as primary source to generate sound pressure in the ear canal. The following is a discussion of why it is so.

The prevalent argument to exclude the bone transmission of speech as a cause for the occlusion effect is that the occlusion effect is infinitesimal when the earmould is inserted tightly into the bony ear canal, an example is plotted in **Figure 6.5**.

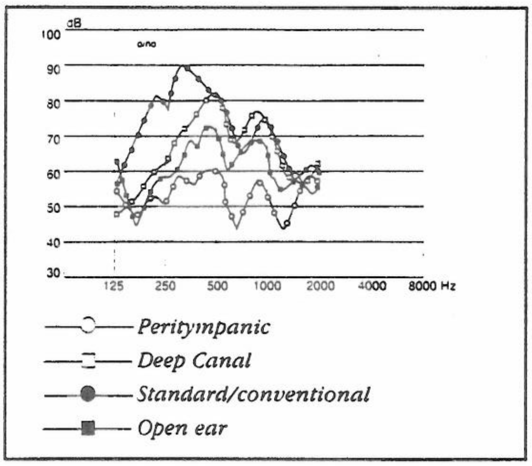


Figure 6.5. Occlusion effects from one's own voice with open ear, a standard earmould, a deep inserted mould and a peritympanic mould (2mm from the nearest point on the eardrum). From [Garcia, 1994].

This was addressed already in 1941 of Békésy [Tonndorf, 1972] and again in 1953 by Zwislocki [1953] but without any great notice. Not until the past 10 years the usefulness of deeply inserted earmoulds has been a subject of interest. Furthermore, it is clinical experience that persons who have tried a deep fitted earplug, claim that their own voice sounds more natural than with a standard earplug, [Nielsen, 1997].

This leads to the conclusion that the occlusion effect by one's own voice is created mainly by vibrations in the soft part in the ear canal and not the bony part. But the bony ear canal might contribute anyway because according to Békésy [Tonndorf, 1972], there are three explanations of why the occlusion effect by bone conduction gets infinitesimal with deep inserted earplugs:

1) - the uncovered surface of the ear canal gets smaller, hence radiation source gets smaller.
2) - the resonance in the occluded ear canal shifts towards higher frequencies when the occluded ear canal gets shorter.
3) - the eardrum cannot move very well and hinders vibration transmission in the middle ear.

Ad. 1)
It is true that when part of the ear canal is covered by the earplug the active area gets smaller, but occluded ear canal volume will also be smaller and it implies an increased sound pressure, so the conclusion is not evident. However, the effect can be estimated with a simple model. The occluded ear canal is regarded as a rigid cavity in parallel with the

eardrum and middle ear, which is modeled with a modified Zwislocki network. A standard earmould reaches about 8 mm into the ear canal and the cavity has the dimensions: length = 23.5-8 mm and diameter = 7.1 mm. A very deep insertion could be an additional 15 mm, leaving a very small volume, alias a very high impedance. The sound source can be modeled as a simple volume velocity generator symbolizing the vibrations of the soft ear canal walls. Presuming that the volume velocity of the ear canal walls decreases linearly with the insertion depth with the gain factor = 1-(insertion depth/length of open ear canal) then Békésy's statement is true: the sound pressure decreases with deeper insertion (alias smaller cavity).

Ad. 2)
The length resonance frequency of the ear canal increases when the occluded ear canal becomes shorter. The resonance frequency of a 23.5 mm long tube is theoretically $f = c/4l$ = 3.6 kHz and a (23.5 - 8) =15.5 mm tube has a resonance at 5.5 kHz. Tonndorf [1972] measured the occlusion effect with varying lengths of open and occluded ear canals and his results indicated that the shift in resonance frequency affects the mid-frequency range but it cannot explain the change in occlusion effect at low frequencies.

Ad. 3)
If the occlusion effect is caused by eardrum movements alone and the eardrum and middle ear had a high impedance, then the occlusion effect should still be present for a deeply inserted earmould. But the eardrum and middle ear have a lower impedance than the occluded ear canal and the mechanical properties will therefore be affected by a change in the load impedance.

When the air pressure in the canal changes, the eardrum get stiffer and the middle ear bones will not move so easily, [Huizing, 1960]. Huizing states that making the eardrum stiffer did *not* change the sound pressure in front of the eardrum. Tonndorf [1972] made an interesting point. He measured the sound pressure in the ear canal produced by bone conduction on cats. Removal of the eardrum did not change the sound pressure significantly at $f < 1$ kHz. In other words, both Huizing and Tonndorf's experiments suggest that the sound pressure in the ear canal is *not* generated by vibrations of the eardrum stimulated from the inside.

The conclusion is that the body conducted sound in the ear canal must mainly be created by vibrations of the ear canal walls.

6.5 THE OUTER EAR COMPONENT

As stated in the latter sections, the prevalent hypothesis is that body conducted sound radiates into the ear canal via vibrations of the cartilage tissue. This theory has in the earlier days been argued by Békésy and Ayres and Morton [in Hood, 1962]. Békésy meant that the occlusion effect was due to periodic contraction of the ear canal walls. Ayres and Morton added to this theory, that the reason, why the occlusion effect is large at low frequencies and decreases with frequencies towards 2 kHz, is that the acoustic impedance in the occluded ear is different from that in the open ear.

6.5.1 The resting jaw

The presence of the jaw alters the vibrations in the cartilaginous part of the ear canal, because the jaw moves relative to the skull. The jaw is not rigidly coupled to the skull - otherwise we would not be able to move the jaw. It means that the jaw does not necessarily vibrate with the same amplitude and phase as the skull does.

Békésy [in Hood, 1962] stated that the inertia of the lower jaw was the cause to the occlusion effect. He presumed that the jaw has a large enough inertia to stay in rest relative to the skull. Then the ear canal volume will be compressed because the lower jaw's condyle (the 'top' of the lower jaw) is near the ear canal wall. This compression would generate a variation of sound pressure in the occluded ear canal.

Békésy's hypothesis conflicts with the findings of Huizing [1960]. The ear canal sound pressure was measured on 1 subject whose lower jaw on one side had been surgically removed. The jaw in the other side was normal. The skull was stimulated with a bone conductor and the sound pressure in the open ear canal was measured. Huizing found that the absence of the lower jaw created a larger sound pressure in the open ear canal at least up to 2 kHz. At 125 Hz the sound pressure was 20 dB higher and at 250-750 Hz it was 8-10 dB higher. If Huizing's measurements are correct, then the production of sound pressure in the ear canal cannot only be due to compression of the cartilage created by the jaw. The sound pressure in the open ear canal would be greater if the lower jaw was not there, thus the inertia of the lower jaw dampens the vibrations of the cartilage.

6.5.2 Vibrations of the jaw

Zwislocki [1953] investigated the influence of a vibrating jaw by mean of hearing thresholds detections. He compared the hearing threshold obtained with an acoustical shielded bone conductor placed on the right side of the jaw 2.5 cm below the right ear and on the right mastoid respectively. The left ear was covered with a headphone with a doughnut cushion and the threshold of the bone conducted stimulus tone was determinated at 250 Hz, 500 Hz, 1 kHz, 2 kHz and 4 kHz. He found that the threshold was lower (the bone conduction sensitivity greater) for the mastoid placement of the bone conductor. This applied to all frequencies except at 4 kHz. When a light earplug was inserted into the cartilaginous part of the left ear canal, the threshold was 8-10 dB lower for jaw placement than for mastoid placement at 250-500 Hz and 4 kHz.

In the case with the doughnut cushion, the occlusion effect is significantly smaller than the occlusion effect obtained with an earplug. If the doughnut cushion was large enough, the occlusion effect would vanish, [Revit, 1992]. The middle ear bones are excited more when the bone conductor is placed on the mastoid because the skull is stimulated directly. The ear canal component is small and the middle ear component is the greatest contributor to the sensation of sound. In the second case, the 'sound pressure' will be increased in the left occluded ear and the outer ear component is dominant. The outer ear is stimulated more when the bone conductor is placed on the jaw, so Zwislocki's results are not surprising. It implies that vibrations of the jaw excites the walls in the ear canal.

This leads to the conclusion that vibrations of the jaw relative to the skull creates an excess sound pressure in the ear canal. The explanation is that the skull excites the cartilage in the ear canal whose vibrations alone as well as the relative movements to the bony part of the ear canal can create a sound pressure.

Presuming that one's own voice sets the mandible into vibrations, then the presence of the lower jaw does influence the sound pressure in the ear canal and the sound pressure created hereby will be greater than the sound pressure created by oscillations of the skull. Franke et al. [1952] concludes from the amplitudes that the lower jaw vibrates similar to a simple oscillator with resonance frequency 110-180 Hz.

Studies of the phase lag between skull and lower jaw vibrations indicated that the mechanism was not so simple. This mechanism seemed to be some kind of damping for example because of the connecting tissue and muscles between jaw and neck.

Far below the resonance frequency of the simple oscillator, the jaw and skull vibrate together and far above the resonance frequency, the lower jaw stays nearly at rest. The lower jaw stays relatively, to the ear canal, at rest because the amplitude of the skull vibrations are twice as large as the amplitude of the lower jaw vibrations at 500 Hz. This observation confirms to some extent Békésy's theory about the lower jaw nearly staying at rest. Békésy talked about the lower jaw in relation to the occlusion effect, thus he might have applied the theory down to as low as 100 Hz.

6.5.3 Mechanism of Jaw movements

The lower jaw can be moved up and down, sideways and rotated. The upper joint of the lower jaw is called the temporomandibular joint and it is very complex. In fact the joint consist of two joints that act as one. The temporomandibular joins the mandibular to the condyle (the top of the lower jaw) of the temporal bone, see **Figure 6.6**.

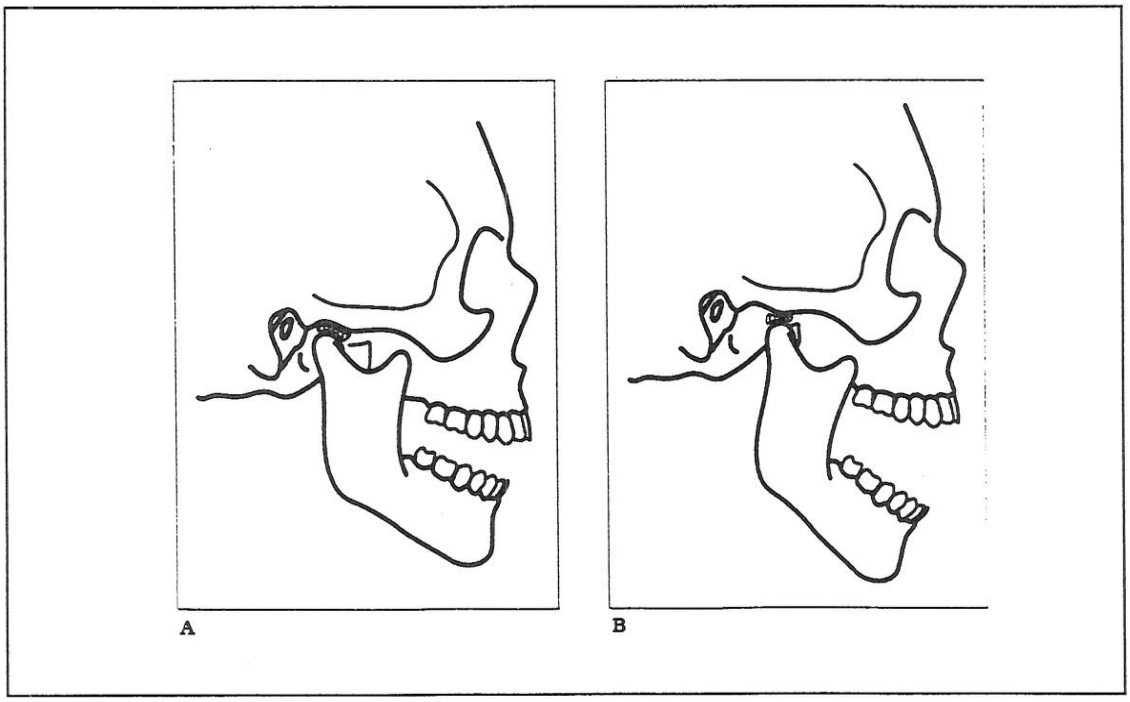


Figure 6.6 The jaw and the temporomandibular joint. A: Slightly opening of the jaw. B: Wider opening causes the condyle to move forward. Based on drawings from [Oliveira, 1995].

Between the mandibular and the condyle is the articular disc, which is made of collagen tissue. When the jaw is lowered just a few mm, the condyle and the disc rotates. The disc should be moving forward but because of the elasticity, it is pulled backwards. These two opposite stresses makes it rotate. Opening the jaw wider still creates a rotation but also produces a gliding movement between the disc and the temporal bone. Both the disc and the condyle moves forward, see **Figure 6.6**. This forward movement can be felt by placing a finger in front of tragus on the condyle while moving the jaw. Otoscopy of the ear canal confirms that the soft ear canal walls also moves inwards when the jaw is lowered [Oliveira, 1995].

Edwards and Harris [1990] measured the three-dimensional movement of the condyle of the jaw during speech. They found that in general, the rotation creates the largest displacement of the condyle and the forward movement is much larger than the vertical movement. The size of displacement differs from person to person. Out of three subjects the smallest forward displacement was 2.5 mm and the largest 13.4 mm for the vowel /aaa/.

The forward gliding of the condyle pulls the anterior soft ear canal wall forward. Oliveira [1995] measured the physical changes of the ear canal caused by jaw movements. Ear impressions between the first and second bend were taken on 6 hearing aid users with the jaw lowered. Afterwards the ear canal impression geometry was measured. The experiment showed that the height (vertical dimension) of the ear canal does practically not change, when the jaw is lowered. The width (anterior/inferior dimensions) changes significantly. When the jaw is lowered 30 mm the width increases up to 25%. For some subjects the width increase was very different in the left and right ear. This is caused by asymmetric jaw movements. The average width increase for 18 ears shows a maximum of 10%. The width increase is greatest in the anterior direction, whereas the inferior wall nearly does not move. This change occurs both for lowering the jaw and protrusion of the jaw [Willigen, 1976].

Franke et al. [1952] placed a bone conductor on vertex (the top of the skull) producing a vertical oscillation of the skull. The ear canal was closed with a rubber earplug and the sound pressure in the ear canal was measured with open and closed mouth, see **Figure 6.7**. Opening the mouth caused only an increase of 2 dB in the open ear canal whereas the sound pressure increased 8 dB in the occluded ear at 200 Hz.

Huizing [1960] measured the ear canal sound pressure on 8 subjects while they clenched their jaws. The stimulus used was a bone conductor placed on the skull (the paper does not say exactly where on the skull). It turned out that the real ear sound pressure increased and decreased at different frequencies and that the pattern was different from person to person. The repeatability was very low and the sound pressure depended on how strongly the jaw was clenched. For example; for one subject the sound pressure decreased with 5 dB at 250 Hz and increased with 5 dB at 500 Hz when the jaw was clenched. **Figure 6.7** illustrates these results.

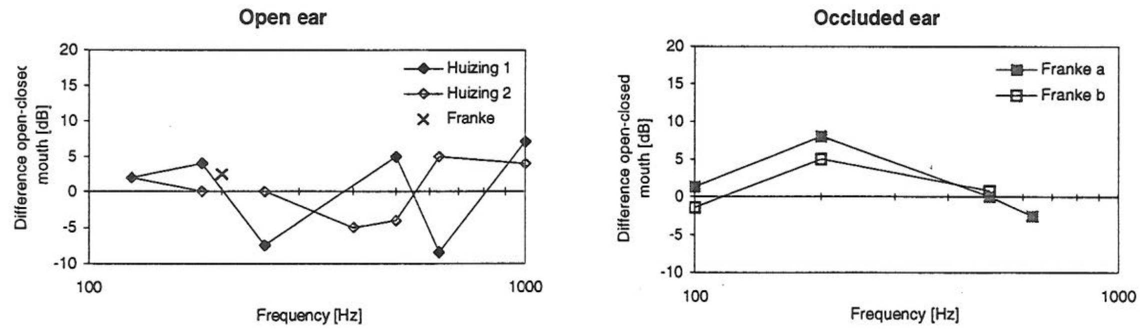


Figure 6.7. Difference in ear canal sound pressure with the jaw open and closed. Left: open ear canal. Right: occluded ear canal. Bone conductor stimulation on the skull. Huizing [1960] 1: clenching the jaw. Huizing 2: wide open jaw. Franke et al [1952]: subject a and b.

The jaw is lowered during speech. It is especially necessary to open the mouth up wide to make the sound /aaa/. During normal speech the jaw does not open very much over time. Borghese et al. [1997] measured the jaw opening during speech with a TV-camera technique. Six subjects had to repeat "/to/, /til/, /tru′/ and /chi/ " (Italian) in random order. The jaw opening was only a few mm. The paper does not print any average data, but a single sequence shows a maximal jaw opening of 6 mm. Roughly taken, the jaw can be lowered at least 45 mm, so during speech, the jaw is far from lowered to the maximum possible.

In the word "/to/ /til/ /tru′/ /chi/ " the shortest time between lowering the jaw and lifting again is 0.85 sec. That is from /to/ to /til/. It takes the same amount of time to lowering the jaw as to lift it. Hence, it takes 0.4 sec. to lower it, see **Figure 6.8**. This gives a frequency of 2.5 Hz - a frequency much lower than the speech range. Conclusively, the jaw movement does create audible vibrations of the ear canal wall. The effect of jaw movements is that a leakage between the earmould and the ear canal wall may occur. Lowering the jaw deforms the ear canal wall during the whole period of the produced sound.

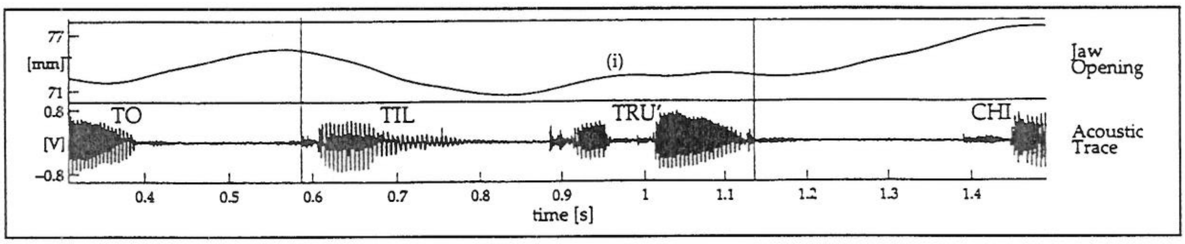


Figure 6.8. Lowering the jaw compared to the generated sound pressure in front of the subject. The jaw opening takes much longer than one period of a vowel. From: [Borghese et al., 1997].

7. HYPOTHESES OF THE OCCLUSION MECHANISM

The previous chapter indicated that one's own voice makes the skull vibrate in the vertical direction. Sound is generated in the ear canal because the ear canal walls vibrate. An important observation was illustrated in chapter 1, even a very small vibration amplitude creates sound pressure levels higher than that of the normal speech level, when the ear is closed. For example, a piston placed in one end of a closed tube only has to vibrate with an amplitude of $4 \cdot 10^{-6}$ mm in order to generate 60 dB SPL at 1000 Hz. Thus, it is indeed possible that the ear canal tissue creates sound by it's vibrations and that an earmould inserted into the cartilaginous ear canal contributes because the mould and the cartilage move relative to each other.

7.1 MEASUREMENT OF TISSUE VIBRATIONS

One's own voice is transmitted through the air and through the body to the ear canal. The body conduction component arises because vibrations from the vocal cords and vibrating air in the vocal tract and oral cavities excite the skull bone and softer tissue. These vibrations are transmitted to the ear canal where it is believed that especially the soft wall of the ear canal vibrates. The vibrations of the cartilage in the ear canal have been measured with a microphone built into the side of an individually made earmould, see appendix F for details.

7.1.1 Measurement method

A microphone, Lectret Type 2141, was built into the shell of a custom made earmould. The microphone was enclosed with a rubber membrane that attenuated 22-36 dB depending on frequency. The earmould was made in a standard design and inserted into the cartilaginous part of the ear canal. The rubber membrane had direct contact with the anterior cartilaginous ear canal wall. This rubber enclosure had a volume of 56 mm^3 and even very small movements of the rubber membrane changed the volume of the enclosure, which create a sound pressure inside the enclosure. The arrangement with a microphone enclosed in a rubber cavity detects only practical vibrations of the ear canal wall and the arrangement will in this section be called a '*vibration meter*' for the sake of convenience.

An electret microphone Knowles EM3356 was built into the tip of the earmould. This microphone was used to record the sound pressure in the occluded ear canal and will be termed '*probe microphone*'. **Figure 7.1** illustrates the earmould arrangement. The sound pressure in the occluded ear canal and in the rubber enclosure was measured simultaneously, while the subject vocalized. Only one subject was tested.

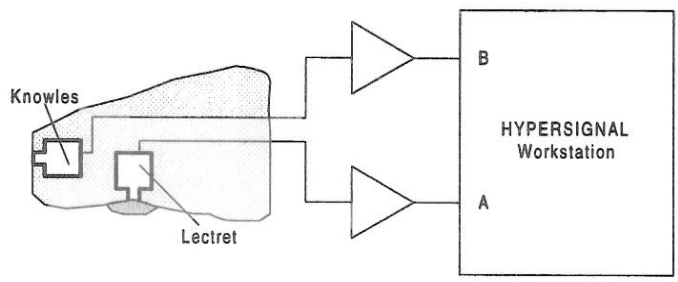


Figure 7.1. The individually made earmould.

7.1.2 Results

The calibrated responses from the Lectret 2141 and EM3356 microphone are drawn in **Figure 7.2** where continuos speech and single words are shown. Except from the lowest frequencies below 300-500 Hz depending on the signal, the spectra obtained with Lectret 2141 (vibration meter) and the EM3356 have nearly the same shape. Even the peaks and valleys at 800-2000 Hz are produced in the signal from the vibration meter. It is especially clear for the words /ice/ and /pil/. At lower frequencies, the occluded ear pressure is lower than the rubber enclosure pressure. The reason is most likely that there is a leakage between the earmould and the ear canal wall, whereas the rubber enclosure stays closed.

Body conducted sound is also generated by chewing. The sound pressures produced when the subject chewed gum is printed in **Figure 7.3**. The characteristics in the speech spectrum are nearly the same for chewing gum. The curves match at the mid-frequencies and the vibration meter signal is larger than the probe microphone signal at lower frequencies.

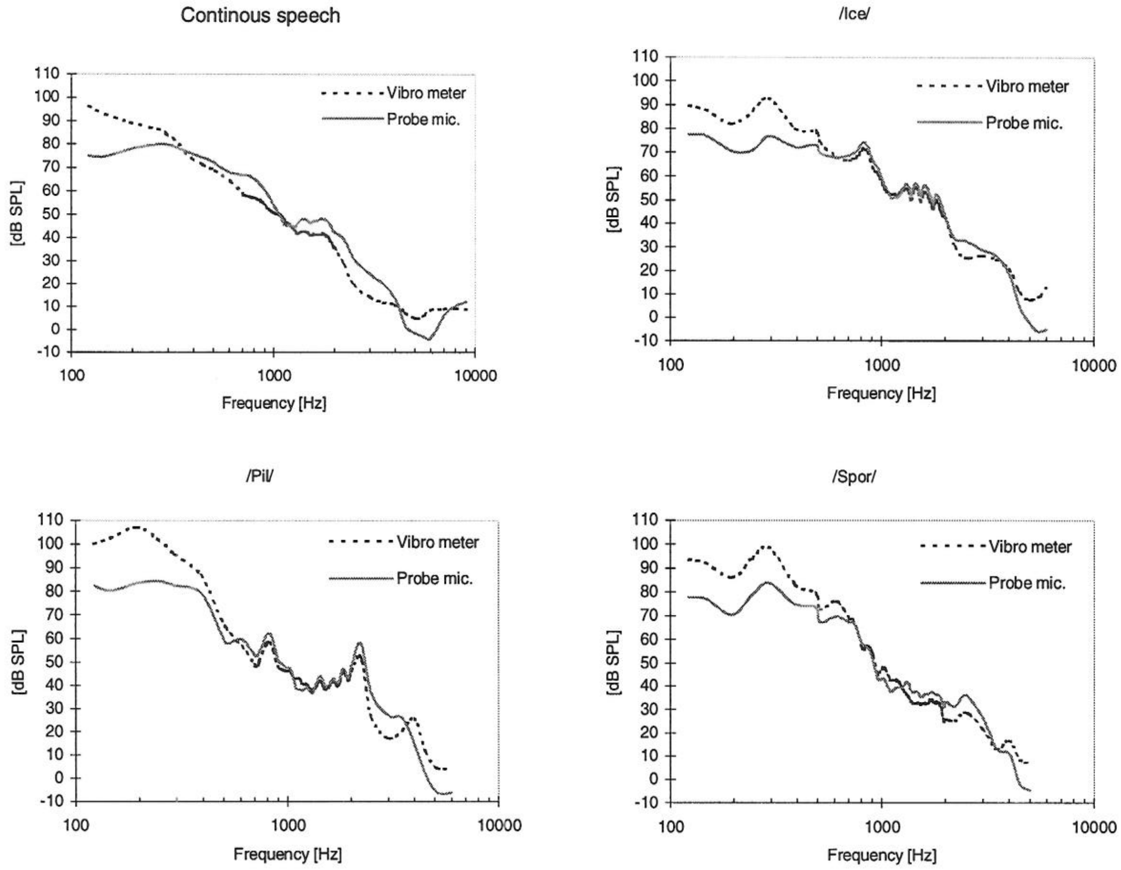


Figure 7.2. Calibrated sound pressure in the occluded ear canal. Measured with the vibration meter and the probe microphone. Continuos speech and short words. 1 subject, subjects own voice. The International Phonetic Alphabet: 'ice' = /aIS/, 'spor' = /Sbo')/, 'pil' = /pi'l/.

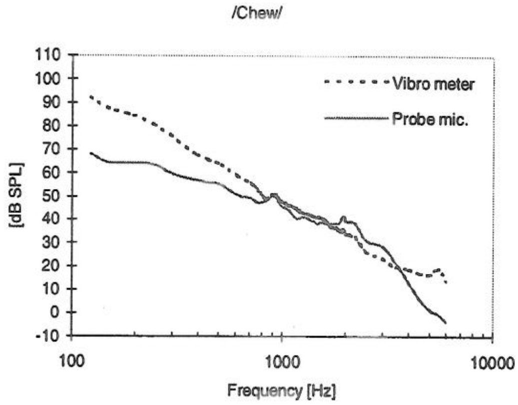


Figure 7.3. Calibrated sound pressure in the occluded ear canal. Measured with the vibration meter and the probe microphone. Chewing gum. 1 subject.

The difference between the probe microphone and the vibration meter response is calculated in **Figure 7.4** for each word. The difference seems to be the same for each of the single words. The difference is clearly smaller for continuous speech than for single words below 1 kHz. It is presumed that the differences here are due to leakage. The ear canal changes form when the mouth is opened, so a leakage between the earmould and the ear canal wall will likely also change it's form. This might be the explanation for the deviation between continuos speech and single words in **Figure 7.4** as continuous speech is averaged over a much longer time than single words. It is necessary to make more measurements with a lot more variety of speech sounds, in order to conclude whether or not the differences are consistent with the speech signal.

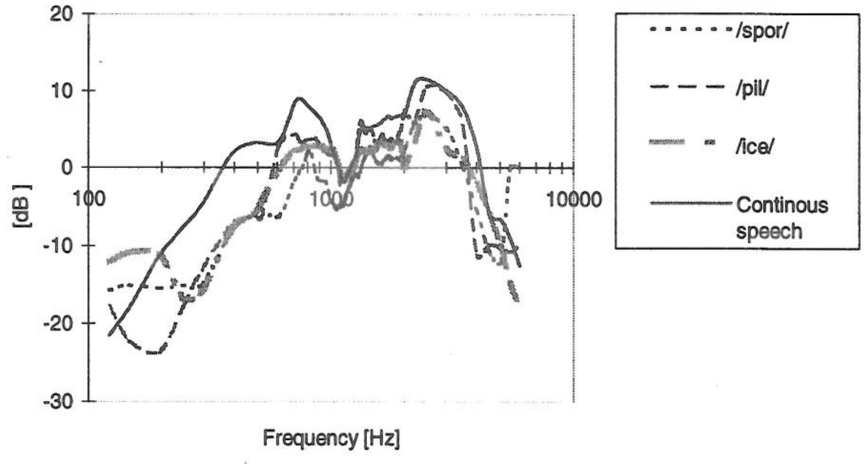


Figure 7.4 Difference between the sound pressure in the ear canal and in the rubber enclosure.

The pressure measured with the vibration meter is about the same level at the mid-frequencies as the pressure measured with the probe microphone. However, this could be a coincidence. If a small cavity was cut in the earmould under the rubber piece and the microphone was placed in the end of the cavity, then the volume, V, would be larger but the surface that touched the ear canal wall will be the same. The result is that the pressure decreases. Hence, the pressure is smaller but the active area that generates sound is the same. Consequently, the sound pressure measured with the vibrometer must be normalized to the sound pressure measured with the probe microphone. In order to do that it is necessary to make the assumption that the whole surface of the ear canal vibrates with the same volume velocity all over. The pressure in a cavity is given by:

$$p(\omega) = \frac{1}{j\omega V / \left(\rho c^2\right)} q(\omega) gain; \tag{7.1}$$

where;
q = volume velocity in the cavity
p = sound pressure in the cavity
V = cavity volume
ρ = air density
c = sound velocity
ω = angular frequency
$gain$ = ('free' surface of the ear canal wall / whole surface of the ear canal wall). The free surface is the surface that is not covered by the earmould.

The ratio of pressure in the rubber cavity and the occluded ear should then be:

$$\frac{p_e}{p_m} = \frac{q_e(\omega)}{q_m(\omega)} \frac{V_m}{V_e} = \frac{A_{occluded\ e} v_e(\omega)}{A_m v_m(\omega)} \frac{V_m}{V_e} \tag{7.2}$$

where;
A = surface area
v = point velocity

If the point velocities of the ear canal tissue is $v_e = v_r$ then:

$$\frac{p_e}{p_m} = \frac{A_{occluded\ e}}{A_m} \frac{V_m}{V_e} = \frac{2\pi r l_{occluded,e}}{b l_m} \frac{b l d_m}{\pi r^2 l_{occluded,e}} = \frac{2 d_m}{r} \tag{7.3}$$

where;
r = radius of ear canal
b, l, d = width, length and depth of rubber cavity
subscript m, e = rubber membrane, ear canal

The ratio then becomes p_e/p_r = (2·2mm/3.5mm) = 1.1, so the pressure in the rubber cavity and the ear canal should be about the same level. This agrees well with the measurements.

The results from the experiment indicate that the earmould and the ear canal wall moves relative to each other. Just as important, the amplitude of these vibrations is great enough to create a sound pressure in the rubber cavity equal to the sound pressure measured in the occluded ear canal, but further studies are needed to find the exact ratio between the pressure measured with a microphone in the ear canal and with a vibration meter on the side. The ratio might depend on the location of the accelerometer, the fitting of the earmould and/or natural leakage between the earmould and the ear canal wall. The ratio will also depend on the insertion depth of the earmould.

7.2 HYPOTHESES OF THE OCCLUSION EFFECT MECHANISM

Now, that it has been indicated that the tissue and the earmould actually do vibrate relative to each other, some hypotheses of the occlusion mechanism will be discussed. But first the effect of the insertion depth will have to be explained.

7.2.1 The occlusion effect depends on the insertion depth

The 'occlusion effect' in a tube excited with a piston in one end becomes greater when the piston is moved further into the tube, so that the volume decreases. The theoretical occlusion effect in a tube as a function of insertion depth of the piston is plotted in **Figure 7.5.**

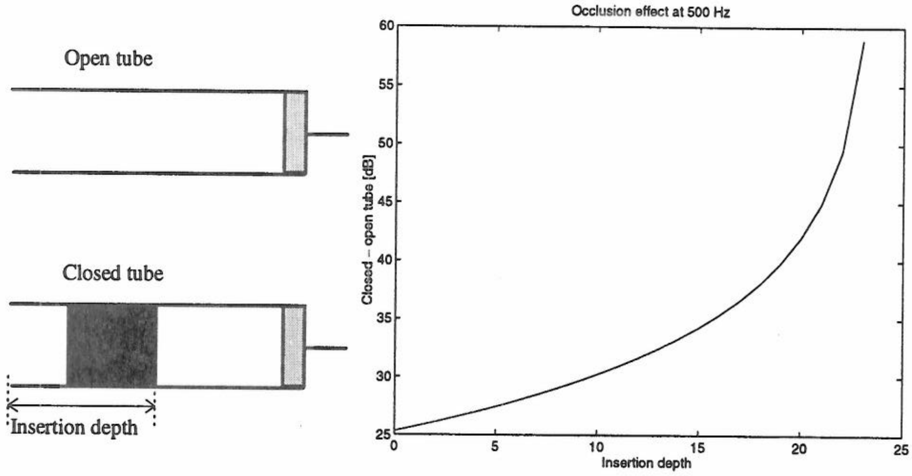


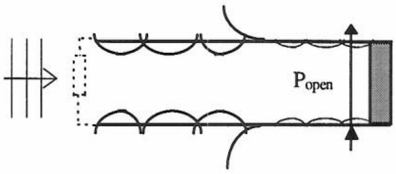
Figure 7.5 Occlusion effect in a rigid tube as a function of insertion depth. Tube length = 23.5 mm, diameter = 7.1 mm. Stimulus: piston in one end of the tube volume velocity = $1 \cdot 10^{-6}$ mm^3/s.

Theoretically, the occlusion effect increases with insertion depth, but in the real ear situation, the opposite happens. The occlusion effect becomes, to some extent, smaller with deeper inserted earmoulds. An important point is that it is possible to fit the earmould so that the occluded ear pressure is the same or even lower than the open ear pressure. But it requires that the mould is inserted into the bony part of the ear canal and that it is fitted very tightly to the bony ear canal. An example of the effect of the insertion depth has already been shown in **Figure 6.5.** The figure also illustrates that even when the occluded ear pressure is lower than the open ear pressure, it is still quite large, 60 dB SPL. A pressure on 60 dB SPL cannot alone be created from sound traveling through leakages or through the earmould materials, thus there must be some sound generated in the occluded ear canal, even when the mould is inserted deep in the bony part.

The occlusion effect mechanism is not quite so simple as indicated in **Figure 7.5**. The soft tissue in the ear canal is excited by vibration energy created from the vocal cords and according to Békésy [1949], the skull is also set into vibration by one's own voice. It is therefore more correct to assume that the sound pressure in the occluded ear canal is a result of the interactions between the bone, the cartilage and the earmould. Several hypotheses of how the occlusion mechanism works can be set up. The influence of the inner ear and the eardrum does not seem to be the primary factors as discussed in chapter 6.

Therefore, the following will only look at the mechanism in the outer ear. Three possible hypotheses are discussed, starting with the open ear.

7.2.2 The open ear



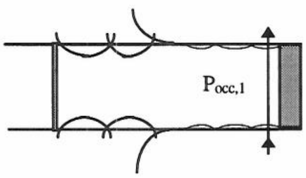
The greatest contributor to the sound pressure in the open ear is the air borne sound, which is radiated from the mouth and nose and transmitted around the head while speaking. But one's own voice is also transmitted to the ear canal through the body. One's own voice is believed to set the skull into vertical vibrations, see chapter 6. The bony ear canal is a part of the skull and will therefore also vibrate vertically. It is presumed that the very thin skin layer is able to vibrate a bit.

The cartilage is set into vibrations partly directly from energy transmitted through the body and partly because the bony ear canal excites the cartilage. The cartilage is elastic, so it will move in another way than the bone. The vibration patterns of the cartilage have not been studied, but it must be assumed that the surface of the cartilage neither moves as a rigid wall nor as an extremely soft material. For example, a great area of the cartilage on the anterior wall is believed to vibrate nearly with the same phase and amplitude but not necessarily with the same phase and amplitude as the posterior wall.

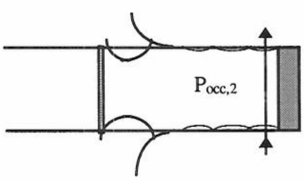
A sound pressure arises in the ear canal because the open ear is acoustically loaded with a radiation impedance (drawn with dashed lines). The sound wave travels towards the point of less loading, and since the radiation impedance (the mass of the air that the wave has to move) is a lot smaller than the acoustic impedance of the eardrum and middle ear, the wave finds it way out of the ear canal. Now, let's look at what could happen when the ear is occluded.

7.2.3 Hypothesis 1: Vibration of the cartilage

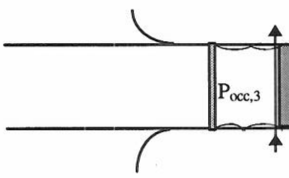
Standard insertion $P_{occ,1} > P_{open}$



Deeper insertion $P_{occ,2} < P_{occ,1}$



Deep insertion $P_{occ,3} << P_{occ,2}$

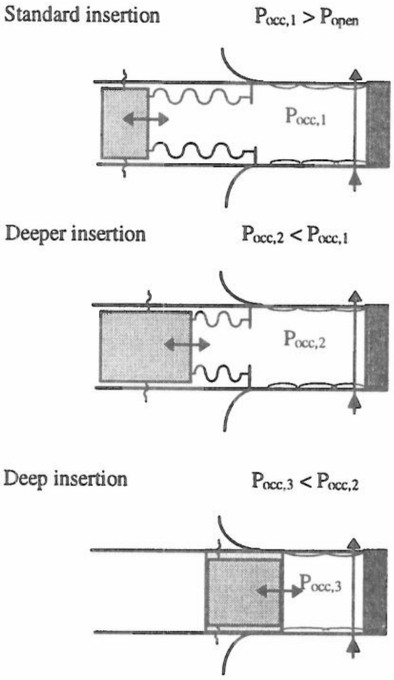


A simple hypothesis is to neglect the mechanical properties of the earmould and regard it as a rigid wall. This approach was used by Schroeter and Poesselt [1986] (see chapter 3). The sound pressure in the occluded ear is entirely created by vibrations of the cartilage, neglecting sound sources in the middle ear.

When the earmould is inserted deeper, the sound pressure drops because the surface of the active cartilage gets smaller. Thus, the volume velocity generated by the cartilage decreases.

When the earmould is inserted into the bony ear canal, the sound pressure drops to nearly zero, because the bony wall vibrates a lot less than the cartilage. It will be shown later that this theory is too simple because the mass of the earmould is important.

7.2.4 Hypothesis 2: Earmould movements



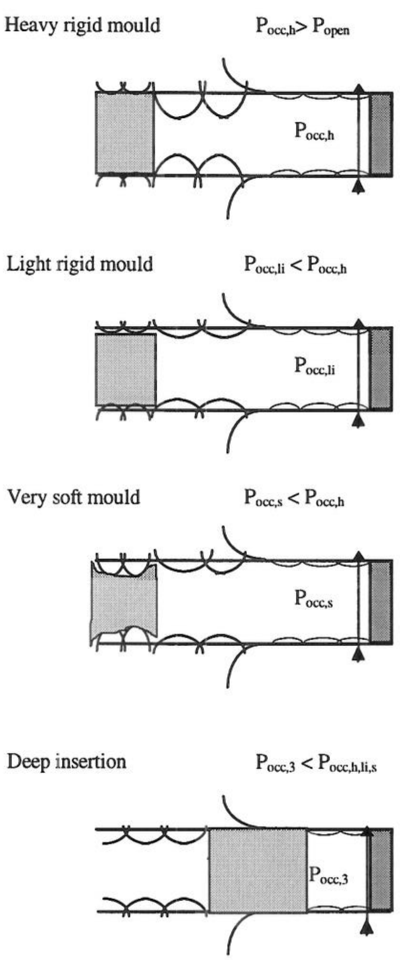
The earmould is in practice never fitted 100% mechanically tight to the ear canal wall and it will therefore be able to move. In a simple hypothesis, the earmould is supposed to move along the length axis of the ear canal. Hereby, the occluded volume change and a sound pressure is generated. The earmould is assumed to be attached to the cavity wall with elastic strings (the cartilage). The compliance of the tissue in the bony ear canal is also shown on the drawing but it is very firm. The bony ear canal moves vertically as a rigid body. The heavier the mould is, the smaller the sound pressure.

With a deeper inserted mould, the sound pressure decreases because the active elastic cartilage becomes smaller and the earmould is not able to move so much.

When the earmould is inserted into the bony ear canal, the active tissue becomes even smaller and the earmould will move less. There is no free cartilage between the mould and the bone to set the mould into vibrations. If the earmould is fitted very tightly - as shown on the bottom figure with light gray signature - the mould cannot move relative to the bony ear canal but must follow the movements of the bony ear canal and theoretically no sound is generated in the occluded ear canal. If the mould was fitted very tightly in the soft ear canal, the bony part would still move relative to the earmould and a sound pressure would be generated.

If this hypothesis is true, then the pressure in the occluded ear will be greater with a light mould than with a heavy mould, because the ear canal wall does not have to vibrate so much so set the light mould into vibrations as with a more heavy mould.

7.2.5 Hypothesis 3: Distributed vibrations



Hypothesis 3 reminds us a bit of hypothesis 1. The difference is that now that mechanical properties of the earmould are taken into account and the vibration amplitudes of the cartilage depends on the loading on the surface. It is assumed that the surface of both the soft and the bony canal vibrates, though the cartilage vibrates a lot more than the skin in the bony ear canal. The temporal bone moves vertically as a rigid body. Furthermore, it is presumed that the total energy of the volume velocity source is constant, so that the tissue area with the smallest loading will vibrate more than the tissue with a greater loading. Hence, the cartilage covered by the earmould vibrates with a smaller amplitude than the free cartilage.

If the earmould has a considerable mass, the covered cartilage cannot move very much and the vibration amplitude of the free cartilage is great because more of the energy is distributed to the free cartilage.

When the earmould is lighter and perhaps also softer, the loading is smaller and the covered cartilage can move more. Consequently, the vibration amplitude of the free cartilage becomes smaller and the sound pressure is smaller than with a heavier or more rigid mould.

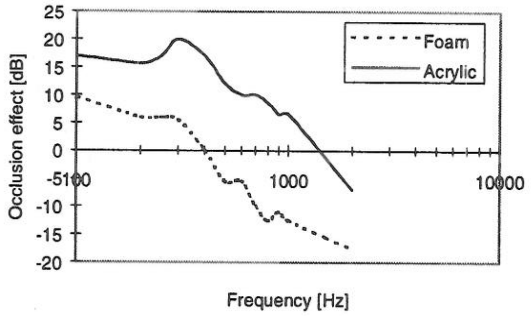
When the earmould is inserted into the bony ear canal, the sound pressure drops because the impedance of the tissue is much greater than the cartilaginous tissue and therefore less able to vibrate. Furthermore, if the deep inserted earmould does not cover the whole outer ear canal, most of the vibration energy will be transmitted to the free cartilage in the outer part. Even when the earmould is tightly fitted, a small sound pressure will still be generated because of the surface vibrations.

In chapter 8 the latter 2 hypotheses will be analyzed theoretically and compared with measurements on real ears. But first, it is useful to look at the observed influence of the mechanical properties of some earmoulds inserted into real ears.

7.2.6 A rigid mould contra a soft mould

The influence of the earmould will be specified by comparing two commonly used earmoulds. A standard acrylic hearing aid mould and an 'over the counter' foam earplug.

Data from the literature indicates that the occlusion effect is different for an acrylic mould and a foam plug, see **Figure 7.6**. The data are obtained from two master thesis projects in relation to the present Ph.D.-work. The acrylic moulds were standard hearing aid moulds but individually moulded to each of the 18 subjects. The occlusion effect was measured with a probe microphone connected to a thin tube that was put through a drilled hole in the earmould. Vaseline was spread on the moulds to make them fit better to the ears and to minimize leakage. The foam plugs were standard plugs used for insertion ear phones and the tube of the probe microphone was put through the sound tube in the middle of the earplug. In this experiment 10 subjects participated. Thus, in both experiments an attempt was made to minimize any leakage created by the insertion of the probe tube.



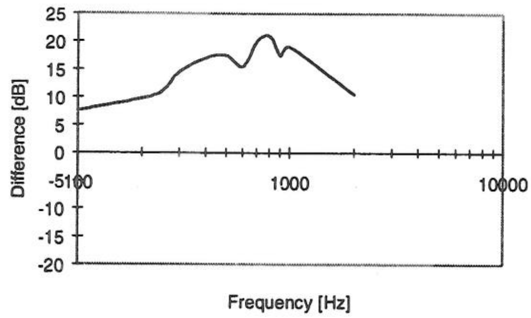


Figure 7.6. Left: Occlusion effect with foam plugs and acrylic moulds. One's own voice. Foam: 20 ears (10 subjects), [Bremmelgaard, 1997]. Acrylic: 36 ears (18 subjects) [Thorup, 1996]. Right: Difference in occlusion effect between acrylic mould and foam plug.

The occlusion effect for acrylic moulds is definitely greater than with foam plugs. The greatest difference is 20 dB at 800 Hz. If these results are also true when the earplugs are fitted to the *same* person, then it is of great interest, because the occlusion effect will be significantly reduced - but not necessarily eliminated - with foam plugs. There could be several explanations or rather a combination of explanations:

A) *Different insertion depths*

The foam plugs were inserted deeper into the ear canal than the acrylic plugs, and as shown earlier the occlusion effect decreases with deeper insertion.

The influence of the insertion depth can be estimated with the same simple model as described in chapter 1. The sound pressure in the occluded ear canal is modeled with a sound source inside a closed cavity. The sound pressure in the cavity depends on the insertion depth. The volume velocity of the sound source is smaller when the insertion depth is larger. When the earmould is inserted deeper, it hinders more of the active ear canal wall to move.

The individual earmould insertion depths were not measured but as they are manufactured as standard moulds, it is reasonable to assume that the insertion depth is about 8 mm on average. The foam plugs were inserted approximately 13 mm[1]. If the insertion depth of the foam plug is corrected to 8 mm by the mean of a mathematical model for occlusion (see chapter 7), then the occlusion effect will not

[1] Bremmelgaard [1997] gives the average insertion depth to 8.7 mm. This is measured by looking at the deformation of the foam just after the foam was pulled out of the ear. The insertion depth was measured from the point where the foam had a distinct compression. This method underestimated the insertion depth by approximately 4 mm because the most distinct compression of the foam is not at the entrance. This argument has not been scientifically proved, but the compression of foam plugs was studied as a single subject.

increase by more than 2 dB. Still there seems to be a great difference in the occlusion effect between foam and acrylic plugs. This estimation is only true if the active vibration area decreases proportionally with the insertion depth in the ear canal, so there is still some uncertainty about the influence of the different insertion depths.

B) *Measurement technique*
The two studies used the same principle of measurement method but the set up and data analysis were very different and could have caused some dB difference. Another cause of the difference is that the experiments were performed with different subjects. For example, two subjects in the acrylic experiments did not have any occlusion effect at all i.e. the occluded ear pressure was the same as the open ear pressure.

C) *Reflection coefficients*
The magnitude of the occlusion effect is determined by the acoustic loading of the ear canal and if the earmould material is absorbent, the occlusion effect is reduced. Since the foam plug is made of a porous material with some coating on the ends, it might dampen the sound pressure in the ear canal.

D) *Leakage*
The foam plug is elastic and it fits more tightly into the ear canal than an acrylic mould, thus there will be a greater leakage between the mould and the ear canal wall with an acrylic mould. A small leakage creates excessive sound in the occluded ear because of resonance at certain frequencies. The 'bump' on the acrylic curve could be caused by a tube shaped leakage. However, a leakage would not give a positive difference between an acrylic and a foam plug at the very low frequencies, for example below 200 Hz.

E) *Influence of the earmould mechanical properties*
The mass and the elasticity of the earmould will influence the sound pressure in the occluded ear canal if hypothesis 2 or 3 is true. According to hypothesis 2, the earmould vibrates. The acrylic mould is mechanically rigid at least at low frequencies and it will therefore not be able to completely follow the movements of the cartilage in the ear canal. This is equivalent to the earmould generating sound as a rigid piston. In hypothesis 3, the cartilage covered by the earmould will move less with the acrylic mould and more with the foam plug. The free cartilage will therefore move with the greatest amplitude with the acrylic mould. In both hypothesis, it is likely to believe that an acrylic mould causes excess sound pressure in the ear canal compared to a foam plug.

It would be more easy to explain the great differences if the occlusion effects with a foam plug and an acrylic mould was measured on the same persons. Explanation B) would be eliminated and A) would be insignificant. No literature was found on soft and hard earmoulds measured on the same persons. Therefore an experiment was set up with 4 subjects fitted with a foam plug and an acrylic mould. The next chapter gives some examples of how the mechanical properties of the earmould can alter the occlusion effect.

8. INFLUENCE OF THE EARMOULD

The effect of the mass and the elasticity of the earmould is analyzed in this chapter. Experimental results have shown that the sound pressure in the occluded ear is greater with a 5 g earmould than with a 1 g earmould and that an acrylic mould creates higher sound pressure than a foam plug. The hypotheses proposed in chapter 7 are analyzed. The analysis strongly suggest hypothesis 3, where the energy of the source volume velocity is distributed over the tissue surface.

8.1 MEASUREMENT OF FOAM AND ACRYLIC MOULDS

The purpose of this experiment was to determine the occluded ear levels with foam plugs and acrylic plugs measured on the same person. Instead of determining the occlusion effect it is sufficient to know the occluded real ear sound pressures in order to compare the performance of foam plugs and acrylic moulds as follows:

$$\frac{oe_{acrylic}}{oe_{foam}} = \frac{\left(p_{acrylic,occluded}/p_{open}\right)}{\left(p_{foam,occluded}/p_{open}\right)} = \frac{p_{acrylic,occlud}}{p_{foam,occluded}} \qquad (8.1)$$

8.1.1 Measurement method

The principle of the measurement method was to measure the level in the right and left ear simultaneously. Real ear levels were recorded in the left ear occluded with an acrylic mould and at the same time in the right ear occluded with a foam mould. Afterwards the measurements were repeated with inter-changed moulds. The experiment took place in an anechoic room which can be considered anechoic at the speech frequency range from 200 Hz. The set up is illustrated in **Figure 8.1**. A detailed description of the experiment is given in appendix C.

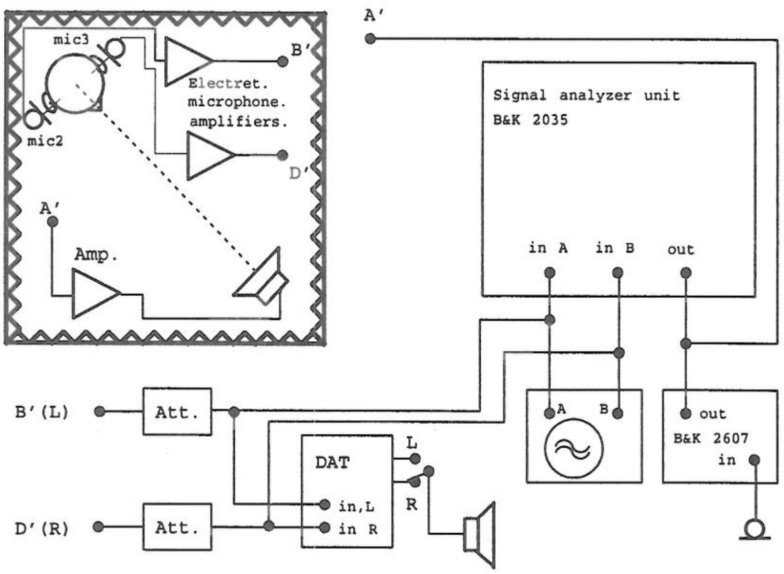


Figure 8.1. Experimental set up. The level in the right and left ear was recorded simultaneously with two probe microphones. The loudspeaker in the anechoic room was used to measure the insertion loss. The loudspeaker and microphone outside the anechoic room was used to communicate with the subject in the anechoic room.

The acrylic earmoulds were made from ear impression of the subjects ears and the foam plugs were standard plugs used for earphones. Four male subjects participated, all with normal hearing and a normal middle ear, except for subject 2, who had a minor abnormality in the tympanogram.

Real ear levels were detected with hearing aid microphones (Knowles EK 3024) attached to a soft tube with an outer diameter of 1.5 mm. The tube was inserted through a drilled hole in the acrylic moulds and through a manufactured plastic tube in the foam plug. This way it was ensured that the probe tube did not create leakage, which was confirmed by measurements. All signal analysis and recordings were made with a Dual channel signal analyzer unit, (Brüel&Kjaer Type 2035) and as a security, all measurements were also recorded onto a DAT-tape.

Simultaneous control of the recordings were done partly by looking at the oscilloscope screen and listening to the output via a loudspeaker.

The loudspeaker in the anechoic room served partly the purpose to communicate with the subject and to generate sound for insertion loss (attenuation) measurements. The subject was placed with the nose pointing towards the center of the loudspeaker. The signal analyzer generated a multi-sine signal which was sent to the loudspeaker. Cable connections between the anechoic room and the control room went through a cross field and thus made it possible to close the door to the anechoic room completely.

The attenuation of the earmoulds when they were inserted into the ear was measured with the subject sitting in front of the loudspeaker, the nose pointing towards the center of the loudspeaker membrane while one ear was occluded and the other open. The difference between the occluded and the open ear is called 'insertion loss'.

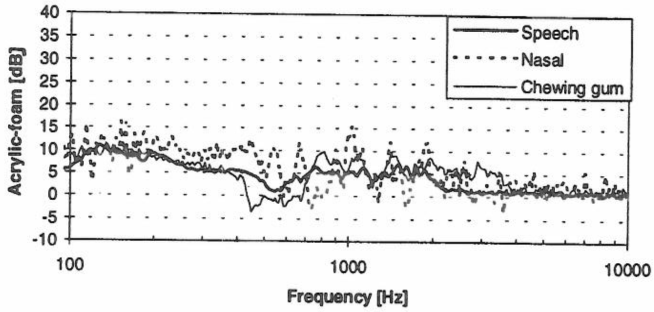
8.1.2 Sound pressure level in occluded ear levels

The occluded ear levels were recorded while the subjects read a text aloud (in Danish), sustained an /aaa/ sound without opening the mouth (nasal-sound) and chewing gum.

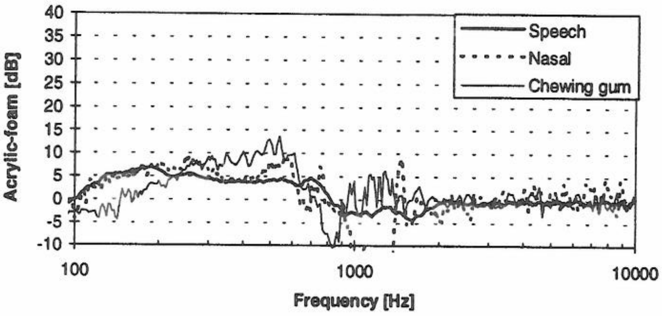
The difference between the sound pressure in the ear with the acrylic mould and in the ear with the foam plug is printed in **Figure 8.2.** There is one figure for every subject and the difference obtained with continuous speech, nasal sound and chewing gum are plotted in the same figure. Continuous speech is an average over 60 seconds and the difference curve is therefore smooth compared to nasal sound and chewing gum, which was only averaged over a few seconds.

The difference is a lot smaller than estimated from the experiments presented in chapter 7 and that the difference between the acrylic moulds and the foam plugs lie around 5-15 dB for f > 200 Hz, but the curve shape is not the same for all subjects. In general the differences are greater than the error due to the measurement technique, see appendix C.

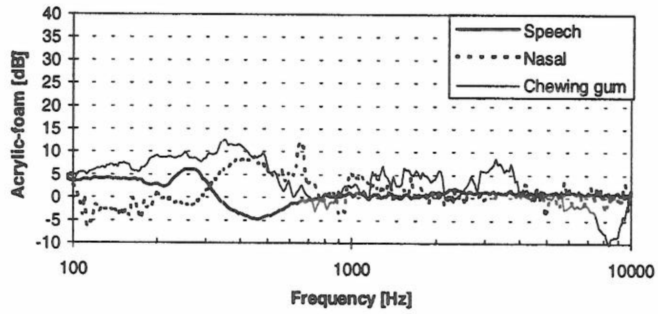
Subject 2, R: acrylic, L: foam



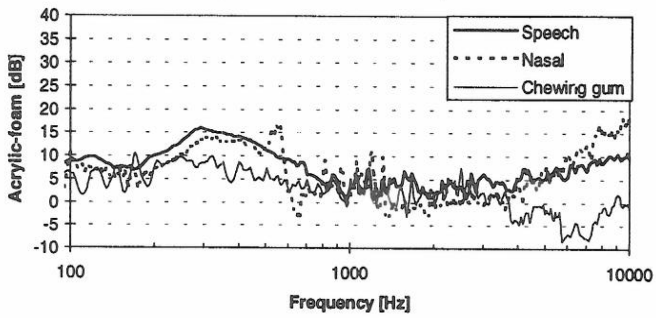
Subject 2, R: foam, L: acrylic



Subject 3, R: acrylic, L: foam



Subject 3, R: foam, L: acrylic



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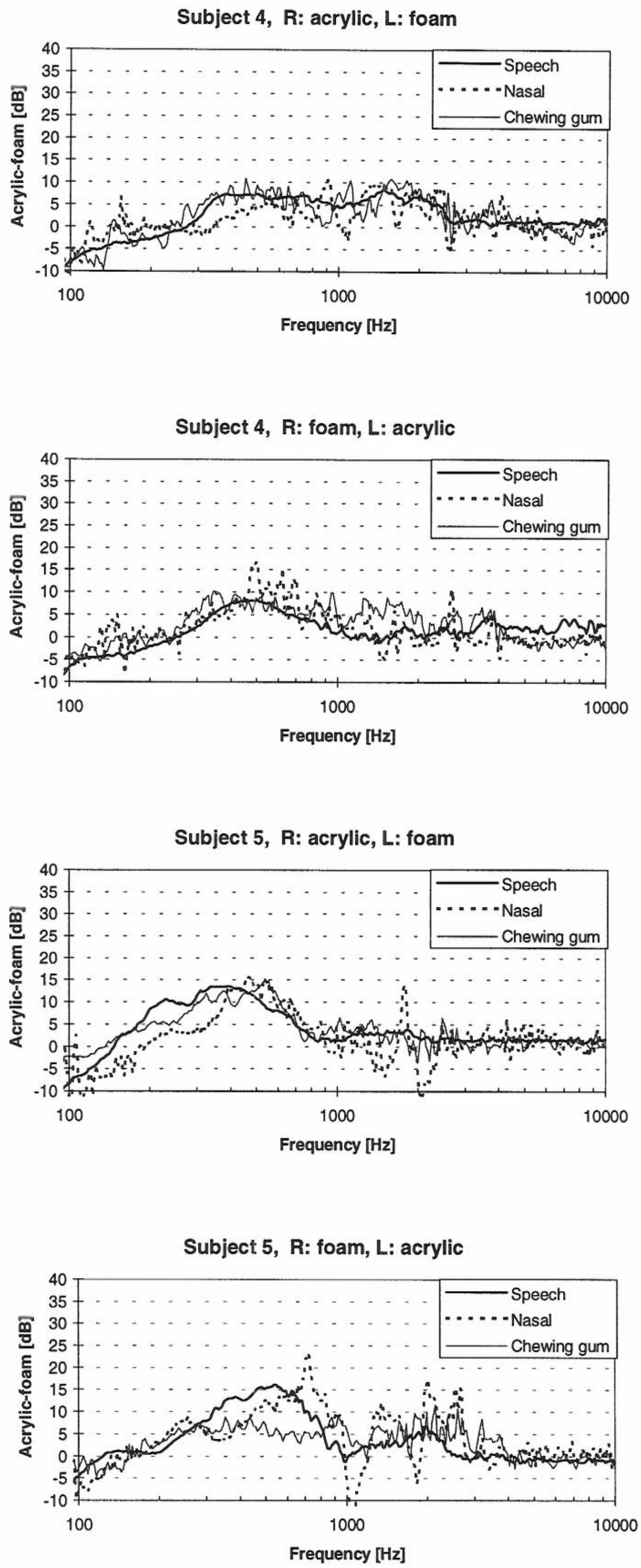


Figure 8.2 Difference in sound pressure level between one ear occluded with an acrylic mould and the other ear occluded with a foam plug. The three stimuli were produced by the subject. R = right. L = left.

The idea of using chewing gum and nasal sound was to observe the situations where the sound was generated mainly by jaw movements and mainly by the speech organs respectively. When you chew gum, the sound is generated because the jaw is clenched. In nasal sound the sound is generated with the speech organs but the jaw is closed. **Figure 8.2** shows that in general the difference for chewing gum is less or equal to the difference for nasal sound. Comparing speech and nasal sound gives also generally the same result.

The results from the chewing gum measurement is a lot less reliable than vocalized sounds. The reliability was not measured in this study, but experience from the pilot study in part I of this Ph.D.-project, [Hansen, 1997], showed that it was difficult to repeat the same sound while chewing on a crisp bread. The results in the present experiment is however better, partly because the crispy sound from the bread is not present with the chewing gum and the sound is averaged over a long time and partly because the open and occluded ear pressure is measured simultaneously and the subject did not have to repeat the chewing.

Right and left ear results are not exactly the same for the same subject, but the shape of the curves are similar. If the earmoulds were fitted identically in the right and left ear, the differences should be exactly the same, providing that the subject is anatomically symmetrical. The reason to the observed right and left ear differences is that the earmoulds are not fitted perfectly. Firstly, the acrylic moulds are not perfectly symmetrical, and one mould might fits a bit tighter than the other mould. Secondly, the foam plugs are inserted into an estimated insertion depth from the corresponding acrylic moulds, so if for example the right mould is a bit longer than the left mould, the right foam plug will be inserted a bit deeper than the left mould. Thirdly, even when the insertion was done with great care, it was not possible to insert the foam plug in exactly the same position every time. Note, that the foam plugs are inserted more shallow than recommended by the manufacturer, because the insertion depth had to be equal the to insertion depth of the acrylic mould.

The curves in **Figure 8.1** look differently from subject to subject, but the trend is the same: the occluded sound pressure level is greater with an acrylic mould than with a foam plug. It is presumed that the differences are due to the materials, leakage or mould and tissue vibrations. In the following, each of these explanations will be considered.

8.2 IMPEDANCE OF MOULD MATERIALS

The occlusion effect can be understood as a change in the acoustic impedance looking from the inside and outside of the ear canal. This was explained briefly in chapter 1. The sound pressure in the occluded ear canal depends, among others things, upon the acoustic impedance of the earmould. If the closed ear canal is regarded as a simple cavity (which is true only at low frequencies), the sound pressure generated from a piston placed in the wall of the cavity is given by:

$$p = q\frac{Z_{mould}Z_{cavity}}{Z_{mould} + Z_{cavity}} \qquad (8.2)$$

where;
q = volume velocity of the piston
p = sound pressure in the occluded tube
Z_{mould} = acoustic impedance of the earmould
Z_{cavity} = acoustic impedance of the cavity

A porous earmould material absorbs some of the acoustic energy and Z_{mould} declines. In the extreme situation, $Z_{mould} = 0$ and $p = 0$. A harder mould absorbs less energy and reflects a portion of the incoming sound waves. If $Z_{mould} \Rightarrow \infty$ the total impedance is the same as the rigid cavity.

8.2.1 Method

The impedance of some commonly used earplug materials was measured with an electro-acoustic transducer for measuring acoustical impedance (Tesla MAI-1). A detailed description of the experimental method is given in appendix B. The impedance of the earmould was determined with an audio analyzer (Brüel&Kjær Type 2012) by detecting the sound pressure generated with a piston of known velocity.

8.2.2 Results

Four devices were measured: a rigid cavity (reference), a 2 mm slice of an EAR foam plug, a normal EAR foam earplug and a foam earplug with a tube put through it with the tip of the tube extending 5 mm out from the plug and the other end closed with the tube of a hearing aid microphone.

The three earplugs have the same magnitude in impedance from 100 Hz to 10 kHz where the difference is less than 2 dB, see **Figure 8.2**. The curves are so alike for the three earplugs that it is very difficult to distinguish between the plugs. The phases are also practically the same (not shown in the figure). If the coating at the end of the EAR plug is cut off so that the air holes in the foam are no longer closed, the impedance will only change with 2 dB at 100 Hz. In comparison, the impedance for a 2 mm thin piece of EAR foam is shown. At 100 Hz the difference from a rigid surface is 16 dB.

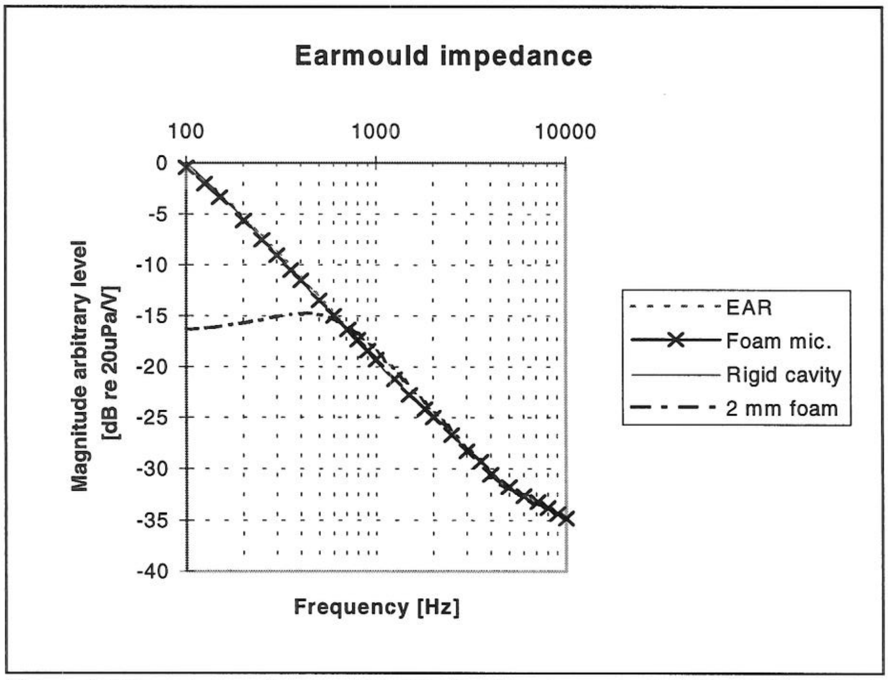


Figure 8.3 Impedance of earmoulds. The level is chosen so that the rigid cavity is 0 dB at 100 Hz.

The cavity made of metal has in theory a reflection coefficient of 1, i.e. an acoustical rigid cavity, implying that the earmoulds also can be considered as acoustical rigid surfaces when inserted into the ear canal. If a foam plug has an acoustical rigid surface, then an acrylic mould will also be acoustically rigid. This leave now 3 mechanisms that can affect the occlusion effect: leakage, earmould vibrations and tissue vibrations.

8.3 INFLUENCE OF THE MASS OF THE EARMOULD

Now, that it has been shown that both the foam and the acrylic mould has acoustical rigid surfaces, the observed sound pressure differences are still not so easy to interpret. The differences are due to different fitting of the two earmoulds. And it is not easy to tell if the differences are mainly due to leakage or the mass of the earmould. In order to eliminate some variables, a small experiment was performed with 1 subject where the only variable was the mass of the earmould. Experimental details are documented in appendix D.

8.3.1 Experimental method

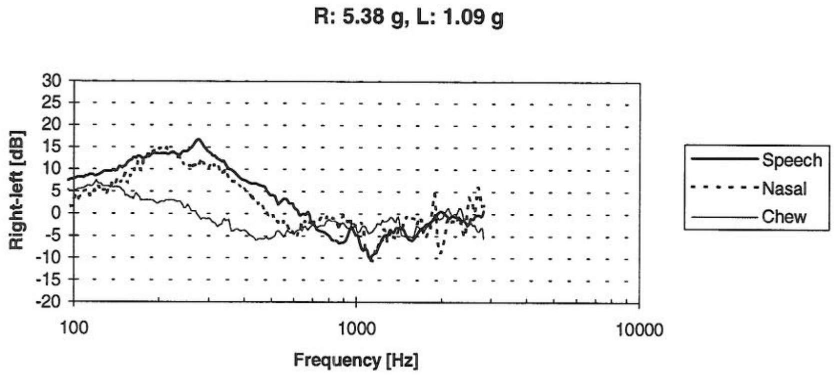
A pair of ITEC shells were moulded. The shells were filled either with foam material or with steel balls and some impression material. This made a light mould (about 1 g) and a heavy mould (about 5 g). The probe tubes were inserted through a drilled hole in the moulds. The sound pressures in the occluded ears were measured while the subject either read aloud for 30 s, sustained a nasal sound for 5 s, or chewed gum for 30 s. The occluded sound pressure was measured in both ears simultaneously while the right ear was occluded with a heavy mould and the left ear with a light mould and vise versa. Since the shell for one ear is exactly the same for light and heavy stuffing, the fit of the mould is the same and only the mass changes between the two situations. Each measurement was done three times and between the measurements the earmoulds were taken completely out of the ear and inserted again.

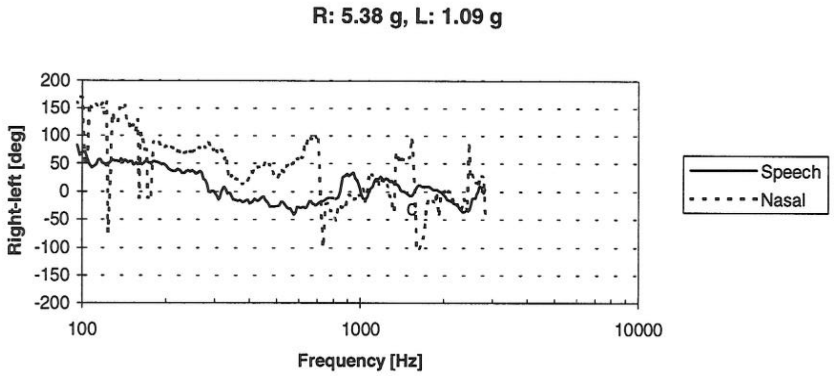
8.3.2 Results

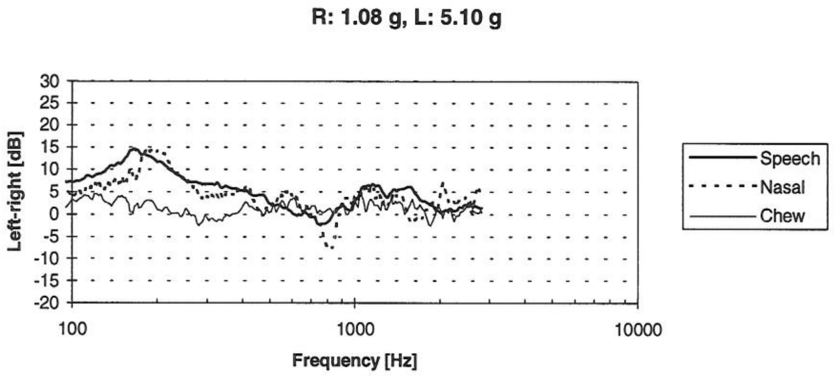
The prime data is given in appendix D. The repeatability was best for long-time averaged speech. The standard deviation for 3 repetitions below 1 kHz is up to 5 dB for speech, but the most frequent deviation was less than 3 dB. Between 300-1000 Hz the standard deviation for the nasal sound is 5-7 dB. The difference between the ears where both ears are occluded with a light mould or a heavy mould, the speech deviates a maximum of 5 dB below 700 Hz and in average 2.5 dB. The nasal sound and chew is less reliable, because the average time is only a few seconds and fluctuations are not averaged out.

The differences between the ears for nasal and speech stimuli goes up to 15 dB and are definitely larger than the differences caused by measurement uncertainties. Chewing gum does not give a significant difference. The magnitudes and phases are plotted in **Figure 8.4**. The curves for the nasal and speech are averages over three measurements. The phase for chewing gum is not shown because the phase fluctuates a lot.

These data suggest that the mass of the earmould alters the occlusion effect and secondly that the mould influence is not due to jaw movements.







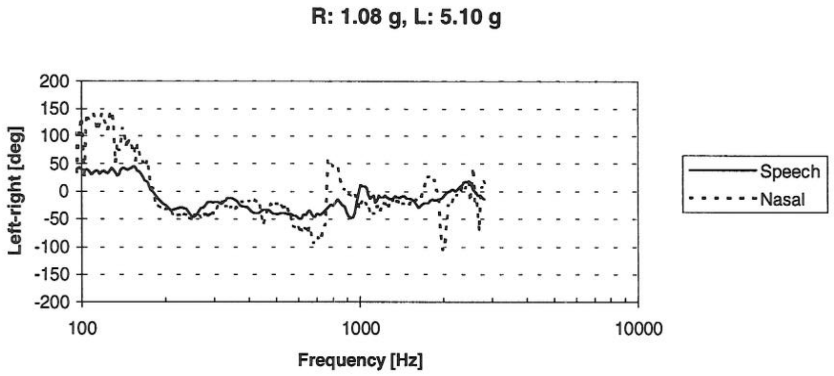


Figure 8.4 Magnitudes and phase differences for the sound pressure in left (L) and right (R) ear. Nasal and speech are averages of 3 repetitions. 1 subject.

8.3.3 Discussion of the results

The data in **Figure 8.3** shows that the occluded ear pressure increases when the earmould mass increases at least within the range of the masses used here, i.e. 1-5 g. It agrees with hypothesis 2 because the earmould-cartilage system resonances at a lower frequency when the mass of the mould increases.

Another explanation could be hypothesis 3 where the cartilage covered with an earmould is less able to move when the earmould has a mass of 5 g than a 1 g mould. With a 5 g mould, more of the vibration energy might be transmitted to the free cartilage, which then vibrates with a greater volume velocity.

Both the acrylic-foam and the (5 g - 1 g) mould experiment show that the mechanical properties has important influence on the sound pressure generated in the occluded ear. The influence of the earmould can be understood as a movement in the vertical plane (y-direction) caused by the tissue and another movement in the horizontal plane (x-direction) from the earmould. **Figure 8.5** shows the principle in this theory.

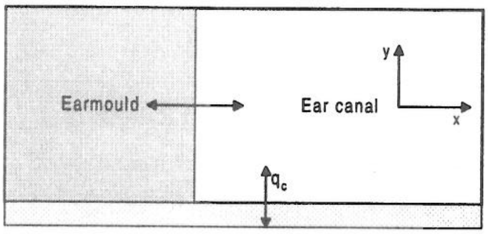


Figure 8.5 Principle of sound generation in the occluded ear. The mould moves in the x-direction. The tissue moves in the y-direction.

Bárány (probably in 1938) [Hood, 1962] proposed that the occlusion effect was due to the inertia of the earmould. Békésy [1949] showed that the skull oscillates vertically when it is excited by one's own voice. Oscillations of the bone in the ear canal excites the cartilage and because of the elasticity of the cartilage, it may vibrate differently than the bone. An earmould inserted into the cartilaginous ear canal will then be pushed by the cartilage and it will vibrate either with the cartilage or relative to the cartilage. The result is that the occluded volume changes because the bony ear canal oscillates relative to the cartilaginous ear canal and again relative to the earmould.

The principle is the same when the jaw moves. The cross-sectional shape of the cartilaginous ear canal changes and the ear canal walls pushes on the earmould which then moves. But as explained previously, the jaw movements during speech are so slow that the pressure generated by slow earmould movements are not audible.

Vibrations of the earmould has been studied for the purpose of hearing protection research. Air transmitted sound sets the earplug into vibrations and that creates a sound pressure inside the ear canal, [Schroeter and Els, 1982]. The result is that there is a limit for how well the specific earplug is able to attenuate the sound and thereby how well it protects your hearing. Shaw [1982] made a very simple model to calculate the transmission loss through an earplug due to vibrations and this model was later repeated by Schroeter [1983]. But in the case of one's own voice, the body conducted sound contributes much more than air excited mould vibrations. Therefore, only the body excited mould vibrations will be modeled and a literature review did not succeed in finding any references on a model of this mechanism.

8.4 EARMOULD VIBRATIONS (X-DIRECTION)

The relative vibrations between the earmould and the ear canal depends on the mechanical impedance of the tissue on the ear canal wall. Unfortunately most literature on human tissue vibrations are concerned about very low frequencies (f < 200 Hz) or frequencies in the MHz range. There is very few data on vibrations of human tissue at the auditory frequencies therefore data from 6 studies are compared in appendix I. The only data that was found on the ear canal, came from Schroeter and Els [1982] who had a special transducer that covered all the outer 10 mm surface of the ear canal wall. More data was found on the impedance of the mastoid, which is a standardized measurement.

The general model used for the human tissue impedance is calculated by , [Ishizaka et al., 1975] (in acoustic units):

$$Z_{tissue} = \left(R + jM\omega - j/(C\omega)\right)/A^2 \quad [Pa \cdot s/m] \tag{8.3}$$

where;
A = cross-sectional area of the ear canal
M = mass of the tissue
R = resistance of the tissue
C = compliance of the tissue

8.4.1 Tissue impedance influence on earmould vibrations

The earmould is assumed to vibrate in the x-direction and to be pushed by the cartilage in the ear canal, which in this example is modeled as a simple cavity. The model and the results are shown in **Figure 8.6**.

The ear canal is modeled with the default parameters and the pressure is measured 3 mm behind the earmould. The transducer is a circular disk placed on the mastoid. The tissue values are converted to a reference area of 1.75 cm^2 as prescribed in IEC 373 and then normalized to the area covered by the earmould = $(2\pi r)(7.8)$mm^2 and then converted into acoustic units.

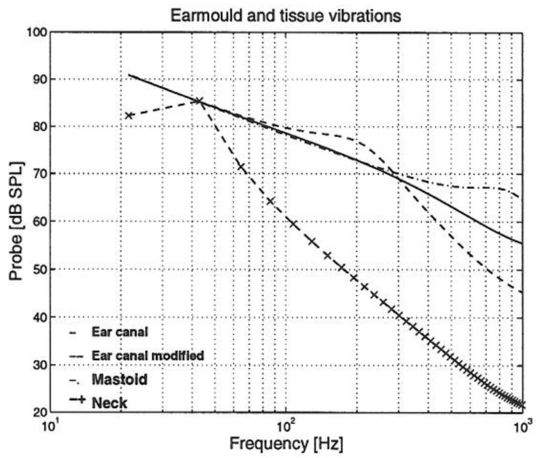


Figure 8.6 Pressure in the occluded ear canal generated by earmould vibrations when the tissue impedance is varied. Mechanical values: Earmould = 3 g. Ear canal tissue: M = 0, C = $1.89 \cdot 10^{-4}$ m/N, R = 11.9 Ns/m. Ear canal tissue [Schroeter and Els, 1982] modified: M = 0, C = $1.89 \cdot 10^{-4}$ m/N, R = 3 Ns/m. Mastoid, [Wagowske et al., 1995]: M = $2.58 \cdot 10^{-4}$ Ns^2/m, C = $1.09 \cdot 10^{-5}$ m/N, R = 10.5 Ns/m. Neck, [Ishizaka et al., 1975]: M = $2.16 \cdot 10^{-4}$ Ns^2/m, $2.25 \cdot 10^{-2}$ m/N, R=0.2 Ns/m. Input: body source volume velocity, q = 10^{-6} mm^3/s.

The example shown here is with a 3 g rigid earmould (C_m = 0). The calculations confirm what must be expected, that if the volume velocity of the tissue is given, then a more elastic tissue makes the mould vibrate less than a more rigid tissue. In **Figure 8.6** Schroeter's model of the ear canal (see chapter 3) is used and a modified version where the friction is set to a lower value (3 Ns/m) which corresponds better to the measured one. Decreasing the friction provokes the resonance magnitude of the system, here around 100-200 Hz. The neck impedance is only shown for reference, because Williams and Howell [1990] uses this impedance in their model of the ear canal. In the present model, the soft ear canal tissue will be estimated with Schroeter's model of the ear canal plus a mass, Ms = $0.5 \cdot 10^{-4}$ Ns^2/m and the bony part with Warskowge's model of the mastoid.

8.4.2 Earmould vibrations

The movement of the earmould in and out of the ear canal (x-direction) is modeled with the same model as used in **Figure 8.6** but where the ear canal is modeled with a tube.

The amplitude of the earmould oscillations will depend upon the impedance of the cartilage, the mass of the earmould and the elasticity of the earmould. Furthermore, the velocity of the earmould will also be influenced by friction between the ear canal cartilage and the earmould material. The occluded ear sound pressure as a variation of the earmould mass is shown in **Figure 8.7**. The moulds with masses 1,5 and 100 g are rigid moulds and the 0.4 g foam plug has the same elasticity as and E-A-R foam plug. If the mould is rigid but very heavy (100 g), the sound pressure decreases, as expected.

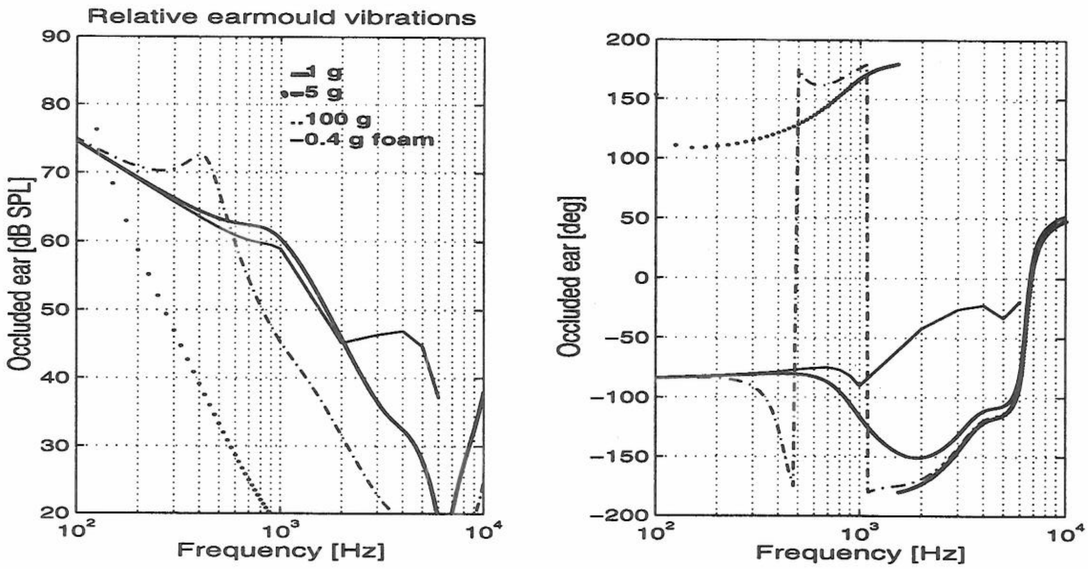


Figure 8.7 Estimated effect of mould vibrations. Rigid moulds (1-100 g). Foam mould (0.4 g). Sound pressure behind the earmould. Default parameters. Input: body source volume velocity, q = 10^{-6} mm^3/s.

The difference between the 1 g and the 5 g curve in **Figure 8.7** does not look like the difference observed in the experiment with a light and a heavy mould. Firstly, the resonance frequency for the 5 g mould is too high. Secondly, the sound pressure rolls off fast above the resonance frequency, so that now the 1 g mould creates a larger sound pressure which does not agree with the measurements. Nor is the difference between the 0.4 g and the 5 g curves similar to the difference measured with a foam plug and an acrylic mould in the same persons. Consequently, the hypothesis about the occlusion being created by movements of the earmould must be rejected.

However, it does not mean that the mould is immobile. Subjective observations report that sometimes the mould is even so loose, that it is possible to actually feel that the mould vibrates.

8.4.2.1 "My head vibrates !"

In the acrylic-mould experiment a subject spontaneously said: "It sounds like my head vibrates !", when he tried the earmould for the first time. The same thing happened in each of the two following session weekly apart. Listening to the sound pressure in the subject's occluded ear through the probe microphone confirms that scratching / resonance sounds were clearly audible. The phenomena was only present in the subject's left ear and can therefore be illustrated partly by looking at the occluded sound pressure in left and right ear simultaneously, see appendix C, and partly by looking at the occlusion effect for the left and right ear.

The occlusion effect in the right and left ear of subject 3 is plotted in **Figure 8.8**. In session 'm', strong resonances occurs in the left ear, whereas there is no resonances in the right ear. In session 'g' the earmoulds were greased in a thick layer of Vaseline. The effect of using Vaseline is clear; the resonances in the left ear are nearly gone! Conclusively, the Vaseline hinders the earmould to resonate.

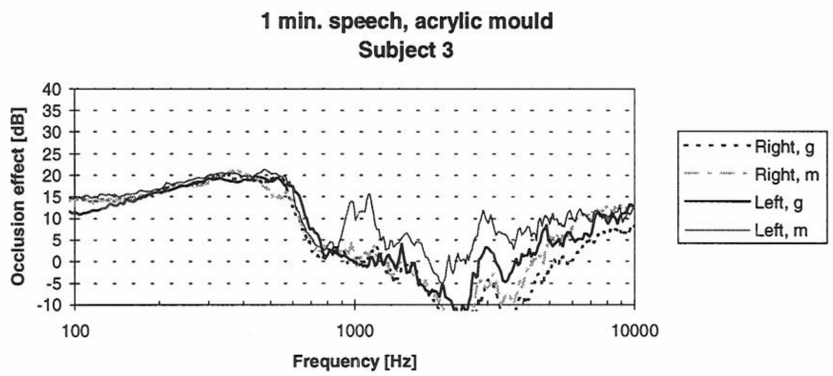


Figure 8.8 Occlusion effect in right and left ear. Acrylic moulds. Measurement 'm' and 'g' was performed with 1 week apart. In session 'g' the earmould was greased with a thick layer of Vaseline.

8.5 DISTRIBUTED VIBRATIONS (Y-DIRECTION)

Previously in this report, the body conducted sound was modeled as a volume velocity generator located in one single point. The model presumed that the earmould could be regarded as a rigid wall (see chapter 1). The value of the velocity was then weighted with a gain factor of the insertion depth: gain=$(1-l_{insertion}/l_{ear\ canal})$.

Another approach and possible explanation of the experimental results is to imagine that during speech, the vocal cords produce a constant volume velocity that is distributed through the whole body. Looking at the ear canal, the volume velocity will then be distributed over the ear canal surface depending on the loading at each point of the ear canal. Now, looking in the y-direction at **Figure 8.5** the tissue wall can be divided into 3 sections: the bony wall, the free cartilage and the cartilage loaded with the earmould as illustrated in **Figure 8.9**. The principle of dividing tissue was also used in the middle ear model by Shaw and Stinson [1981], where the eardrum was divided into 2 pistons, see appendix H.

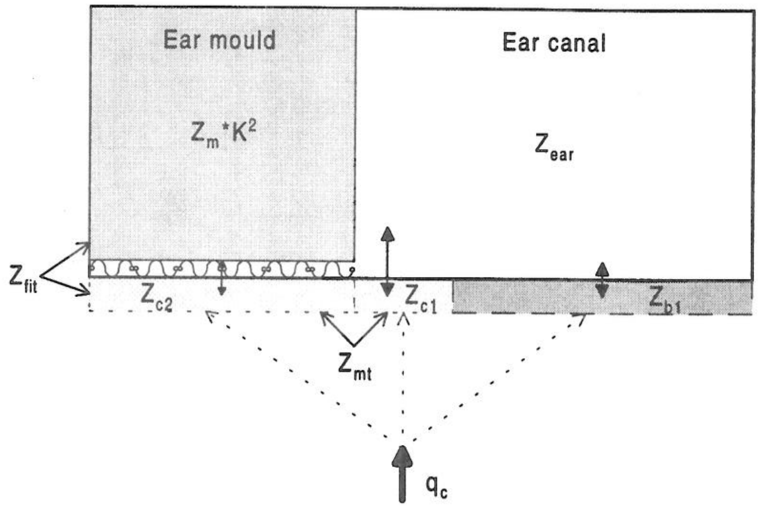


Figure 8.9 Illustration of the hypothesis of distributed volume velocity source.

One obvious problem here is that the tissue is assumed to move in the y-direction. The occlusion effect model is 1-dimensional and presumes that the sound wave travels in the length direction of the ear canal (the x-direction). The areas A_2 an A_1 is the surface tissue area covered by the earmould and the non-covered (free) tissue surface respectively. The acoustical impedance network is drawn in **Figure 8.10**. The free cartilage is named Z_{c1} and the free bone Z_{b1}. The covered tissue is called Z_{c2} and Z_{b2} respectively.

The tissue is assumed to be a piston divided into 2 major parts, the covered part, A_2, and the free part, A_1, and it is therefore necessary to normalize each part with the respective piston area. The normalizing coefficient is called K. Both the tissue and the ear canal volume depends on the tissue surface area but the earmould mass does not. Consequently, the earmould impedance, Z_m, should not be transformed with K, and the impedance is therefore weighted with K^2 so that the transformer has no effect on the earmould.

The covered tissue and the free tissue is, of course, connected and the coupling between the two parts is represented with the impedance Z_{mt}. The value of this impedance is not known but it must logically be similar to the tissue impedance. In the cartilaginous part, Z_{mt}, the cartilage impedance is weighted with the ratio between the covered area and the area of the total tissue surface. If this Z_{mt} is used, then the model predicts that the sound pressure in the occluded ear has a maximum at 200 Hz with a 5 g mould inserted 7.8 mm which agrees very well with the experimental results. If $Z_{mt} = 0$, the resonance occurs at 100 Hz and if $Z_{mt} = \infty$, the resonance disappears. In the bony part the coupling is stiff and Z_{mt} is infinite.

The impedance Z_{fit} models the connection between the earmould and the ear canal wall. It is the most difficult impedance to interpret because it depends on the shape of the earmould. For example, if the earmould tip is tapered, the resonance in the occluded pressure shifts upwards in frequency, [Mueller, 1994], and it is not due to an increase in the effective occluded volume, because this would decrease the resonance frequency.

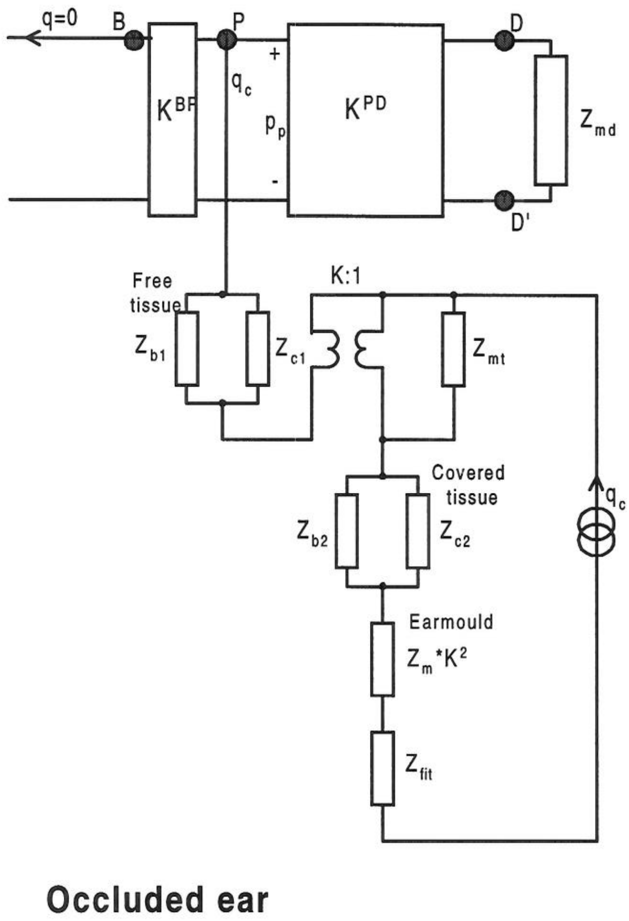


Figure 8.10 Acoustical network to simulate the distribution of a given volume velocity source. The body conducted sound is presented with the volume velocity, q_c, and the output is the sound pressure, p_p, measured 3 mm from the earmould. The proportion of the tissue impedances Z_{c1}, Z_{c2}, Z_{b1} and Z_{b2} depends on the insertion depth. K^{BP} and K^{PD} is the occluded ear canal. Z_{md} is the middle ear. See text for more details.

The model assumes here that:

- the generator is an ideal constant volume velocity source.
- the volume velocity is constant and is distributed over the tissue surface depending on the tissue impedance and the loading alone.
- that the earmould is sealed to the tissue and follows the movements of the tissue.

In case of an open ear, the earmould is removed and the rigid wall ($q = 0$) is substituted with the outer part of the ear canal, the concha and radiation impedance.

The predicted sound pressure from the model in **Figure 8.10** in the occluded ear canal is plotted in **Figure 8.11** for different insertion depths. When the mould is inserted into the bony ear canal, the sound pressure level decreases. At 11 mm insertion, the phase has shifted and the phase shifts at 10.6-10.7 mm. The sound pressure drops significantly from 11 mm to 13 mm in accordance with the hypothesis.

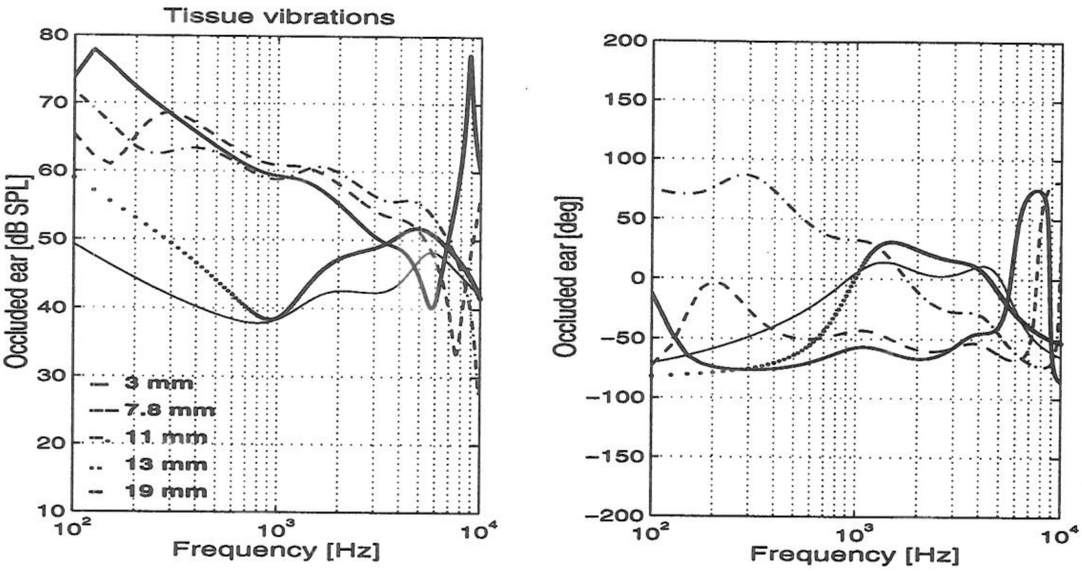


Figure 8.11 Estimated pressure in the occluded ear canal (3 mm behind the earmould) for different insertion depths. Source: tissue vibration. Default parameters. Mould mass = 3 g. Input: body volume velocity source = 10^{-6} mm^3/s.

The idea in hypothesis 3 is that the distribution of the volume velocity source depends on the load and that the sound pressure depends on the earmould mass, as shown in **Figure 8.12** for rigid acrylic moulds (1-100 g) and a foam plug (0.4 g). The mechanical properties of the foam plug is the same as for E-A-R plug foam material, [Berger, 1990], the calculation of the foam plug impedance is given in appendix C.

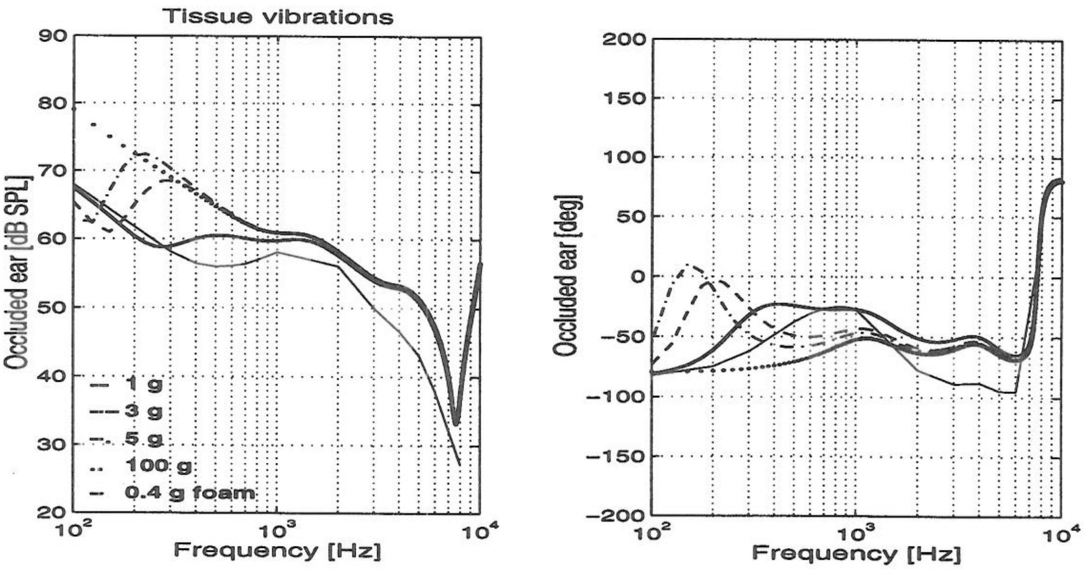


Figure 8.12 Estimated pressure in the occluded ear canal (3 mm behind the earmould) for different earmoulds. Rigid moulds (1-100g). Foam mould (0.4 g). Generator: tissue vibration. Default parameters. Insertion depth = 7.8 mm. Input: q = 10^{-6} mm^3/s.

8.5.1.1 A light and a heavy mould

From **Figure 8.12** it is seen that a more heavy mould creates a larger sound pressure in the ear canal than a light mould. In fact a very heavy (100 g) and rigid mould creates the largest sound pressure because the tissue is unable to move and all the energy will be transmitted to the free tissue. The lightest and most elastic earmould (the foam) will therefore create the least sound pressure because the energy also is transmitted to the covered tissue and does not radiate into the occluded ear canal.

This contradicts the effect of the earmould vibrations, where the sound pressure decreased with mass below the resonance frequency, see **Figure 8.7**. But the tendency is the same as measured with a heavy and a light mould as well as a foam plug. The estimated differences between a 5 g and a 1 g mould is shown in **Figure 8.13** together with the measured data. The maximum difference occurs at different frequencies and the estimated difference curve lies between the two measured differences. Hence, the estimation also agrees well in the magnitude except below 200 Hz. The estimation rolls off more towards lower frequencies and becomes negative whereas the measured data rolls less sharply off and stays positive.

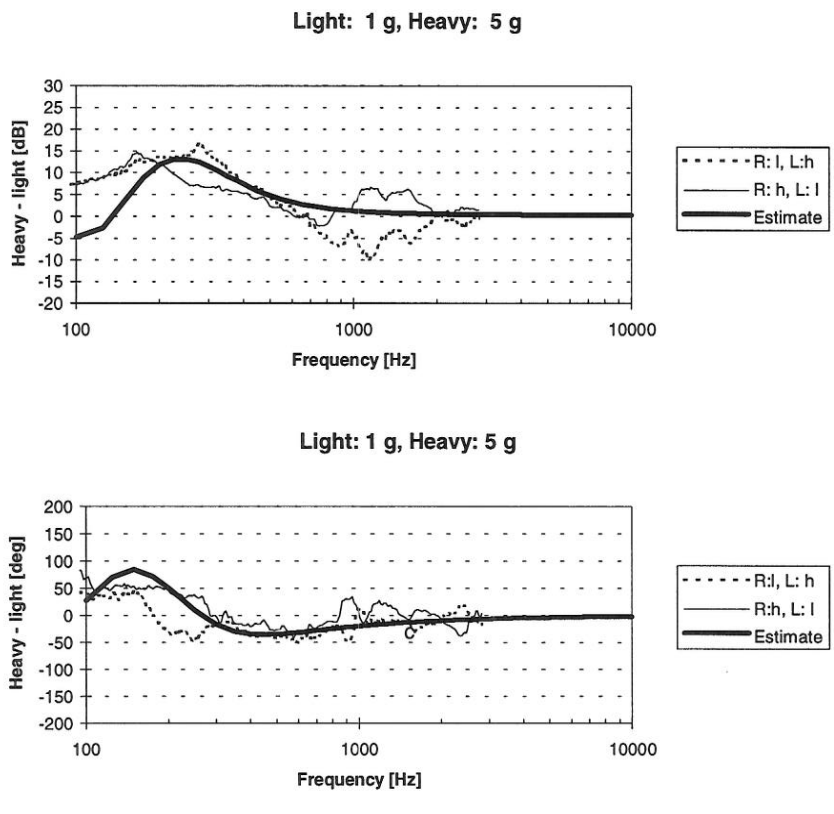


Figure 8.13 Estimated and measured difference of the sound pressure in the occluded ear with a heavy mould (approx. 5 g) and a light mould (approx. 1 g). Stimulus: subject's own voice, continuous speech. 1 subject. Estimated with default model values. R = right, L = left, h = heavy, l = light. Top panel: magnitude. Bottom panel: phase.

The estimated and measured phase difference agree well. The phase is negative from 300 Hz and about -50 deg and then it goes slowly towards 0 deg difference. Below 300 Hz, the phase-estimation agrees best with the heavy mould in the right ear and the light mould in the left. It is possible to adjust some of the values in the model to give the estimated curve a better fit. But it is not justifiable to try and fit the curve to data measured on only 1 subject. The values in the model, will therefore be kept as the values as described and based on literature data. Above 1 kHz the difference becomes negligible because the mould mass influences the low frequencies only. Furthermore, the measurements are less reliable at higher frequencies, for example the sound pressure is about 40 dB less at 3 kHz than at 300 Hz.

8.5.1.2 An acrylic mould and a foam plug

The estimations and measured data with acrylic and foam plugs are shown in **Figure 8.14**. Only the case with the foam in the right ear is shown because subject 3 behaves different from the general tendency when the right ear is fitted with the acrylic mould, see appendix C. The foam plug weights 0.4 g and the mass of the acrylic mould is included in the model as 4 g. The subject's moulds weighted between 3.2 and 5.7 g with an average of 4.4 g. Only frequencies below 1 kHz are estimated because, the mechanical data on the foam plug was only available up to 1 kHz.

The measured sound pressure level differences go up to 15 dB and are shaped differently between subjects and the estimated difference lies in the middle of the measured curves. Except from subject 2, who shows quite a flat difference curve, the estimated curve follows the shape of the measured data.

The estimated phase follows subject 4 and 5 nicely above 200 Hz. Below 200 Hz, the estimation decreases towards lower frequencies whereas the measurements show the opposite.

It might be possible to fit the estimate better to every single subject. The deviation between subjects are namely caused by several factors, such as the size of the occluded ear canal, the weight of the earmould (subject 5 has the heaviest mould and subject 4 the lightest), leakage between the earmould and the ear canal wall, and the individual shape and fit of the mould. The latter is very difficult to model but the ear canal size and the weight is possible to adjust in the model.

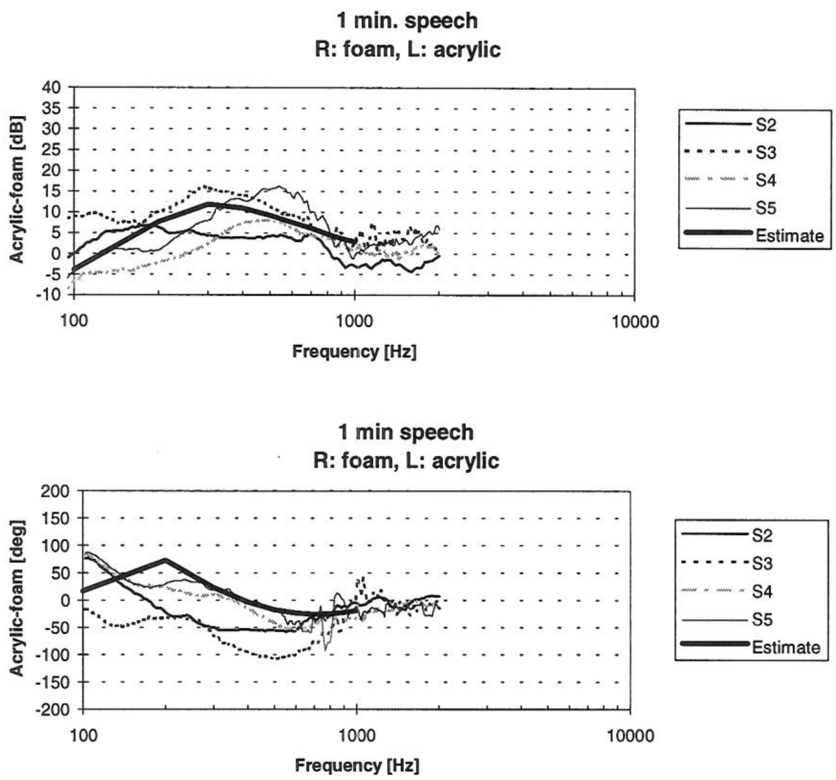


Figure 8.14 Estimated and measured difference of the sound pressure in the occluded ear with an acrylic mould and a foam plug. Stimulus: subject's own voice, continuous speech. Estimated with default model values. Top panel: magnitude. Bottom panel: phase.

The occlusion effect estimated with this model agrees well with measured occlusion effects. This will be shown later in chapter 12. At the lowest frequencies (below 100-500 Hz) the estimated occlusion effect is in some cases greater than the measured. The reason is that the present model presumes that the earmould is sealed to the ear canal wall but in practice this is, of course, not true. Just a small leakage between the ear canal wall and the earmould decreases the occlusion effect at low frequencies. Leakage is not included here, but it is possible to model a leakage quite well and this will be discussed in the next chapter.

9. NATURAL LEAKAGE

A standard hearing aid mould will never be perfectly tight, there will always be a small leakage between the earmould and the ear canal wall, even when the mouth is closed (static leakage). During speech or chewing the leakage (dynamic leakage) becomes greater, because the jaw moves. Roughly speaking, a leakage reduces the amount of body conducted sound at low frequencies and allows more air transmitted sound to enter the ear canal depending on the size of the leakage.

9.1 MODEL OF A LEAKAGE

To begin with, the theoretical influence of a leakage will be analyzed with a tube model and a slit model. Later, the models of leakage will be compared to experimental data.

9.1.1 Tube model

Traditionally, the leakage is modeled as a tube with constant cross-section (in Oticon is used a diameter = 1.4 mm has been used). The reason is probably that the mathematical model for a cylindrical tube is well-known and is rather simple (see also chapter 1). The open tube is given by:

$$Z_{open\ tube} \cong j\frac{\rho c}{S}\tan kl_{eff} \qquad (9.1)$$

Looking from the inside and out, the tube has a radiation impedance that can be modeled as a piston in an infinite baffle. Because the tube leakage is so small, the imaginary part of the radiation impedance corresponds to a prolongation of the tube and the real part to a resistance, [Jacobsen, 1993]. The additional length is:

$$\Delta l \cong 0.85a \qquad (9.2)$$

where;
a = radius of the tube

The radiation resistance, R_{rad}, is:

$$R_{rad} = \frac{\rho c}{\pi a^2}\frac{(ka)^2}{2} \qquad (9.3)$$

The tube model is however not a really good estimation of the real situation in the static case, for example the situation with an earmould inserted in an artificial ear or when the hearing aid wearer has the mouth closed. A tube has a length resonance which does not appear in the static case and in order to make the tube model more suitable to the real situation it has been common to put an arbitrary loss impedance across the tube, [Macrae & McAlister, 1989]. In theory, the loss-free tube is only valid for $0.03/f^{1/2}$ < radius < 25/f, so for example, the formula for a leakage with diameter 1.4 mm is only valid above 1836 Hz and below 35 kHz. Consequently, the loss-free tube model (as used for the ear canal) is not valid for the leakage, because the diameter is so small that loss due to viscosity and heat loss at the walls can no longer be neglected. The real wave number, k, must be substituted with a complex number, $k_{small} = k - j\alpha$ in equation (9.1). When only plane waves exist and when $\omega\rho a^2/\eta > 100$ (which is fulfilled for a > 0.5 mm and f < 1000 Hz), the absorption coefficient α is given by, [Kinsler et al., 1982]:

$$\alpha = \frac{1}{ac}\sqrt{\frac{\eta\omega}{2\rho}}\left[1+(\gamma-1)\sqrt{\frac{\kappa}{\eta C_p}}\right] \tag{9.4}$$

where in air at 20°;
η = viscosity = 18.1*10^{-6} Pa s
γ = ratio of specific heats = 1.402
C_p = specific heat at constant pressure = 1006 J/kg K
κ = 2.56*10^{-2} W/m K
a = radius of the tube leakage

The effect of the losses are about 4 dB for a 1 mm diameter tube and 1 dB for a 2 mm diameter tube at the resonance frequency. Thus, for a normal sized vent greater than 2 mm, the losses can be neglected but not for a leakage or a pressure relief vent (0.6 mm).

Note, that the loss due to a small diameter must not be confused with the loss due to mechanical friction between the earmould and the ear canal wall. For example, the loss resistance in Weinrich's model for occlusion effect, see chapter 3, is due to mechanical friction.

9.1.2 Slit model

The reason why the tube model is not working so well, at least not when the mouth is closed, is because the leakage in practice consists of several narrow slits randomly placed, [Macrae and McAlister, 1989]. For practical reasons - and because the number and size and positions of the slits are unknown - the leakage will be modeled as one single rectangular slit as illustrated in **Figure 9.1**.

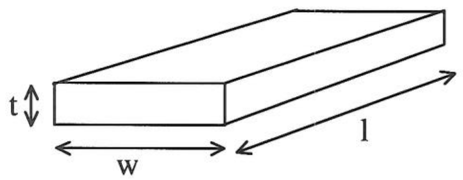


Figure 9.1. A narrow slit leakage.

When the slit is narrow, t << w, then the impedance of the slit is dominated by the friction and inertia between the air particles [Beranek, 1986]:

$$Z_s = \frac{12\eta l}{t^3 w} + j\omega\frac{6}{5}\frac{\rho l}{tw} \tag{9.5}$$

where;
t = thickness of slit
w = width of slit
l = length of slit

This lumped model is valid if $t < 0.003/\sqrt{f}$. Thus, at 100 Hz $t_{max} = 0.3$ mm and at 1000 Hz $t_{max} = 0.09$ mm, [Beranek, 1986]. As shown later, the static leakage is estimated to have about t = 0.2 mm, so the slit model cannot even be used at 1 kHz, at least theoretically seen. But the slit model will be used at higher frequencies anyway, because the general model of a larger slit is not so easy to interpret. When $w < (c/2f)$ only plane waves propagates [Beranek, 1986], so only plane waves propagates below 10 kHz if $w < 17$ mm and below 6 kHz if $w < 29$ mm. An average ear canal has a circumference of 28 mm in the outer part, so the model can at least be used up to 6 kHz.

The main reason to use equation (9.5) is that the general model of a larger slit is not easy to interpret. The slit is flanged at both ends which can be accounted for by adding a correction to the length, getting $l_{new} = l + l_{add}$, where l_{add} is given by, [Beranek, 1986]:

$$l_{add} = 0.85\sqrt{\frac{wt}{\pi}} \qquad (9.6)$$

The radiation resistance in front of the outer outlet of the flanged slit is:

$$R_{rad} = \frac{(kt/2)^2 \rho c}{2\pi (t/2)^2} \qquad (9.7)$$

9.1.3 Acoustic influence of a leakage

When speaking, body produced sound will travel out of the ear, if there is a leakage, and the air conducted sound will enter the ear canal. Hence, the occlusion effect is influenced by a natural leakage. For illustration purpose only, we will look at the acoustic influence on the body conducted sound. **Figure 9.2** shows the simulated sound pressure in the occluded ear canal with a tight mould, a slit shaped leakage and a tube shape leakage.

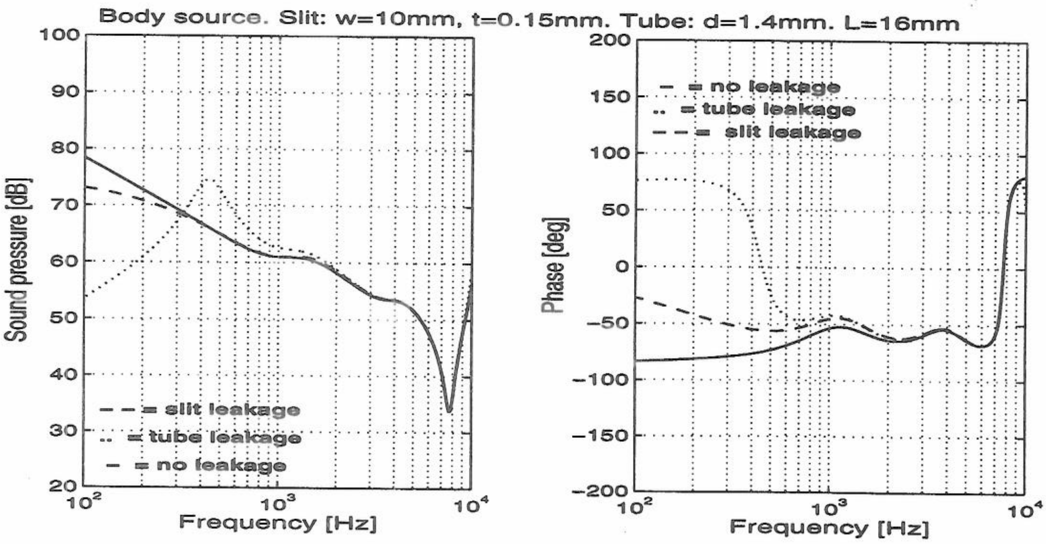


Figure 9.2. Simulated effect of a slit shaped leakage and a tube shaped leakage on body produced sound in the ear canal. Body source = 10^{-6} mm^3/s.

The tube and the slit have the same cross-sectional area (1.5 mm^2) but the acoustic behaviour is clearly not the same. An important point to make is that a tube shaped or a square slit leakage actually increases the occlusion effect at the tube resonance!

9.1.4 Approximation of the slit model to the tube model

The leakage will constantly change it's shape when the person speaks, hence it is reasonable to presume that when the mouth is open, then the slit leakage becomes broader and thicker. A rough estimate of the size of the leakage with open mouth can be based on Oliveira's [1995] data for 5 mm jaw lowering. Assuming that the average ear canal diameter is 7.1 mm, the new diameter is 0.3 mm greater in the anterior-posterior and 0.07 mm greater in the inferior-superior direction. The changes are greatest for the anterior and the inferior walls (the front and the bottom walls).

The simple slit model is not valid for this situation because the slit is too thick compared to the width and the frequency range. However, a transmission line model of a slit is not easy to determine[1] and a tube model is theoretically more correct to use.

When only plane wave propagates the acoustic behavior of the slit can be approximated with that of the tube with radius (t/2)/n, where n is a shape factor, [Attenborough, 1983]. The estimated radius of the tube, then becomes the effective acoustical radius and not the physical radius. **Figure 9.3** compares a tube leakage with a slit leakage, where the slit has a square cross-section ($w = t$). The calculations shows the 'vent response' i.e. the difference of the sound pressure in the occluded ear with and without a leakage. The slit and tube model agree well within a few dB. Of practical reasons and in order to keep just one leakage model in the occlusion effect model, the slit model will be used even in the dynamic case.

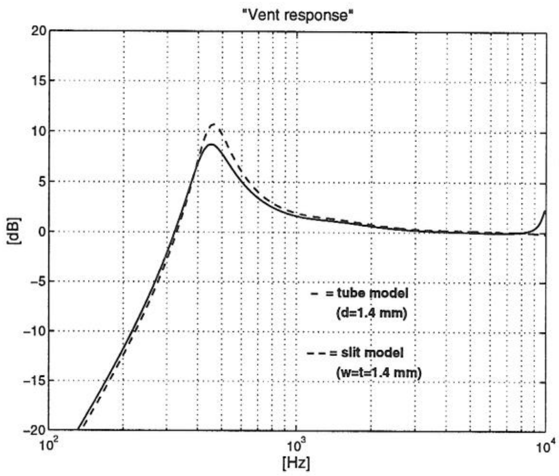


Figure 9.3. Simulated 'vent response' with a tube and a slit leakage.

9.1.5 Variation in leakage dimensions

The attenuation from the occlusion effect depends on the size and shape of the leakage. **Figure 9.4** and **Figure 9.5** shows some examples of the effect of varying the width and the thickness of a slit leakage looking at the sound pressure in the occluded ear with both air and body sound present.

At the lowest frequencies, a larger leakage attenuates the sound in the occluded ear but at higher frequencies the sound pressure increases. For example for a slit of thickness 0.2 mm

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[1] A solution for the rectangular tube regardless of dimensions from the general 2 port expression for a tube given by, [Stinson, 1997]:

$$\begin{bmatrix} p_{out} \\ q_{out} \end{bmatrix} = \begin{bmatrix} \cosh ml & Z_0 \sinh ml \\ (1/Z_0)\sinh ml & \cosh ml \end{bmatrix} \begin{bmatrix} p_{in} \\ q_{in} \end{bmatrix}$$

where m and Z_0 must be derived from equation (80) and (81) in [Stinson, 1991].

the 'critical' frequency for a 10 mm wide slit is 400 Hz compared to a small leakage with width 5 mm. Increasing the thickness and/or the width moves the resonance up in frequency. The phases get numerically smaller for larger slits, which is to be expected. The phases follow the magnitudes so when the sound pressure increases the phases get more negative and when the sound pressure gets smaller towards the reference sound pressure at 60 dB SPL, the phases go towards zero.

The length does not influence the leakage so much. A length of 23.3 mm (a BTE hearing aid) gives a little less than 3 dB higher pressure at 100 Hz than a slit of length 16 mm. Between 300-10,000 Hz the variation is only ±1 dB (not shown).

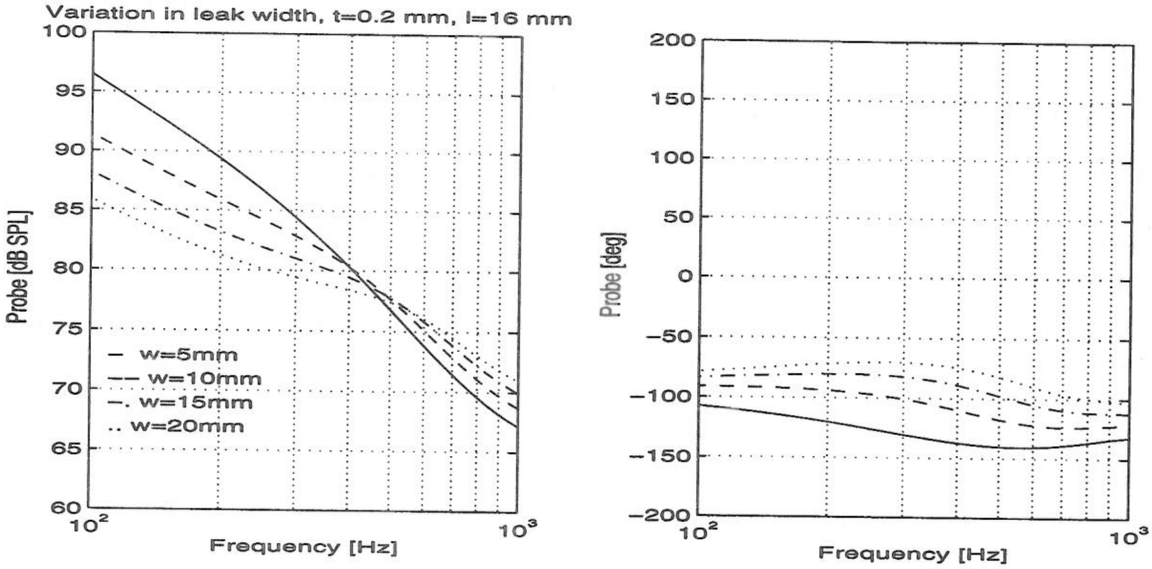


Figure 9.4 Sound pressure in the occluded ear (3 mm behind the earmould) with variation of the slit leakage width. Thickness is 0.2 mm and length=16 mm. Input=60 dB SPL air conduction. See **Figure 9.11** for the value of the body source. Phase reference is a zero phase.

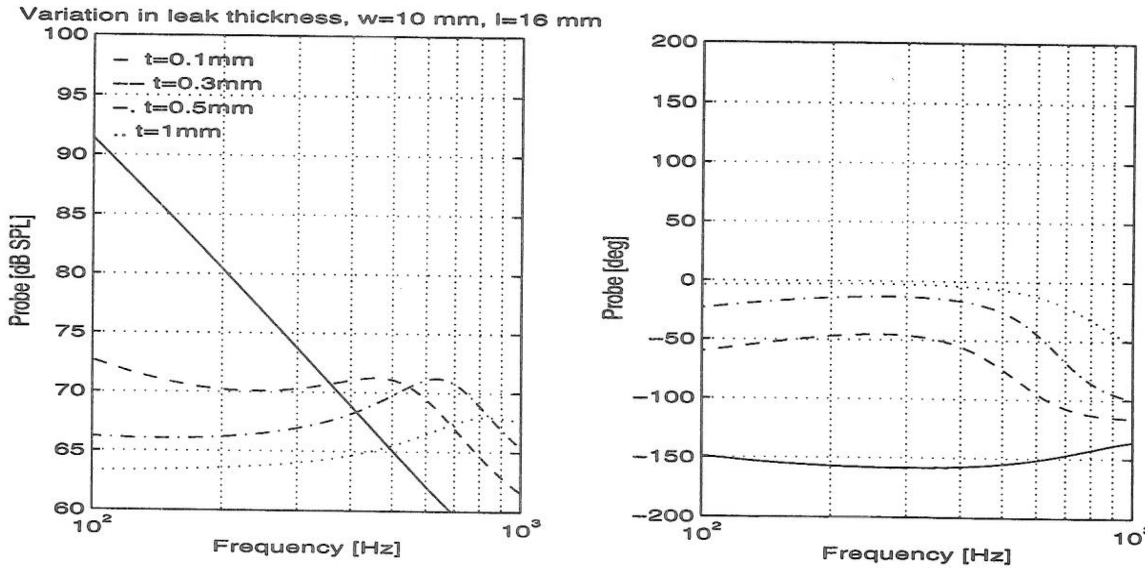


Figure 9.5 Sound pressure in the occluded ear 3 mm behind the earmould with variation of the slit leakage thickness. Width is 0.2mm and length=16mm. Input=60 dB SPL air conduction. See **Figure 9.11** for the value of the body source. Phase reference is a zero phase.

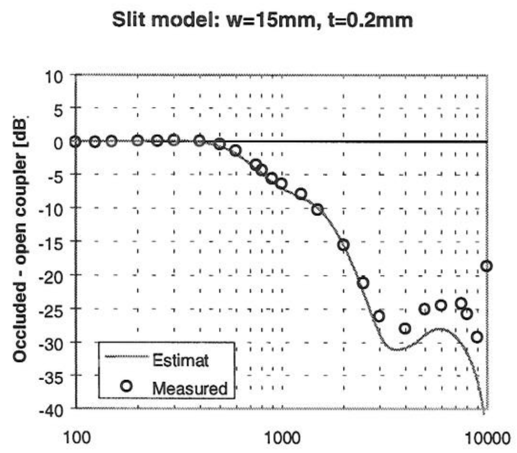
9.2 STATIC LEAKAGE

Static leakage occurs because the earmould is not fitted 100% tight in the ear canal. One way to demonstrate the presence of leakage is to compare the sound pressure in the open ear with the sound pressure in the occluded ear canal when the sound source is placed outside the ear. This method is used in clinical practice for hearing aid fitting to determine the amplification of the hearing aid. The difference between the sound pressure in the ear with hearing aid turned on and in the open ear is called *insertion gain.* Normally, the insertion gain is positive because the occluded ear pressure is higher than the open ear pressure due to the hearing aid amplification. When there is only a mould in the ear and no hearing aid, the occluded pressure is lower than the open ear pressure because the earmould attenuates the external sound and the difference is a negative gain, the so called an *insertion loss.*

9.2.1 Insertion loss in a coupler

Before moving on to real ears, the model estimate will be compared with the coupler measurements described in appendix G. The attenuation of a foam, an acrylic plug and a plug of impression material was measured in an ear simulator B&K 4157 (IEC-711) with an adapter (DB 2012). The coupler and adapter simulates the ear canal and the middle ear impedance. Three acrylic moulds were cast directly in the adapter with the same method used for making earmoulds to real ears. The most inner part of the coupler is tube shaped and the adapter is cone shaped. In a mathematical model of the coupler, the whole open coupler is modeled as a uniform tube of length 23.5 mm and diameter 8.2 mm. The average diameter decreases when the acrylic plug is inserted 14 mm, and the occluded diameter is therefore 7.5 mm. The tube model was terminated with the middle ear impedance.

Leakage between the acrylic mould and the coupler will probably occur as many tiny slits, but of practical reasons, the leakage will be modeled as one slit only. In **Figure 9.6** the best fit of the model to the measured coupler insertion loss is shown, which is a slit of width 15 mm and thickness 0.2 mm.



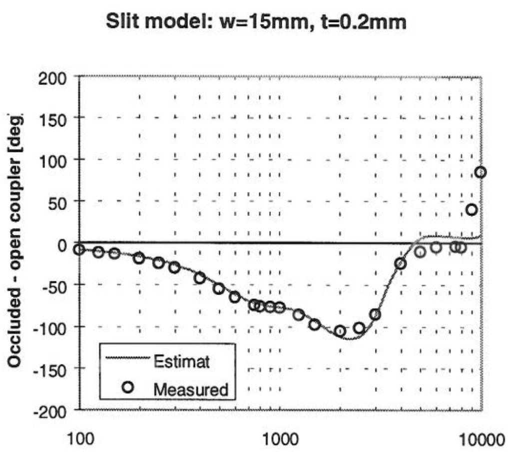


Figure 9.6. Estimated and measured difference between a closed coupler and an open coupler. The coupler was closed with an acrylic earmould. The estimate assumes a slit leakage 15 mm x 0.2 mm. Left: magnitude. Right: phase difference. Stimulus: loudspeaker.

9.2.2 Insertion loss in real ears

Figure 9.7 shows an example of insertion loss in a real ear. In this example a loudspeaker was placed 2 m in front of the subjects nose. The sound pressure in the occluded ear and the open ear was measured at the same time.

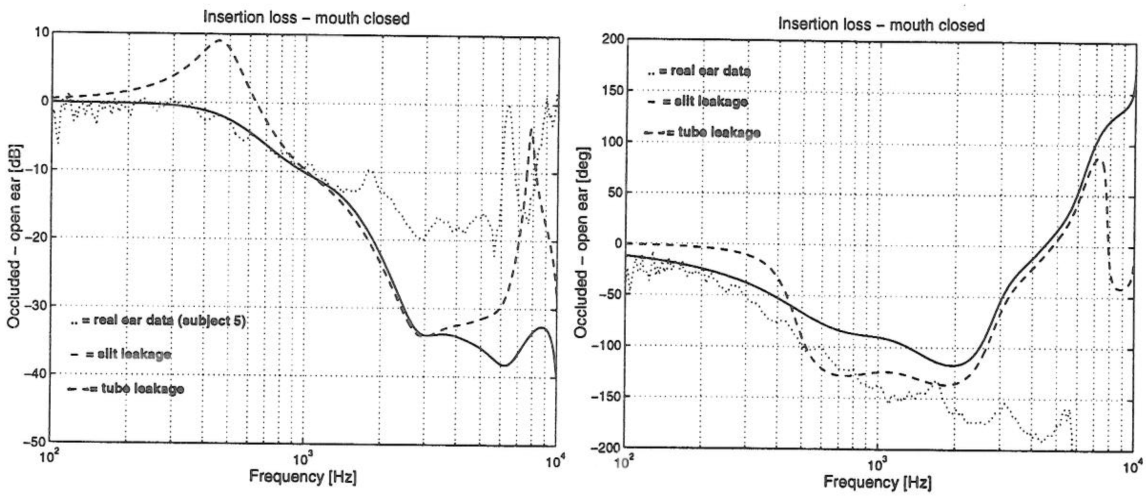


Figure 9.7. Difference between an occluded ear canal and an open ear canal. Real ear data: subject 5, right ear occluded with an acrylic mould. Slit: w=18mm, t=0.2mm. Tube: d=1.7mm. Leakage length = 21 mm. Left: magnitude. Right: wrapped phase.

The curve was measured in an anechoic room (the little one at Department of Acoustic Technology at the Technical University of Denmark). The insertion loss is negligible up to 300 Hz and only around 3 dB at 500 Hz is due to leakage. The acrylic material attenuates about 40 dB at all frequencies (see appendix G), so a tight mould should give an insertion loss of 40 dB at low frequencies too. The attenuation is a lot smaller when the earmoulds are inserted in real ears, partly because of leakage between the earmould and the ear canal wall and partly because of physiological noise (blood flow, heart beat etc.) in the occluded ear.

The real ear data is compared to estimated insertion loss with the slit model and the tube model respectively. It is clear that the slit model fits the real ear data better than the tube model.

The estimated data with the slit model corresponds well with the real ear data up to 1.5 kHz in the magnitude. The phase fit is poorer but decreases a little less than the real ear data. At 2 kHz, the estimated phase does not fit at all. Also the lumped element slit model is not adequate at mid- and high frequencies. The model does not fit the measured data above 1.5 kHz mainly because the insertion loss is not just a function of leakage, hence the model of insertion loss is too simple and may not work above 1 kHz or so. The sound pressure level in the ear canal is a combination of the direct sound through leakage, sound radiation from the ear canal walls, sound traveling through the earmould material, and sound generated by vibrations of the earmould. So besides leakage, the attenuation of the earmould inserted in an ear depends on the absorption in the earmould material, the inertia of the earmould, the friction between the earmould and the ear canal wall (the surface material of the earmould) and the body conducted sound. Furthermore, the slit model assumes that the slit walls are rigid, which is not true as one wall comprises the ear canal cartilage and absorbs more energy than a rigid wall. Notice, that in a small slit the absorption at the walls is important whereas in a tube with a size of the ear canal, the absorption becomes negligible.

These features are not so important when the occlusion effect has to be determined for the use of hearing aid design in relation to own voice. The most prevalent hearing loss is age-related and increases with frequency, the hearing aid will therefore amplify less at low frequencies and more at high frequencies, thus, the amplified sound overtakes the sound produced by the above mentioned features at higher frequencies (above 1 kHz, for example).

Regarding ear protectors, the situation is different. The body conducted sound is greatest at low frequencies, so predictions above 1 kHz must include features like the earmould material etc. This is especially important for singers and players of wind-instruments, for whom it is essential to perceive the whole range of audio- frequencies correctly.

9.3 DYNAMIC LEAKAGE

The term dynamic leakage will be addressed to leakage that occurs when the person produces a sound with the speech organ or the jaw. A sound produced though the nose and the closed mouth (nasal) might create a larger leakage than in the static case because the cartilage is moving. If the jaw also moves, the leakage will become even greater and the shape will change.

9.3.1 Ear canal dimension change

It was explained earlier that lowering the jaw changes the shape of the ear canal cross-section. Because of the elastic fibers in the cartilage, the cross-section gets larger, mostly in the anterior direction (towards the front), Oliveira [1995]. As described in chapter 6, Oliveira measured the ear canal shape on 6 subjects by the means of ear canal impressions. The diameter in the inferior-superior and anterior-posterior direction was measured between the first and second bend at the point where the change in both diameters was greatest. The cross-sectional shape changes were demonstrated by van Willigen [1976] who used a special impression paste to picture the ear canal cross section with the jaw in different positions on a number of subjects (the paper does not say how many). The shape of the cross-section changes, see **Figure 9.8**. An acrylic earmould cannot follow these shape changes and it is likely that if there was a slit leakage with a closed mouth, the slit would then become thicker - and less wide with an open mouth.

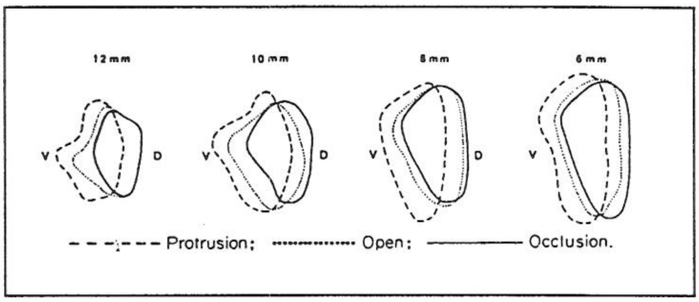


Figure 9.8 Ear canal cross-sections with closed mouth, protruded jaw and 25 mm jaw opening. The cross-sections were cut in planes 6-12 mm from the ear canal entrance. From [van Willigen, 1976].

Borghese et al. [1997] measured the jaw opening during speech to 8 mm, measured as the difference in 3D distance between tip of nose to chin. At present time they have presented primarily data only using one word:'to-til-tru-chi'. It is difficult to say precisely how large the average jaw opening is during running speech, so to get an operational value, the average jaw opening during the word 'to-til-tru-chi' is used and that is 4 mm.

9.4 A FOAM AND AN ACRYLIC MOULD

In chapter 8 it was shown that the occlusion effect is different between acrylic mould and a foam plug. One explanation could be that the leakage is different. The foam plug attenuates the air borne sound better than the acrylic mould does, because the foam plug can be fitted much tighter in the ear canal. A long narrow leakage between the ear canal wall and the earmould will reduce the occlusion effect at low frequencies, which of course is an advantage. If the leakage becomes thicker and wider, it might have the opposite effect. At very low frequencies, for example below 200 Hz, the occlusion effect will still be reduced but then it increases because of the acoustic resonance created by the leak, see **Figure 9.2.** Thus, it is possible that the observed difference between the foam plug and the acrylic mould is caused by leakage.

Insertion loss of the earmoulds were measured on each subject (appendix C). In by far the most cases, the foam plug attenuated better than the acrylic mould, see **Figure 9.9.**

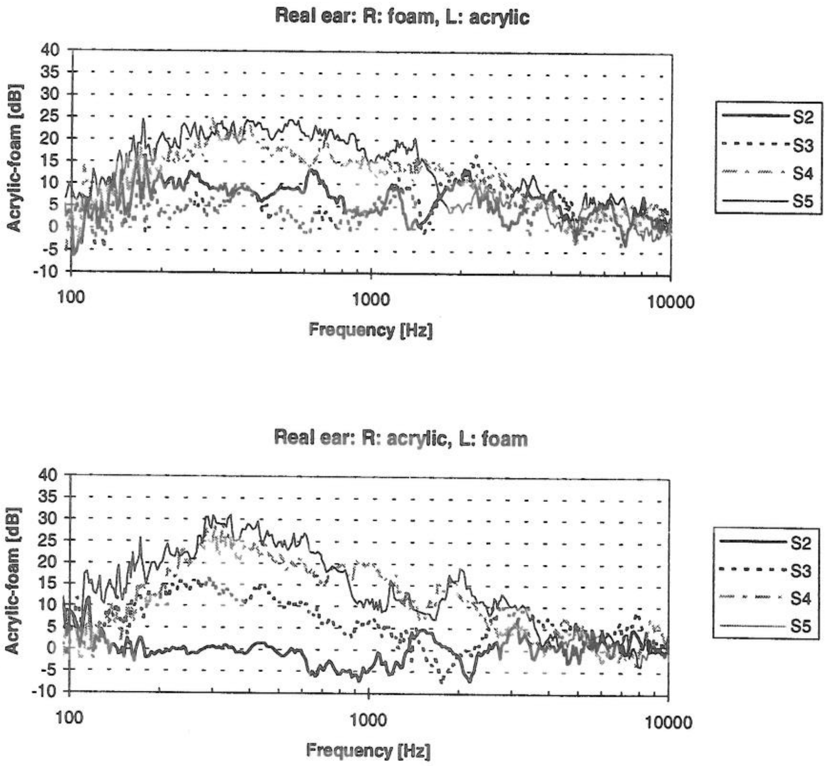


Figure 9.9. Difference between an acrylic mould and a foam plug. Measured simultaneously in right (R) and left (L) ear. Stimulus: loudspeaker multsine signal.

The measurements took place in an anechoic room where the subject was sitting in front of a loudspeaker generating a multisine signal and the sound pressure in the right and left ear was recorded simultaneously by the mean of two probe tube microphones. The probe tubes were inserted through the moulds to assure that the probe tube did not create a leakage.

Below 200 Hz the difference becomes smaller with frequency. Firstly, the foam plug is *not* inserted fully (as recommended by the manufacturer) so at very low frequencies the attenuation is not good and secondly, the anechoic room and the loudspeaker does not perform optimally below 200 Hz. Between 200-1000 Hz, subject 4 and 5 shows the greatest difference which is about 20-25 dB and then it decreases with frequency. Subject 3 shows a difference of 5 dB with the foam plug in the right ear and 10-15 dB with the foam plug in the left ear. Subject 2 has a difference of 10 dB with the foam in the right ear and nearly no difference with the foam in the left ear.

In **Figure 9.7** the static leakage for subject 5 was predicted for the acrylic mould. The foam mould is tightly fitted and it is reasonable to presume that there is no static leakage. **Figure 9.10** justifies this presumption at frequencies below 1 kHz. The measured insertion losses for all subjects, except for subject 2 with the foam plug in the left ear, are not much different from the insertion loss measured in a IEC-711 coupler, where there was, for sure, no leakage. The insertion loss with the foam plug was not measured directly. In order to find the insertion loss for a foam plug, first the difference between the insertion loss for the acrylic mould was measured and then the difference in sound pressure level in one ear occluded with an acrylic mould and the other occluded with a foam plug with an external sound source (loudspeaker) was measured, see **Figure 9.10**.

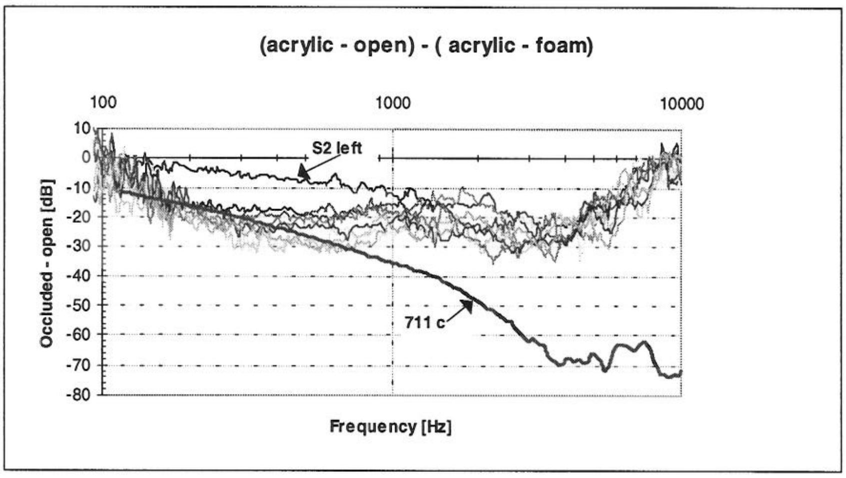


Figure 9.10 Difference between an occluded ear canal and an open ear canal. Four subjects, both ears and a IEC 711 coupler. Foam plug.

In conclusion, the foam plug can be considered as very tightly fitted in the static situation, whereas the acrylic mould is not tight. Therefore, it is reasonable to assume that the leakage for the acrylic mould will be larger than the foam plug also in the dynamic situation. An important point to make is that the difference between the foam plug and the acrylic mould is about the same for nasal sound, continuous speech and to some extent, chewing. Hence, the difference is *independent* on jaw movements. The most likely explanation is that the leakage gets larger for both the foam plug and the acrylic mould when the jaw moves and that the leakage still creates a difference in the sound pressure in the ear canal equal to the situation when the jaw does not move.

The static leakage in **Figure 9.9** shows that the subjects behave very differently and it can only be concluded that the occluded sound pressure is smaller with a foam plug than with an acrylic mould. It is more informative to analyze each subject individually. We will look at 2 examples; subject 5 and subject 2, because they describe the most extreme differences in occlusion effect.

9.4.1.1 Subject 5

The occlusion effect calculated from the occlusion model with an acrylic and a foam plug is shown in **Figure 9.11**. For comparison, the estimation with a dynamic leakage is shown together with the static leakage and it is seen that the dynamic leakage is clearly different from the static leakage. According to **Figure 9.7** the static leakage is about 18 mm wide and 0.2 mm thick. The right panel in **Figure 9.11** illustrates that according to these theoretical estimations, the foam plug also has a dynamic leakage. The measured occlusion effect for the foam plug is not measured directly but calculated in the same way as the insertion loss curve in **Figure 9.10**.

The total effect of leakage on the occlusion effect from own voice is not simple to estimate. It is necessary to make an assumption about the relative contributions from the air transmitted sound and the body conducted sound to the overall sound pressure. For reasons that will be described in chapter 10, it is reasonable to presume that, in the open ear, the body conducted sound is about 10 dB lower than the sound due to air transmission.

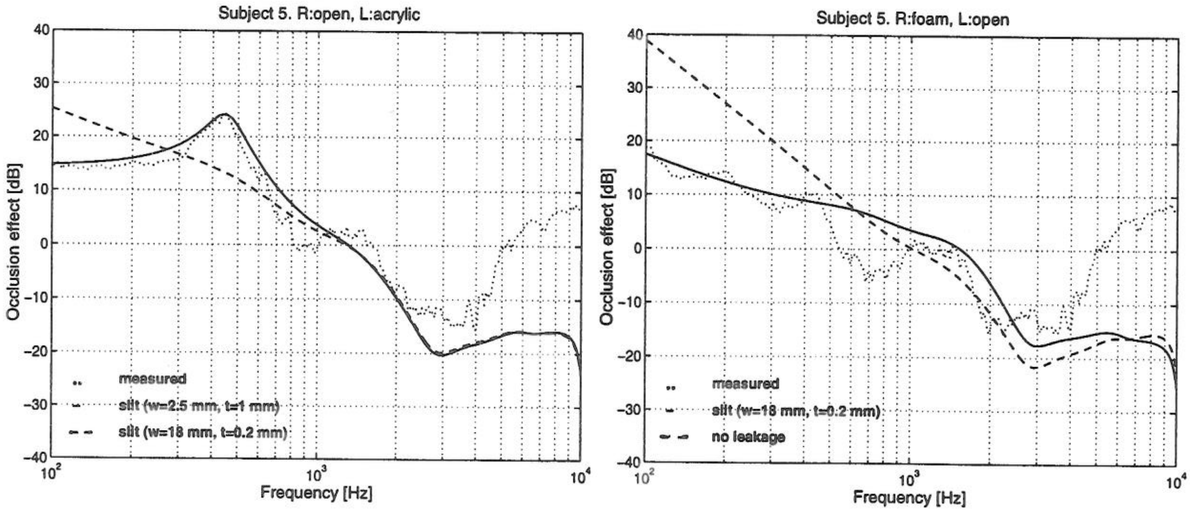


Figure 9.11 Measured and estimated occlusion effects. Subject 5. Left panel: Right ear open, left ear occluded with an acrylic mould. Slit leakage length = 21 mm. Right panel: Right ear occluded with foam plug, left ear open. Slit leakage length = 8 mm. Default model values. Own voice, speech. The body conducted sound is assumed to create a sound pressure in the open ear that is 10 dB lower than the sound pressure created by air borne sound from own voice. Left: Magnitude Right: Phase.

The estimated pressure in the occluded ear with an acrylic mould and a foam plug respectively is compared to the simultaneous measured difference. As shown in **Figure 9.12** the estimation agrees well with the measured data and this indicates that the difference might occur because the dynamic leakage is different with the acrylic and with foam plug fittings.

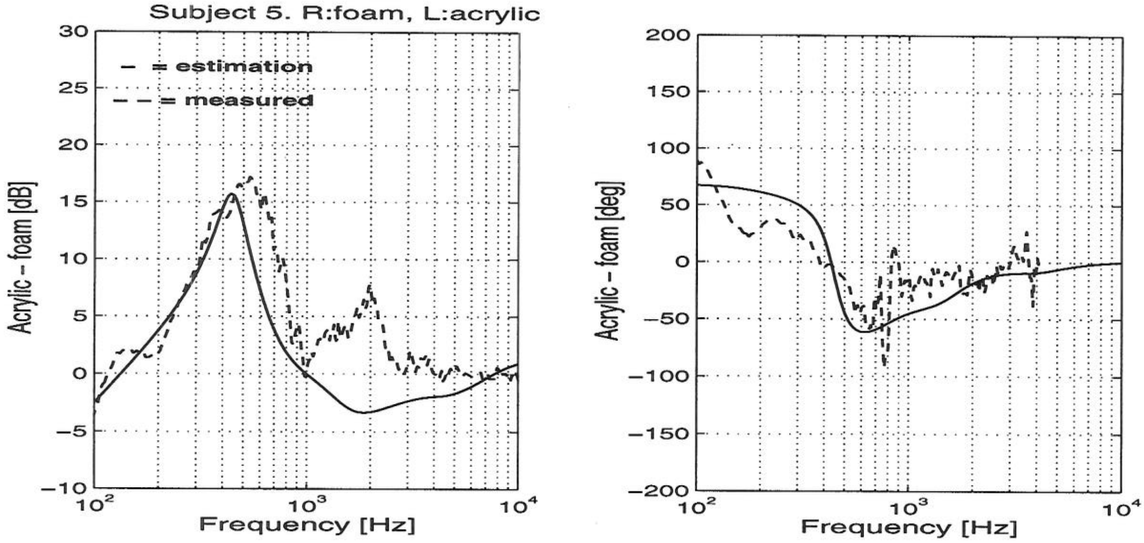


Figure 9.12 Measured and estimated difference in sound pressure level in left ear occluded with an acrylic mould and right ear occluded with a foam plug. Acrylic; slit: w=2.5 mm, t=1 mm, l=21 mm. Foam; slit: w=18 mm, t= 0.2 mm, l=8 mm. Subject 5. Own voice, speech.

9.4.1.2 Subject 2

Subject 2 has a more flat difference between the ear occluded with the foam plug and the acrylic mould. The best fit of the leakage model to the measured data is shown in **Figure 9.13** and it is clear that the estimate does not agree with the measured data.

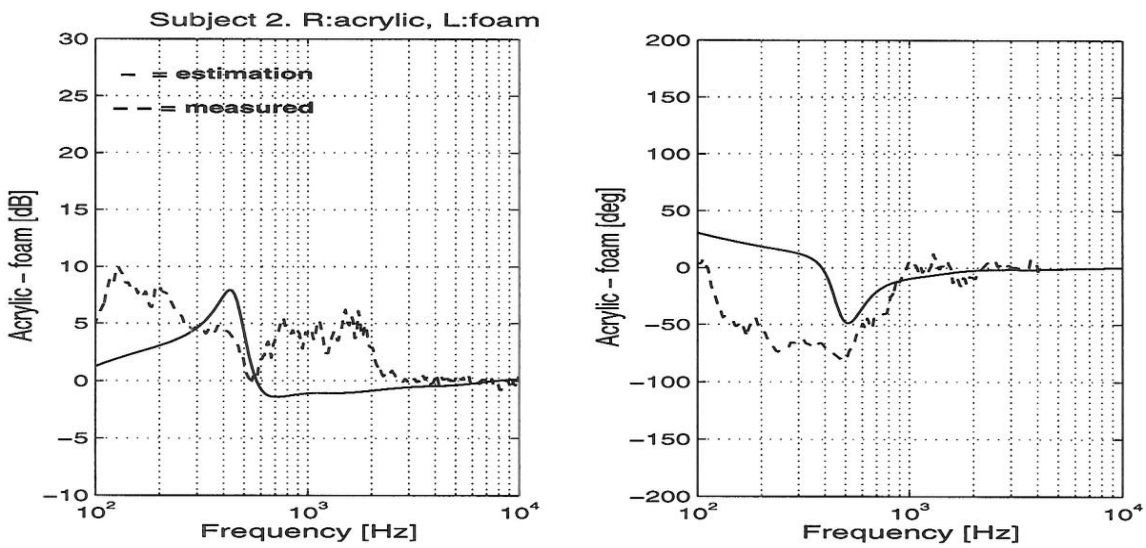


Figure 9.13 Measured and estimated difference in ear sound pressure level in right ear occluded with an acrylic mould and left ear occluded with a foam plug. Acrylic; slit: w=2.8 mm, t=1.3 mm, l=30 mm. Foam; slit: w=3 mm, t= 0.5 mm, l=8 mm. Subject 2. Own voice, speech.

Subject 2's data were estimated with the same procedure as used for subject 5. The problem is that the occlusion effect for subject 2 with a foam plug cannot be estimated very well with the simple model consisting of a uniform tube and the mould as a rigid wall with some slit leakage. This is illustrated in **Figure 9.14** where the acrylic plug estimation is acceptable, but the foam plug is not. For the sake of illustration, estimations for 3 different slit sizes are shown.

but the foam plug is not. For the sake of illustration, estimations for 3 different slit sizes are shown.

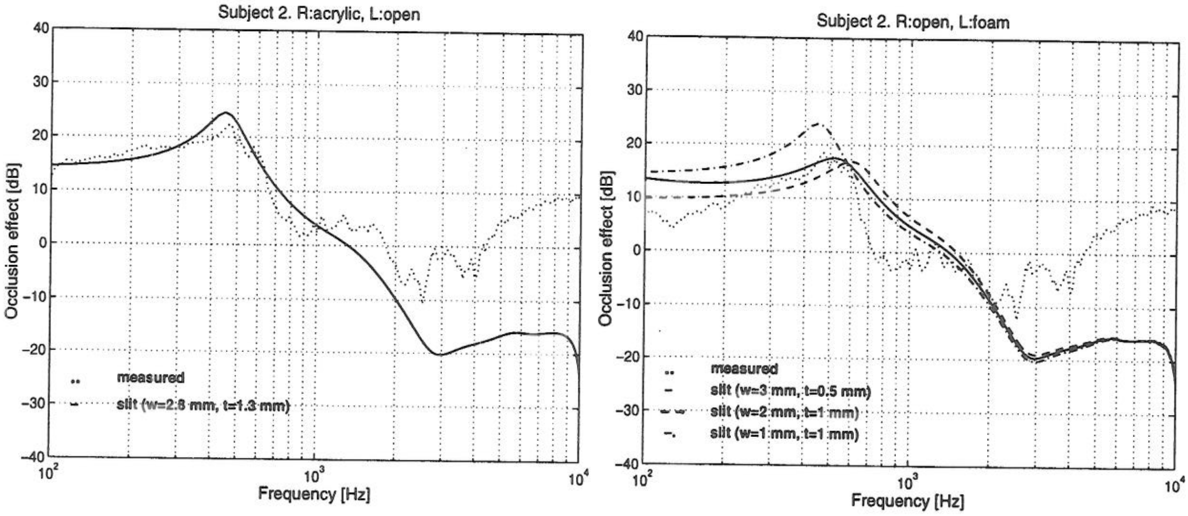


Figure 9.14 Measured and estimated occlusion effects. Subject 2. Left panel: left ear open, right ear occluded with an acrylic mould. Slit leakage length = 30 mm. Right panel: left ear occluded with foam plug, right ear open. Slit leakage length = 8 mm. Default model values. Own voice, speech.

9.5 CONCLUDING REMARKS

It was shown that a slit model of the leakage fits better to the real situation than a tube shaped leakage model. The previous analyses of leakage shows that the physical shape of the static leakage is different than the physical shape of the dynamic leakage. The dynamic leakage is greater than the static leakage. The observed differences with a foam plug and an acrylic mould can be explained by the effect of different leakage in some cases, but not in all cases. The occlusion model with a leakage shaped as one single slit and ignoring the earmould properties seems to be too simple to estimate the occlusion effect for all subjects.

10. RATIO OF AIR BORNE AND BODY CONDUCTED SOUND

In the previous chapters the body conducted sound source was set to produce a sound pressure in the open that was 10 dB lower than the sound pressure produced by the air conducted sound from own voice. This chapter describes why it is so.

When we speak, sound reaches our ears both through the air, from mouth to ear, and through our body, as vibrations. The mathematical model includes both sound sources, but the relationship between the two was not known a priori. In previous models in the literature of the occlusion effect, only one sound source is present at a time. Own voice complicates the situation because both sound sources naturally are present at the same time. Especially with leaky fitted moulds, such as standard hearing aid moulds, it is important to know the relationship between the airborne and body conducted sound, p_c/p_a. In the occluded case, air borne sound will enter the ear canal through the leak, and body conducted sound will escape through the leak. It is important to address the perception of own voice in order to fully understand the experience of the occlusion effect. The ratio between the air borne and body conducted sound has been studied in a pilot experiment followed up by a master thesis project. A model of occlusion effect will be used in the following. The model is based on hypothesis 1 in chapter 7, where the earmould is regarded as a rigid wall. This is done in order to minimize the variable parameters in the model.

10.1 INTRODUCTION

The hearing organ detects both environmental sounds and the sounds we produce with our own body. The total sound pressure in the ear canal created by our own voice is thought to be mainly a sum of two components: An air conducted signal, p_a, propagates from mouth through the air to the ear while a body conducted, p_c, component propagates as vibrations from the larynx through the neck and skull to the soft cartilage of the ear canal, where it radiates into the canal, see **Figure 10.1**. For more information see chapter 4 for speech production, see chapter 5 for the anatomy of the ear, and see chapter 6 for body conduction.

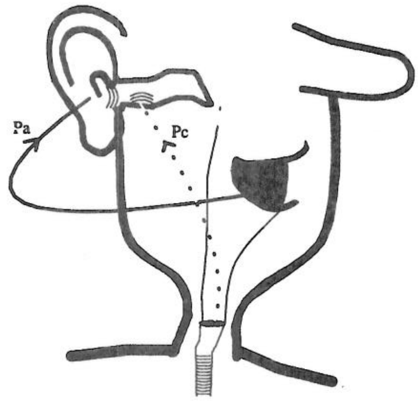


Figure 10.1 Sketch of a human head showing the air conduction (Pa) and body conduction (Pc) pathways. In the following Pc is measured about 12.5 mm from the eardrum.

Although the occlusion effect has been measured by various authors, additional knowledge of the air conducted and body conducted components is required to help understand and to model the occlusion effect. An independent measure that is particularly pertinent is the relative magnitude of the body conducted sound to the air conducted component.

The literature provides little real ear data on body conducted sound produced by one's own voice. The most relevant work is that of Békésy [1949]. Békésy tried to eliminate the air conducted sound by means of large tubes placed on the ears. These tubes attenuated the air conducted sound by 30 dB between 150 Hz and 4 kHz, without causing an occlusion effect. The difference in loudness of short words between open ears and covered ears was measured using the subject's own voice. The loudness of short words was compared to the loudness of a 1 kHz tone produced by a bone-conductor on the forehead. Békésy found that the body conducted component was 0 dB to 10 dB lower than the air conducted component, depending on the sound produced. The experiment was performed on 4 subjects.

This experiment needs some comments. First of all the ratio was only measured indirectly, no real ear pressures being determined. Assuming that the loudness measurements are done using standard methods, tissue and bone act as a lowpass filter and the voice will therefore sound deeper when only the body conducted sound is heard. The difference in loudness between open and covered ear is therefore not necessarily a measure for the loudness at the same frequencies. Secondly, a reference microphone was not used to assure that the subjects spoke with the same level in the two situations. The present author [1997] has performed an experiment showing that people automatically change their speech level when the ear is occluded. When the ear is covered by a device that does not give occlusion effect it must be anticipated that subjects still raise their own voice by a few dB. This phenomena means that the real ratio is a few dB smaller than that measured by Békésy.

It was therefore decided to perform a pilot experiment in order to get an estimate of the ratio between air and body conducted voice sounds, making use of the probe tube technology that has developed since the time of Békésy. But first some theoretical considerations and estimations will be made.

10.2 THEORETICAL CONSIDERATIONS

In a visco-elastic medium (which could be a model of human tissue) sound waves will be transmitted best at low frequencies. However, it does not mean that the ratio p_c/p_a decreases constantly with frequency, because the air borne sound entering the ear canal also decreases with frequency above 3 kHz. The ratios found by Békésy [1949] might immediately seem quite small because one tends to compare the body conducted sound pressure with the air borne sound pressure that one hears when another person speaks. The actual air borne sound pressure to compare is the source pressure in the back of the oral cavity. Békésy [1949] who measured the sound pressure from the inside of the mouth to the outside and compared that to a reference microphone placed 1 cm out from the lips. On average the sound pressure at the lips is **15 dB** lower than in the back of the mouth cavity.

The radiation from the mouth decreases with about 6 dB/octave and furthermore the sounds are 'filtered' from the mouth to the ear around the head. For example on a Head-and-Torso-Simulator the sound pressure drops 32 dB from about 3 to 7 kHz, [Brüel&Kjær,

1989]. The air borne sound is reduced with 18 dB from the mouth to the eardrum in the frequency range 100-1000 Hz, which is the same order of magnitude found by Békésy.

In conclusion the air borne sound from the back of the oral cavity to the ear canal entrance is damped about 35 dB below 1 kHz. So, even when a ratio of *pc/pa* -10 to -15 dB seems to be a rather small, it has to be remembered that the air borne sound from own voice is damped significantly from the back of the mouth cavity, out of the mouth, around the head and into the eardrum.

The ratio between the air borne sound and body conducted sound in **Figure 10.1** can be understood as a transfer function. The transfer function from the vocal source to the ear canal of air conducted sound can be written as a series of transfer functions:

$$p_a(f) = S(f)_{\text{vocal source}} T(f)_{\text{vocal tract}} R(f)_{\text{oral cavity to mouth and nose radiation}} HRTF(f)\, E^{e\text{-}d}(f)_{\text{ear canal entrance to eardrum}}$$

The body conducted sound is transmitted to the ear through the body (BRTF) via:

$$p_c(f) = S(f)_{\text{vocal source}} T(f)_{\text{vocal tract}}\ BRTF(f) E^{x\text{-}d}(f,x)_{\text{point x in ear canal to eardrum}}$$

where;
HRTF = head related transfer function, sound from mouth around the head to the ear
BRTF= body related transfer function, the sound from the vocal tract to the ear canal wall

The ratio then becomes:

$$\frac{p_c}{p_a} = \frac{BRTF \cdot E^{x-d}}{R \cdot HRTF \cdot E^{e-d}} \qquad (10.1)$$

The transfer function for different points at the ear canal has already been calculated, see for example appendix A. R is accessible in the literature and HRTF is well known. But the BRTF is not known. Sound transmission in muscles and soft tissue at the audio frequencies has not been given much attention. In the medical services ultrasound frequencies are more interesting. In relation to bone-anchored hearing aids some measurements of the point skull impedance has been made, e.g. Håkansson and Carlsson, [1986]. But here the bone is directly stimulated and the results cannot be applied directly to speech production. Ishizaka et al.[16] measured the mechanical impedance on the cheek and neck in order to estimate the impedance of the vocal tract walls. Unfortunately they only measured up to 160 Hz. The impedance of the skin and soft tissue has been measured as a point impedance in most cases on the finger or the arm, [Gierke and Brammer, 1995]. None of these data provide enough information to be able to make a reasonable estimate of p_c/p_a.

10.3 ESTIMATING PC/PA

The body related transfer function BRTF is unknown so another approach will be used. During speech with the ear canal open, the sound pressure in the canal is a sum of an air borne component p_a and a component p_c passed through the body, see **Figure 10.1** The available equipment could not measure the phase so to begin with it was therefore assumed that these components were uncorrected, the total pressure is then $(p_a^2 + p_c^2)^{1/2}$. In the following, we consider the effects of blocking the air borne path, both in producing the occlusion effect and as a means for establishing the relative magnitude of the two components.

10.3.1 The Occlusion Effect

The occlusion effect, OE, is calculated as the difference in sound pressure level (11.8 mm) in the ear canal between the closed ear and open ear condition, when the canal is blocked with an earplug or hearing aid mould. With the ear closed, the air conducted component is reduced by the attenuation of the mould giving a sound pressure αp_a, where α is a factor due to attenuation of the inserted earmould. On the other hand, closing the ear canal tends to increase the sound pressure due to the body conducted component, at least at lower frequencies. The acoustic load presented by the closed canal is higher than the load for the open canal and with an effective volume velocity source, higher sound levels are anticipated. With the canal blocked, the sound pressure due to body conducted sound is βp_c, where β is an occlusion gain factor. The occlusion effect ratio, *oe,* can then be calculated as:

$$oe^2 = \frac{p_{closed}^2}{p_{open}^2} = \frac{\alpha^2 p_a^2 + \beta^2 p_c^2}{p_a^2 + p_c^2} \qquad (10.2\ )$$

Some measurements of the occlusion effect are shown in **Figure 10.2.** For these, Wimmer [1986] and Thorup [1996] both inserted the probe tube through a hole in the earmould, avoiding leakage. May and Dillon [1992] placed the probe tube between the ear canal wall and earmould.

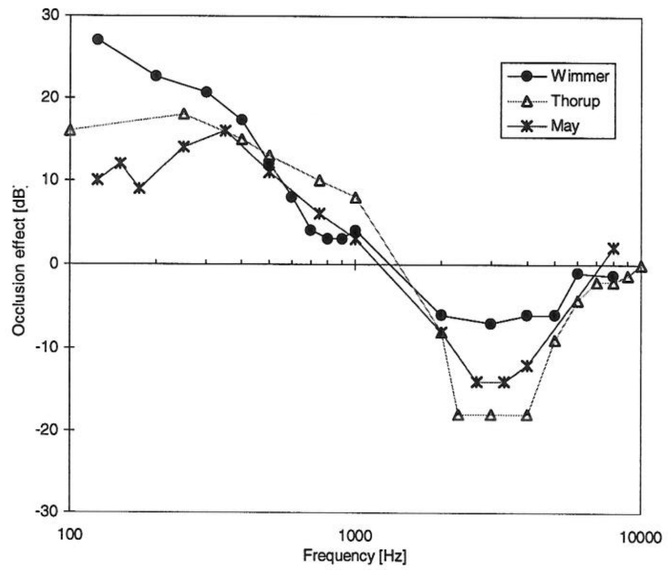


Figure 10.2. Occlusion effects, own speech.. Data from: Wimmer [1986], Thorup [1996] and May [1992]

The β factor has not been measured previously. As seen from equation (10.2), it is a quantity that can be derived once measurements of both *OE* and p_c/p_a are available. It is possible, though, to estimate β theoretically if assumptions are made about how the body conduction is taking place. Initial calculations show that $\beta \gg 1$ and if the earmould attenuates enough so that $\alpha \ll 1$, then β can be found by:

conduction is taking place. Initial calculations show that $\beta \gg 1$ and if the earmould attenuates enough so that $\alpha << 1$, then β can be found by:

$$oe^2 \cong \frac{\beta^2 (p_c / p_a)^2}{1 + (p_c / p_a)^2} \Leftrightarrow \qquad , \beta^2 > oe^2 \qquad (10.3)$$
$$(p_c / p_a)^2 \cong \frac{oe^2}{\beta^2 - oe^2}$$

The air conducted and body conducted sound has been assumed to be uncorrelated in order to make some estimates. However, p_c and p_a are highly correlated but it has not been possible to find any reports on measurement of the phase created by vocalization, although some data on phase differences of signals traveling from one ear to the other and sound velocity in the skull has been reported and will be commented in section 10.7.2.

10.3.2 Procedure to Measure p_c/p_a

The basic idea of the experiment was to eliminate the air borne sound so much that the sound pressure in the ear canal could be regarded as due to body conducted sound only. If the air borne component can be reduced by a factor γ, without causing an occlusion effect, then, the pressure with the attenuation box in place p_{boxed} is given by:

$$\frac{p_{boxed}^2}{p_{open}^2} = \frac{\gamma^2 p_a^2 + p_c^2}{p_a^2 + p_c^2} \qquad (10.4)$$

Now, if $\gamma \ll 1$, p_c/p_a becomes:

$$(p_c / p_a)^2 \cong \frac{p_{boxed}^2}{p_{open}^2 - p_{boxed}^2} \qquad (10.5)$$

so a measurement of p_{boxed} and p_{open} yields the desired ratio p_c/p_a.

Several methods of how to eliminate p_a were discussed. A device to put over the mouth was tried but it was difficult to make it tight enough because the mouth and jaw are moving during speech. Finally, it was decided to use a technique similar to that used by Békésy [1949] except for two main differences. First, the ear canal sound pressures are measured directly, using probe microphones. Secondly, an attenuation box, rather than large tubes, is used to reduce the air conduction component.

The phase difference could not be measured in this pilot study because p_{boxed} and p_{open} could not be measured simultaneously.

10.4 MEASUREMENT METHOD

This section describes the set up and equipment used in the pilot study. The three main issues are the design of the box, precision of the reference microphone position, and influence of the cushion pressure.

10.4.1 Set Up and Calibration

The test set up is sketched in **Figure 10.3**. The subjects were seated in a chair in an anechoic room. The ear canal sound pressure was measured with a probe microphone used for standard clinical equipment. The reference signal was picked up near the opposite ear with a ½" Brüel&Kjaer microphone, Type 4134. The probe microphone signal was amplified with a Stanford Research System amplifier Model SR640. The reference signal was amplified in a Brüel&Kjaer Measuring Amplifier, Type 2610. Both signals were lowpass-filtered with a Stanford Research System amplifier Model SR640 at 9 kHz in order to avoid aliasing. The signals were recorded onto DAT-tape with a sampling frequency of 44.1 kHz, which was later down sampled to 2,050 Hz before analyzing the signals in the program 'Matlab'. The supervisor could check the signals via an external loudspeaker and a scope (not showed on the figure).

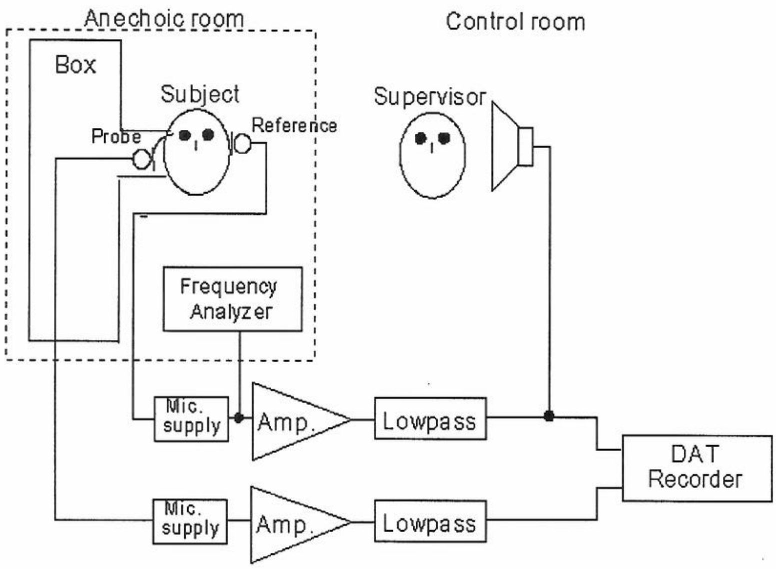


Figure 10.3. Test set up

Figure 10.3 shows the subject attached to the attenuating box. In the free field situation, the box was taken out of the anechoic room. The sound pressure in the ear canal was measured with a standard probe microphone with a soft flexible probe tube. The probe tube was inserted into the subjects ear canal to a depth corresponding to a standard earmould (about 8 mm + 3 mm). As only one probe microphone was available, it was necessary for the subject to repeat the same sound twice. Continuous speech (10 Harvard sentences) and the cardinal vowels /i/, /a/ and /u/ were used. The challenge here was to maintain identical conditions (formant frequencies and levels) for the two cases, with and without the attenuating box in place. The subject could look at the monitor of a real time frequency analyzer, where the fundamental frequency of the first pronounced vowel was stored and the subject then had to hit the same fundamental frequency again. A successful reproduction required a few tries every time.

10.4.2 Box Design

The box was made of 2 cm thick plywood panels with the dimensions: height=104 cm, width=54 cm and depth=54 cm. The outside and inside of the box were covered with 10.16 cm (4") fiberglass in order to minimize reflections. A hole was cut in the front panel and covered with a removable rectangular Plexiglas duct installed. The Plexiglas tube was 15.5 cm x 15.5 cm with a depth of 10.5 cm. A doughnut cushion was attached on the front plate. The cushion was placed over the subject's ear and the plate was attached to the head with Velcro straps. It was then easy to put the probe microphone in the ear canal and buckle the plate to the box. It was assured that the plate was attached tightly to the box. The box attenuates **more than 25 dB** for frequencies between 300 Hz and 3 kHz.

The signals picked up by the reference microphone in the free field conditions were compared to those with the box in place. At frequencies below 1 kHz, the disturbance from the box was in the order of the measurement error and the rms-level is the same for the reference microphone with the box in place and in the empty room. The cushion was fit around the pinna, so the effect of the pinna on the radiated sound is the same as for an uncovered ear.

10.4.2.1 Pressure from the Cushion

The subject had to lean against the cushion to avoid leakage between the head and the cushion. When the subject presses against the cushion, there is a risk though that the mechanical motion of the bones and tissue will be different than if the head was not attached to the cushion.

The effect of the cushion pressure was measured with the subject sitting in the anechoic room. The front plate was detached from the box and strapped on the subject's head. The sound pressure in the ear canal was measured while the subject read aloud with the front plate very loosely attached, and with the plate very tightly attached. The difference in sound pressure between a loosely attached and a tight attached plate is a measure for the effect of cushion pressure.

If the ear canal is open, the difference is 2.5 dB or less below 2 kHz. This is within the range of repeatability error. Hence, the effect of the cushion pressure can be ignored in the open ear.

10.5 RESULTS

10.5.1 Continuous Speech

The long-term power spectrum of continuous speech was calculated with the Welch method, see appendix E. An overlap of 50% was used.

The boxed ear and free field ear measurements were not measured simultaneously and the subject may have spoken with a different rms-level in the two situations. Therefore it was necessary to normalize the boxed ear sound pressure level with respect to the free ear level, using the difference in sound power of the reference microphone signals. The signals have been normalized using the power ratio, given as:

$$\text{ratio} = \frac{1/2\pi \sum S_{\text{boxed}}(n)}{1/2\pi \sum S_{\text{freefield}}(n)} \qquad (10.6\,)$$

where the spectra S is obtained from the reference microphone.

With the above normalization, the ratio between the sound pressure in the open ear and in the boxed ear was calculated using equation (10.5). The ratio p_c/p_a is shown in **Figure 10.4** for the left and right ear of the subject who participated in this pilot experiment.

10.5.2 Continuous speech

The ratio p_c/p_a estimated according to equation (10.5) is shown in **Figure 10.4**.

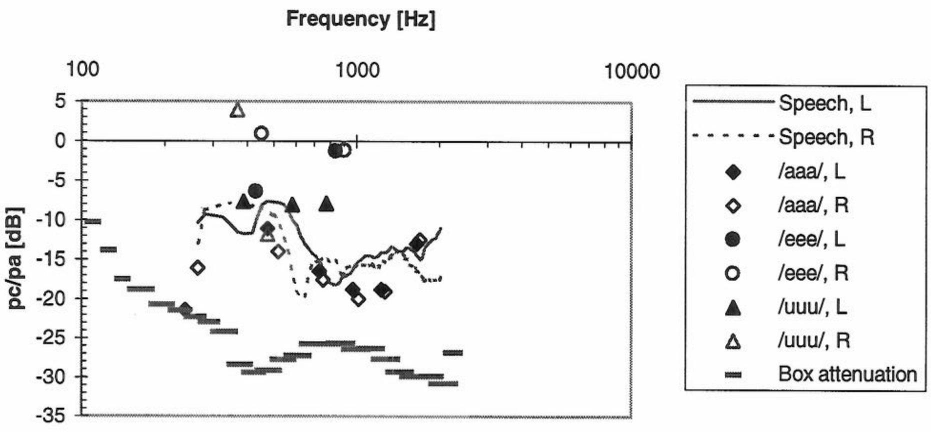


Figure 10.4 Ratio of boxed ear to free ear levels produced by speech and vowels. Speech 1/3 octave smoothed.

The results are printed from 200 Hz since the data is not reliable at lower frequencies because the room is not anechoic below 200 Hz and the box does not seem to attenuate enough. The vowels are printed up to that frequency where the harmonics seemed to agree reasonably between the two repetitions. In general the vowels and continuous speech agree well. The spectra for the right and left ears are similar but not exactly the same. It could be due to anatomical asymmetry and uncertainty in the measurements.

The levels were also reproduced. It is known that the speech spectrum changes when a voice is softer or louder than the normal speech level, [Elberling and Nielsen, 1993]. But in the pilot experiment a soft, a normal and a loud level vowel did not give a consistent change in the ratio between the sound pressure in the boxed ear and the free ear. This indicates that the ratio of body to air conducted sound is level independent as long as the sounds are produced in the same way, e.g. whispering might not give the same result.

Also shown in **Figure 10.4** is the measured box attenuation. Between 250 Hz and 2 kHz, the box attenuation is in the worst case only 6 dB below the pressure ratio (/aaa/ right). If the box attenuation is included in the computing, the ratio is 1.3 dB higher. From 300-2000 Hz the maximal calculation error was found to be 0.8 dB when it was assumed that $\gamma \ll 1$.

10.5.2.1 Comparison with the literature

Continuous speech will be compared to estimations of the ratio p_c/p_a by the mean of the occlusion effect data found in the literature. If β can be estimated, equation (10.3) is used to transform the occlusion effect data of **Figure 10.2** to give predicted ratios p_c/p_a. A rough estimate for β is derived from the model of the open ear and closed ear, see equation (10.7) when only body conduction is present, β is the ratio between the impedance of the occluded ear, Z_{occ}, and open ear, Z_{open}, as seen from the inside of the ear, see also chapter 1:

$$\beta = \frac{p_{occ,c}}{p_{open,c}} = \frac{q_c Z_{occ}}{q_c Z_{open}} = \frac{Z_{occ}}{Z_{open}} \qquad (10.7)$$

where;
subscript '*occ,c*' = occluded ear canal, body conducted source
subscript ' *open,c*' = open ear canal, body conducted source

So the β factor is independent on p_c/p_a . The estimate uses the values: ear canal length = 23.5 mm, diameter = 7.1 mm, insertion depth = 7.8 mm, probe located at earmould. The very simple model is used where the earmould is regarded as a rigid wall and the body sound source depends linearly on the tissue that is not covered by the earmould. Therefore a gain factor for the volume velocity source, gain = 1-(ins.depth/ear canal length), was introduced.

Data from Wimmer, [1986], and Thorup, [1996], was measured with very tightly fitted earmoulds, but the data from May and Dillon, [1992] was measured with standard earmoulds and the effect of leakage between the ear canal wall and the earmould has been accounted for in the estimations (slit leakage length= 21.5mm, width = 13mm, thickness = 0.35mm)[1]. These estimates are consistent with our measurements. Even if attempts were made to make the earmoulds by Wimmer and Thorup really tight, there will always be a small leakage when the jaw moves e.g. during speech and the earmoulds are made of hard materials. Hence, the estimated β values for Wimmer and Thorup are too large because they assume that there is no leakage. These values are used anyway for the sake of comparison. The computed ratios are compared with those measured in **Figure 10.5**. The data from May provides the best fit and the Thorup and Wimmer data lie as expected a few dB below the measured ratio at f < 800 Hz.

[1] The size of the slit was estimated by fitting a leakage model to measured data

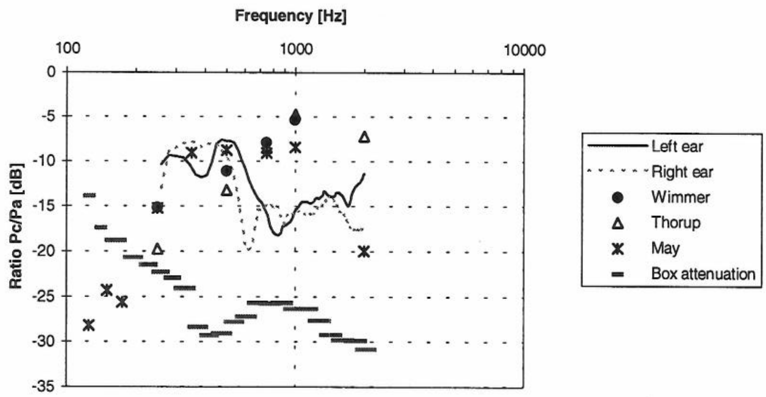


Figure 10.5 Ratio of body conducted and air borne sound. Continuous speech. Measured ratios are compared with estimated values derived from literature data.

10.5.3 Vowels

The ratio p_c/p_a was measured for the three cardinal vowels /a/, /i/ and /u/. Each vowel was measured three times. The subject was trained to reproduce the fundamental frequency within 30 Hz precision, but it was not always possible to reproduce higher harmonics so well. The difference between the boxed ear and the free ear spectra had to be compared harmonics by harmonics. It is rarely possible just to compare the spectra without any manipulation. Appendix E tells more about how this was done. Vowels are periodic signals and the assumptions for using equation (10.5) does not hold for vowels. The best estimate is to take the ratio between the boxed ear pressure and the open ear pressure.

The ratio for the 3 cardinal vowels are printed in **Figure 10.6**. It was only possible to reproduce up to the third harmonics acceptably well for normal as well as soft and loud phonation level. The fundamental of /uuu/ is not shown because it is located below 200 Hz where the box attenuates to little and the anechoic room was not free from reflections.

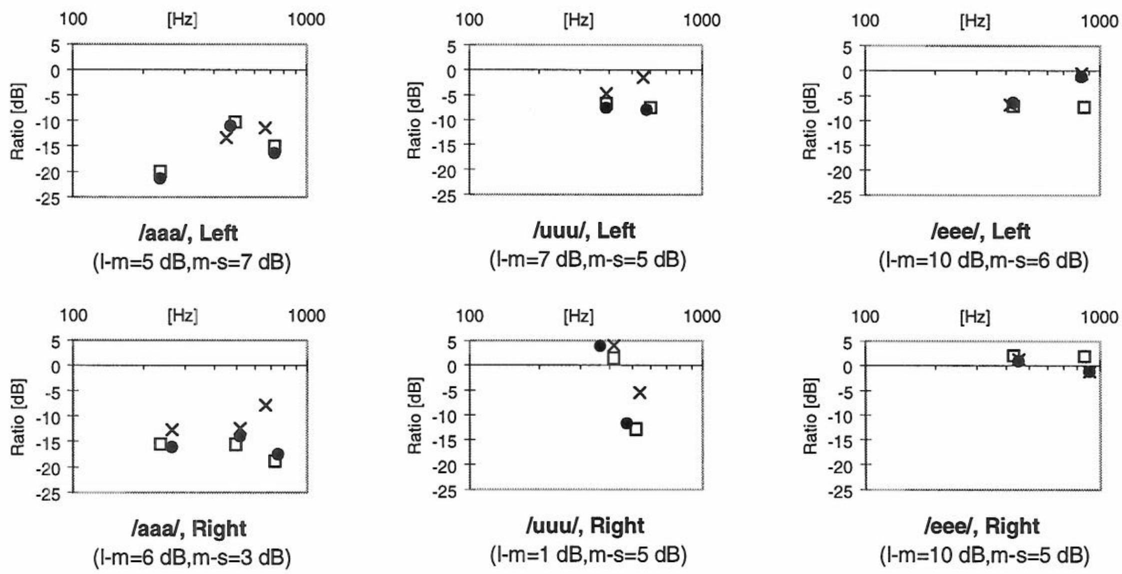


Figure 10.6 Ratio of body conducted and air borne sound. Sustained vowels: = loud, • = normal level and × = soft,. The bottom text lines show the fundamental level difference for the reference microphone in dB (l-m) = between loud and normal, (m-s) = between normal and soft.

10.5.3.1 Comparison with the literature

As mentioned in the introduction, Békésy [1949] measured ratio p_c/p_a by the mean of an equal loudness method. In the Békésy experiments many sustained vowels were used. In **Figure 10.7** shows the three vowels /aaa/ as in father, /uuu/ as in pool and /eee/ as in team, these are the same vowels used in the pilot experiment and should be compared with **Figure 10.6**.

Be aware that Békésy used a subjective method whereas the data in **Figure 10.6** was measured with an objective method (apart from using the subjects own voice). Even with the two different methods, the results agree that the ratio p_c/p_a of vowels lies generally within a range from -20 dB to 3 dB and that the ratio is smallest for the vowel /aaa/ and largest for the vowel /eee/.

Békésy explains that the ratio is lower for an /aaa/ sound because the mouth is fairly open and a lot of sound can be radiated. In contrary to /eee/ where the mouth is not so open and less sound is radiated, see also chapter 4 for the shape of the mouth and oral cavity.

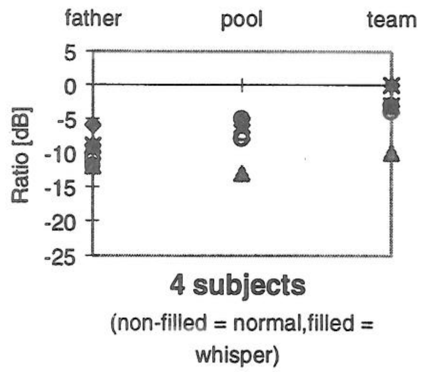


Figure 10.7 Loudness difference of the subjects' own voice caused by elimination of air borne sound. Stimuli: /aaa/ as in father, /uuu/ as in pool and /eee/ as in team. Non-filled signs are normal speech, filled signs are whispering. 4 subjects. Data from: [Békésy, 1949].

10.5.3.2 Effect of phonation level

Another agreement is that p_c/p_a is independent on speech level. Békésy compared normal speech with whispering and for only one subject whispering caused a clearly lower ratio. Excluding that particular subject, the ratio for normal speech and whispering lies within 3 dB for each subject. The data in the pilot experiment in **Figure 10.6** shows the same tendency. Except from the /eee/ at the left ear, the ratio is the same within 3 dB for loud, normal and soft speech up to the 2'nd harmonic. The 3'rd harmonics agree well for normal and loud phonation but it is more difficult for soft phonation. Soft phonation deviates up to 10 dB at the 3'rd harmonics.

10.6 DISCUSSION

Several factors affect the accuracy of the speech spectra being evaluated. The reference signal was measured with an estimated uncertainty of 2 dB. Considering all sources of uncertainty, the measured pressure ratio p_c/p_a is believed to be accurate to within 5 dB.

The measurements indicate that a significant fraction of the sound pressure in the ear canal is due to body conduction. To show that the sound pressures due to body conduction are physically reasonable, the required motion of the canal walls can be estimated using the model of the open ear canal. The body source is treated as a piston in the ear canal wall. The volume velocity entering the canal is $q = p_C / Z_C$, where p_c is the pressure generated in the canal and Z_C is the acoustic impedance presented by the canal. The corresponding piston displacement for angular frequency ω has an amplitude:

$$x = q/(A\omega) \tag{10.8}$$

where the effective piston area is A. The result of this calculation is shown in **Figure 10.8.**

Imagine that a point on the wall of the ear canal moves outwards and the contra-lateral point also moves outwards. The two points lie on a diameter and the growth in diameter corresponding to twice the point displacement and it can be compared with Oliveira's [1995] data. The displacement across a diameter is then $1\text{-}40\cdot10^{-6}$ mm depending on frequency. This is much less than the 20% change in diameter of the ear canal that was observed by Oliveira during opening of the mouth. It must therefore be concluded that the measured sound pressures correspond to cartilage displacements that are physically reasonable.

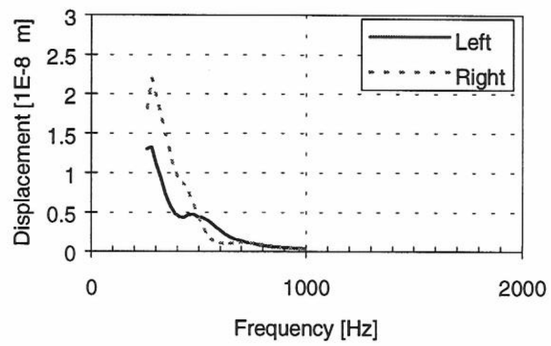


Figure 10.8 Displacement of the ear canal wall for body conducted sound due to one's own voice. Note, the linear frequency axis.

The decrease of the displacement with frequency is consistent with the known glottal source spectrum and equation (10.8). It is also known that the impedance of soft tissue increases with frequency, [Gierke and Brammer, 1995] up to about 15 kHz.

There are several body conduction routes that need to be addressed in developing a model. For example, consider **Figure 10.9** in which the open ear spectrum for a vowel sound is compared to the spectrum obtained with the ear closed with the attenuation box.

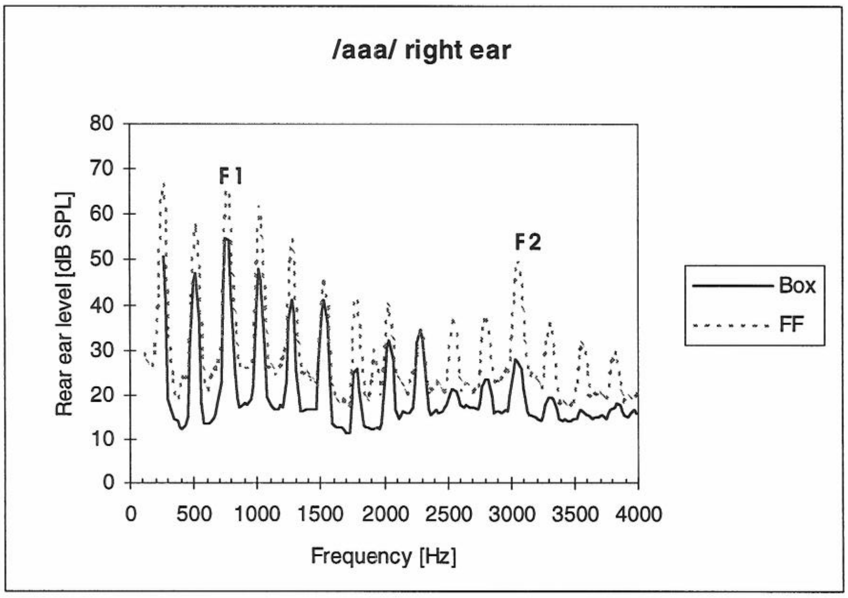


Figure 10.9 Spectrum of /a/ measured first in the ear canal on the free ear and then in the boxed ear. Note, the linear frequency axis.

The boxed ear spectrum shows the same resonant structure as the free ear spectrum and the formant (F1 and F2) are vaguely represented in the boxed ear signal as well. The formants are an effect of the acoustics of the vocal tract length and the shape of the oral cavities. This comparison indicates that the body conducted sound is produced, not only by vibration of the vocal cords and attached structures, but through pressure changes in the vocal tract and oral cavity. This is an important observation for the purpose of determining the BRTF but very little is known and there is still a long way to go.

10.7 PC/PA MEASURED ON 10 SUBJECTS

The results from the experiment are important in order to be able to model the sound transmission in the ear canal and the occlusion effect. But only one subject was used and the experiment must be considered as a pilot study. Since, the results agree with the literature and theoretical predictions, it was decided to do the experiment on a larger scale in order to get some statistical data. In a masters degree project by Bremmelgaard [1997] the ratio p_c/p_a was measured in nearly the same way as in the pilot experiment but using 10 subjects (21 to 33 years). A new, smaller but better attenuating box was made. The box had the inner dimensions 29.9 cm x 23.4 cm x 18.6 cm and was covered both on the inside and the outside with absorbing material. The box attenuated 40 dB in the frequency range from 100-2000 Hz. For this project 2 probe microphones were available which gave the opportunity to measure the right and left ear simultaneously. A set up with 2 probe microphones has two major advantages, namely that the subject does not have to repeat the same sound and that it is possible to determine phase information from the ratio p_c/p_a.

A statistical t-test confirmed that the sound pressure levels in right and left ear generated by the subjects own voice were significantly the same. This means that it is reliable to measure the sound pressure level in the right and left ear simultaneously.

10.7.1 The magnitude of PC/PA

Bremmelgaard estimated p_c/p_a as the simultaneous complex difference between the real ear sound pressures:

$$|H| = 20\log_{10}\frac{|p_{boxed}|}{|p_{open}|} \cong 20\log_{10}\frac{|p_c|}{|p_a|} \quad , \quad \angle H = \varphi_{boxed} - \varphi_{open} \cong \varphi_C - \varphi_A \qquad (10.9)$$

where;
H = complex ratio of body conducted and air borne sound
φ = phase

The subject who participated in the pilot study were also measured by Bremmelgaard. **Figure 10.10** plots both data sets for the right and left ear.

The right ear data agree well within ±5 dB, except at 600 Hz, where the pilot study shows a sudden dip. The left ear data does not agree so well, especially not between 250-400 Hz. Bremmelgaard's data are 7-10 dB below the pilot study data.

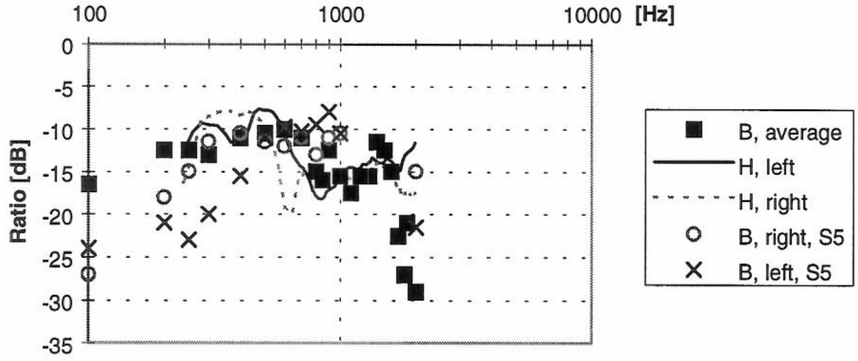


Figure 10.10 Ratio of body conducted and air borne sound. Continuous speech. H = subject in the pilot study. B, S5 = same subject in the master thesis project. B, average of 10 subjects.

Data points for the average ratio p_c/p_a for the 10 subjects are plotted as solid squares in **Figure 10.10**. The statistical deviation is mostly 5 dB in the frequency range from 100-2000 Hz. This (rather large) deviation is due to great individually differences of body and air conducted sound.

Some of the differences between the pilot study data and Bremmelgaard's data can be found in the way the ratios were computed. The method according to equation (10.9) has some advantages as already mentioned, but unfortunately it also introduces an error because p_c is not so much smaller than p_a that the body conducted sound can be neglected in the free field case. In a worst case the ratio plotted in **Figure 10.10** is numerically 3 dB too small[2], especially between 200-800 Hz, e.g. p_c/p_a should be -7 dB instead of -10 dB. The pilot study data (marked 'H') are computed with equation (10.5) which does account for the mistakes that lies in assuming that p_c is much smaller than p_c. The disadvantage is that the phase information is not included. Still the data corresponds well and agrees with literature data. The real ratio p_c/p_a lies within ± 5 dB of these measured data.

10.7.2 Phase difference of PC/PA compared to the literature

An estimate of the phase difference are derived from literature data on phase differences of signals traveling from one ear to the other. Sound velocities in the skull are also reported.

10.7.2.1 A literature review

Zwislocki [1953] did an experiment in order to estimate the acoustic attenuation between the ears. A subject received a tone in one ear through a receiver mounted in a doughnut cushion, which should create a minimal occlusion effect. The subject should then adjust the amplitude and phase of a tone in the other ear and compensate for the other tone. The phase shift was approximately a straight line with slope 0.00734 deg/Hz. This phase shift indicates a sound velocity between the ears of 260 m/s. A dip at 400 Hz and one at 2,000 Hz and a peak at 800 Hz are, probably caused by resonance.

[2] If p_a is 60 dB SPL and (p_{box}/p_{open})= 20Log$[p_c/(p_C+p_a)]$ =-10dB, then p_c=53 dB and 20Log$[p_c/p_a]$=-7dB

In 1981 Tonndorf and Jahn repeated Zwislocki's experiment. They also found dips in the phase shift curve around 400 and 2,000 Hz and a peak at 800 Hz. In both cases the sound velocity was frequency independent above 2,000 Hz. They calculated the sound velocity to 330 m/s. **Table 10.1** lists several sound velocities in the skull and it seems that the experimental set up and the age of the test person are decisive for the result.

Reference	Sound velocity	Comments
Békésy (1948)	540 m/s at 1,800 Hz 570 m/s at 1,000 Hz	Vibration pick ups on skull
Zwislocki (1953)	260 m/s at 100-2,000 Hz	Subjective compensating method with headphones. Subject a young man
Franke et al. (1952)	80 m/s at low freq. 150 m/s at 500 Hz, rapid increase 500-1,000 Hz to 300 m/s	Vibration pick ups on skull forehead to vertex
Wigand (1964)	2600 m/s phase velocity	Dry bones
Tonndorf (1981)	330 m/s at 100-2,000 Hz	Subjective compensating method with headphones. A 66 year old person.

Table 10.1 Measured sound velocities in the skull.

10.7.2.2 Measured data

The measurements by Zwislocki and Tonndorf use subjective perception of sound. The velocity must then include the sound transmission through bone and softer tissue and therefore their measurements will be used here. The relative phase between p_c and p_a depends on the travel length and the sound velocity in air and human tissue and the relative phase can be expressed as:

$$\left(\frac{d\varphi}{df}\right)_c - \left(\frac{d\varphi}{df}\right)_a = \frac{2\pi l_c}{c_c} - \frac{2\pi l_a}{c_a} \quad ;[rad / Hz] \qquad (10.10)$$

where;

l_a = distance in air from the inner of the oral cavity, around the head into the ear canal, assumed to be 30 cm

l_c = distance from the inner of the oral cavity through the body to the ear canal, assumed to be 15 cm

c_a = sound velocity difference in air = 343 m/s

c_c = sound velocity in human tissue

$d\varphi$ = change in phase

df = change in frequency

If c_c is set to 330 m/s then equation (10.10) gives the slope -0.150 deg/Hz when transforming from *rad* to *deg*. If c_c is set to 260 m/s then the slope is -0.107 deg/Hz. Thus, it is expected that the measured relative phase has a slope within this range. The measured phase difference and linear regression line is shown in **Figure 10.11**. The slope is -0.108 deg/Hz which corresponds to the predicted slope using c = 260 m/s. The measured data agree best with c = 260 m/s as measured by Zwislocki probably because the subject in the present experiment was young and Zwislocki used a young subject whereas Tonndorf used a 66 year old subject.

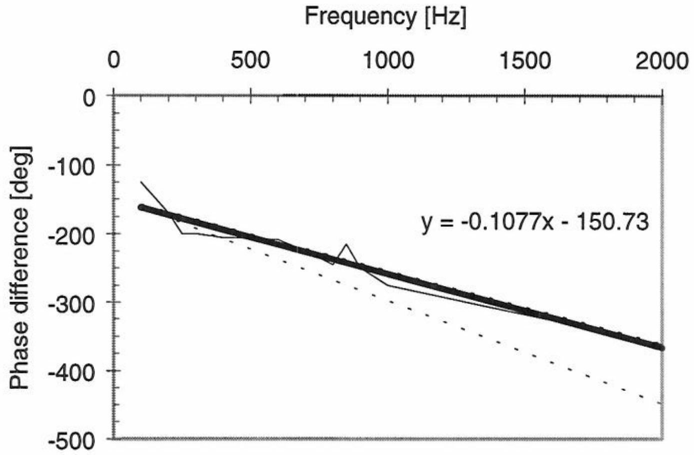


Figure 10.11 Phase difference between body conducted and air borne sound from one's own voice (thin line). Linear regression line shown on the plot (thick line). The dotted lines are the slopes estimated from the literature. Average of 10 subjects. Data from [Bremmelgaard, 1997]. Note, linear frequency axis.

10.8 A MODEL OF PC/PA

One purpose of the experiments was to determine the ratio p_c/p_a to be used as input in the mathematical model of the occlusion effect. Therefore, a simple expression for p_c/p_a will be determined.

10.8.1.1 Phase model

The phase decreases linearly with frequency, and an expression for the phase difference is already found (as shown in **Figure 10.11**) to be:

$$\varphi_c - \varphi_a = -0.108f - 151, \quad [\text{deg}] \tag{10.11}$$

where;
φ_c = phase of the body conducted sound
φ_a = phase of the air transmitted sound

10.8.1.2 Magnitude model

The individual ratios can vary a lot from the average shape and the ratio p_c/p_a depends most likely on the age of the subject because the human tissue changes with age. Bremmelgaard's data was measured on subjects in their twenties and thirties. Bremmelgaard measured up to 2 kHz, but the data from 1-2kHz has very large standard deviations. The most simple model could be as illustrated in **Figure 10.12** - 13 dB over the whole frequency range.

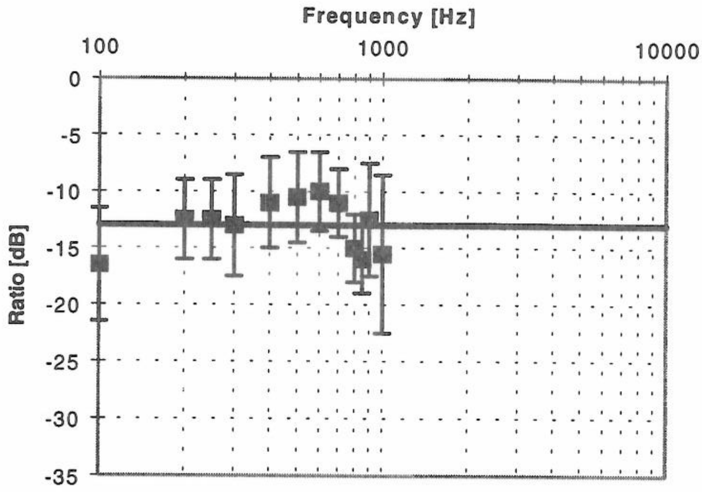


Figure 10.12 Ratio of sound pressure in the boxed ear and the free ear. Filled squares = data points, average of 10 subjects. Vertical lines = standard deviation. Solid line = simple model. Data from [Bremmelgaard, 1997].

The occlusion effect can now be estimated with the same model as used to derive the predictions in **Figure 10.5** but where $p_c/p_a = -13$ dB and the relative phase in **Figure 10.11** is used. The volume velocity of the body source is derived from a model of the open ear. When the ear is occluded the volume velocity depends on the insertion depth of the earmould with a gain factor = (1-insertion depth/ear canal length). The estimate is compared to occlusion effect measured on real ears in **Figure 10.13**. The slope of the occlusion effect is, as excepted from the simple calculations in chapter 1, approximately decreasing with 40 dB/decade.

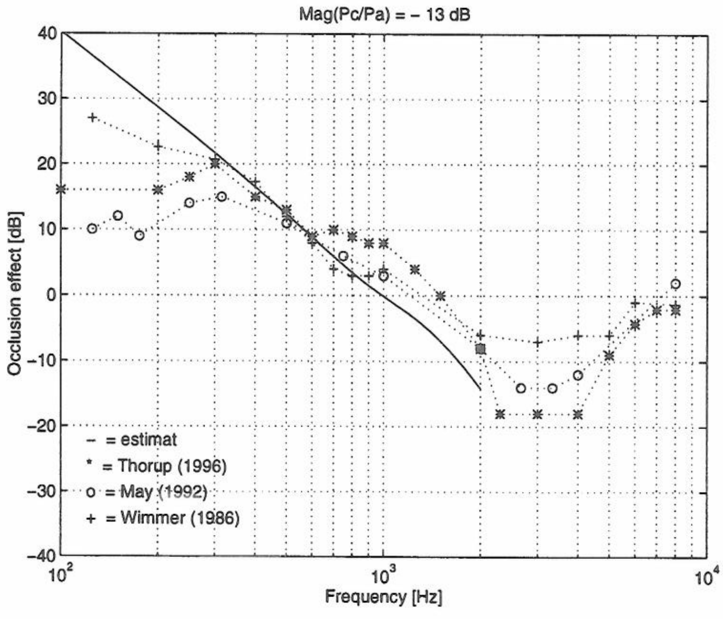


Figure 10.13 Estimated occlusion effect compared to measured data.

Above 2 kHz less is known about the ratio. The signal to noise ratio is a problem. As explained previously, the air conducted sound decreases with frequency above the ear canal resonance and the ratio is then dependent on which of the two signals p_c and p_a that decreases most rapidly. In fact, judging from the measured occlusion effect, it looks like the ratio increases above the ear canal resonance. But the increase in occlusion effect could also be caused by a low signal to noise ratio. Conclusively, it is unknown what the actual situation is above 2 kHz and more experiments are needed. This problem will not be addressed further in relation to hearing aids because most hearing aid users have a high frequency hearing loss. The natural sound of one's own voice at high frequencies is therefore not important.

Below the fundamental frequency (about 125 Hz for men and 200 Hz for women) the ratio is difficult to determine because the speech signal decreases rapidly towards lower frequencies. From **Figure 10.12** it looks like the ratio is lower at 100 Hz than at 200 Hz and also the estimations based on measured occlusion effects in **Figure 10.5** indicates that the ratio decreases at 100 Hz. A better indication than looking at the average ratio is to look only at the male spectra because the signal to noise ratio here is better below 200 Hz. A look at the individual spectra from Bremmelgaard strongly suggest that the ratio rolls off from the fundamental frequency to lower frequencies. Judging from **Figure 10.13** the occlusion effect rolls off at low frequencies which is mainly due to leakage between the earmould and the ear canal wall. In addition, it could be that the ratio rolls off too, but it does not make a difference in the estimated occlusion effect, so it is just as good to keep the simple frequency independent model.

10.8.2 How to estimate the volume velocity source

The body source is given as a complex pressure ratio of the air conducted sound. By means of the model, the corresponding volume velocity of the vibrations of the ear canal walls is determined. The body source is, of course, distributed over the whole ear canal surface but the distribution pattern is unknown. Some parts of the ear canal wall might move more than other parts. For the sake of simplicity and practical incorporation, it is presumed that the body source can be gathered at one single point at some cross-section in the ear canal. This point will hereby be defined as the point where the probe tube is located, i.e. 3 mm behind the insertion of an earmould. This point is called 'P' and this point is used because the ratio $\hat{P}_C \,/\, \hat{P}_A$ is estimated from measurement of the pressure in this location, see model in chapter 1.

The body source is computed with use of the model for the open ear. The reference point called 'E' in the middle of the concha is given a constant frequency independent pressure, for example 60 dB SPL. The pressure in point P is then found by transforming the reference point pressure into point P:

$$p_a^P = p_{ref}^E H^{EP} \qquad (10.12)$$

where;
p_a^P = air borne pressure in point P inside the ear canal
H^{EP} = transformation from point E to point P, i.e. concha to point P in the ear canal

The pressure in point P due to body conducted sound, p_c, is then found by $p_c = p_a(\hat{P}_C \,/\, \hat{P}_A)$. The volume velocity source in point P is:

$$q_c = \frac{p_c}{Z_{ear}^P} \tag{10.13}$$

where;
Z_{ear}^P = the impedance of the open ear seen from point P

10.9 CONCLUDING REMARKS

The body conducted sound is 13 dB lower than the air borne sound from one's own voice, but as a preliminary estimate the ratio 10 dB was used. These results agree with the data from Békésy [1949], who writes that these results demonstrate that hearing one's own voice by bone conduction is of the same order as hearing by air conduction. However, Békésy took the difference in loudness between the free ear and the tube covered ear. In the free ear both air conduction and bone conduction is present, so for example for the word 'team' the bone conduction component might theoretically be even larger than the air conduction component.

The problem is that very little is known about body conduction of one's own voice. In contrary, *perception* of speech from other people has had a lot of attention. The perception of one's own voice under different listening conditions (such as an occluded ear) has been overlooked and there are enough aspects for years of studies in this field.

Knowledge about body transmitted sound created from one's own voice is of interest for the basic research of the human body but should also be of interest for applied research. It has been shown here that the body conducted sound to the ear canal does matters how one's own voice is perceived, hence a significant change in the outer ear condition could cause a change of perception of one's own voice. It has obvious interest in audiology, for the fitting and new developments of hearing aids. Persons with pathological abnormal conditions of the outer ear, might perceive their own voice in an 'unnatural' way, for example when the ear canal is swollen or in the more extreme cases such as congenital abnormalities.

11. MODEL OF THE EAR

A chart of the total model is presented here to give a better overview of the model. The occluded ear dimensions are analyzed. Until this point in the report, the open ear has been modeled as a uniform tube. The open ear will now include concha, no pinna and the radiation from a point in the head. The reader who is interested in modeling the ear should also read appendix H.

11.1 OVERVIEW OF THE MODEL

The model is based on the two-port theory, which is explained in appendix A, where the associated Matlab program code is also presented. A 2-port network of the open ear and occluded ear is illustrated in **Figure 11.1**. The ear canal and concha are modeled with acoustic transmission lines and the middle ear with a lumped element model. The inner ear is not included in the model, because the effect of the inner ear component on the occlusion effect is negligible [Schroeter and Poesselt, 1986]. The model presumes only that plane waves exist and one-dimension is therefore adequate. When only plane waves exist, then the sound pressure will be the same in all points in one cross-section and the model can be divided into a number of reference planes.

Default settings:

Ear canal: diameter = 7.1 mm, length (C to D) = 23.5 mm
Concha: diameter = 22 mm, depth (R to C) = 6 mm
Reference point in plane R
Z_{rad} = piston in a sphere, diameter = concha diameter = 22 mm
Acoustic length of ear canal (R to D) = 23.5+ 0.85*(diameter/2) mm
Cartilage of ear canal = ½ of ear canal length = 11.8 mm
Earmould : insertion depth (C to B) = 7.8 mm
Probe microphone additional depth = 3 mm (B to P)
Earmould length: BTE = 23.3 mm, ITE = 21.5 mm, ITEC = 20.2 mm
Radiation of leakage and vent included
The body source, q_c, is calculated from the sound pressure in the open ear canal generated by air borne sound, p_a, and body conducted sound, p_c:
$|pc/pa| = -13\text{ dB}; \ \varphi_C - \varphi_A = -0.108f - 151, \ [\text{deg}]$
Z_{md} = modified Zwislocki network, eardrum and middle ear

The ear canal is a chain of cylindrical rigid tubes represented by 2-ports between planes C-B, B-P and P-D. The length C to B is the insertion depth of the earmould. The concha is also implemented as a rigid cylindrical tube from R to C. Only the inner 6 mm of concha is considered because the reference point is defined to be positioned 6 mm from the ear canal entrance (see appendix H session. H3.1). The reference point is chosen to be located in the middle of the concha because it is approximately the location for the microphone on an ITE hearing aid[1].

[1] The placement of the microphone can vary with hearing aid design and the form of the moulded shell.

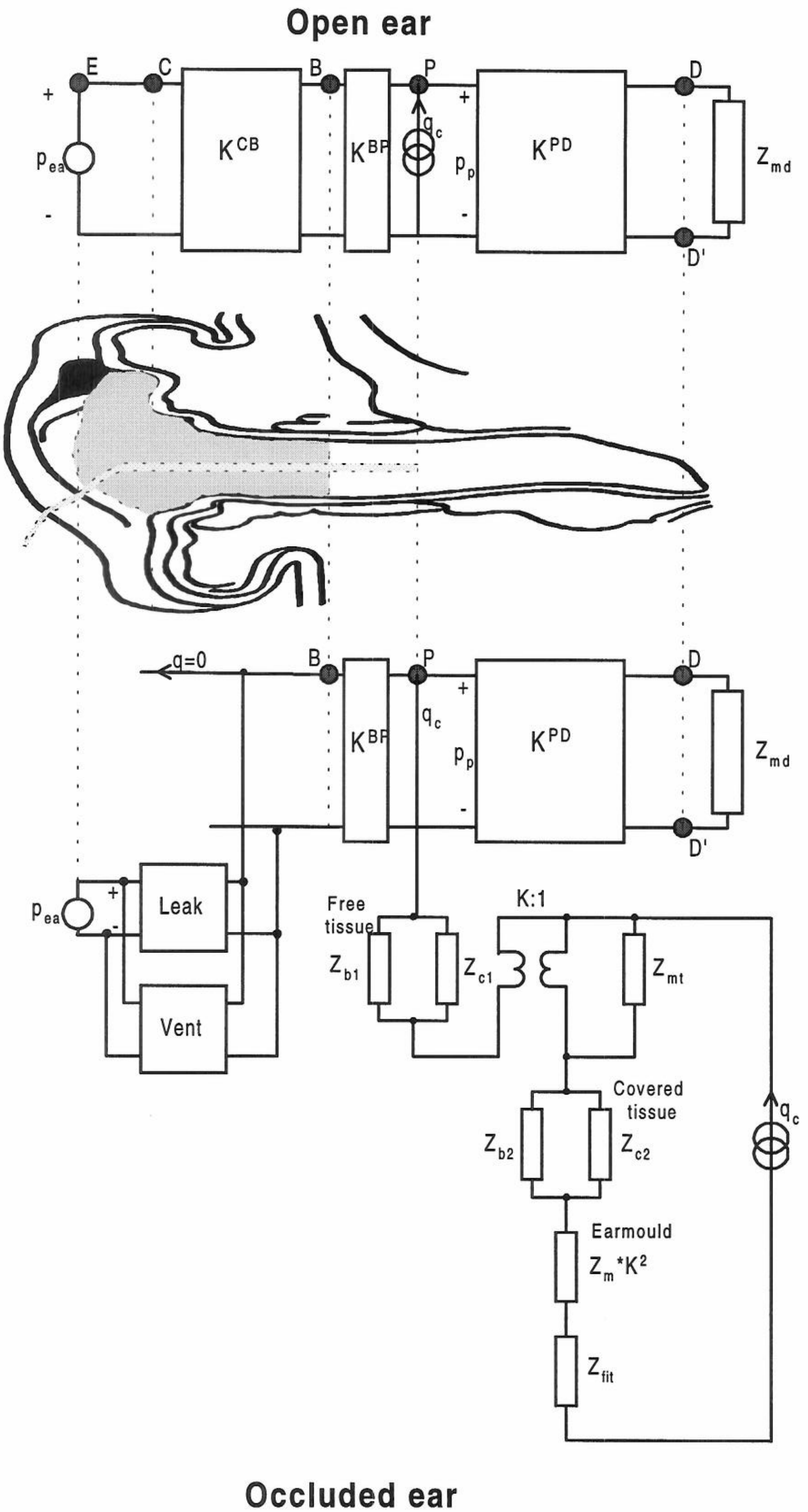


Figure 11.1 Acoustic model of the occlusion effect from one's own voice. Prefix K = transmission line, prefix Z = lumped element representation. Note that the outer ear is not drawn with the correct proportions. The model of the occluded ear is described in more details in chapter 8. P_a = air borne incoming sound, q_c = volume velocity of body transmitted vibrations.

The radiation from the ear canal can be simplified as an open tube in an infinite baffle (see appendix H section. H3.2). The radiation is compensated by increasing the length of the ear canal in order to get the correct ear canal resonance frequency at 2.6 kHz. The pressure is measured in plane P with a probe microphone. For the sake of ease the body source is also located in plane P. The residual ear canal behind the earmould goes from plane B to D. The eardrum and middle ear, Z_{md}, is modeled with a modified Zwislocki network, (see appendix H section. H.1). The modified Zwislocki model presumes that the eardrum works like two pistons instead of one piston as in the original Zwislocki network.

11.2 MODEL OF THE OPEN EAR

11.2.1 The outer ear

The outer ear consist of the pinna and ear canal terminated with the eardrum. The anatomy is described in chapter 5.

The ear canal is not a simple geometry. It has 2 bends and varies in diameter along the length and it is closed in one end with the eardrum that is angled to the ear canal walls. The ear canal shape and dimensions even differ significantly from person to person. Despite these complexities some simple acoustical models can be used.

Even when the radiation impedance for the concha part is included in the model, the effective acoustic length of the ear canal is not correct. If the model only includes the physical ear canal length, the ear canal resonance will be estimated incorrectly to 3.1 kHz instead of the real ear resonance at 2.6 kHz. It is therefore still necessary to correct the ear canal length. Shaw´s [1974] physical model of the ear canal with concha and pinna also showed this phenomena. He measured the resonance of the first mode in the system (the ear canal resonance) to correspond to an effective length of the ear canal of 28 mm instead of 25 mm which was the physical length. Hence, the acoustic effective length of the ear canal cannot be neglected. If the ear canal was infinite flanged, the effective acoustic length would be approximately 0.85·radius of the ear canal. The primary reason for the disagreement between the theoretical and measured ear canal resonance is that the concha, the tragus and the crus helias interacts with the ear canal. In children the resonance frequency is higher because the ear canal is shorter. The adult resonance frequency around 2.6 kHz appears around the age of 20 years [Ballachanda, 1995].

A tube is an acoustic transmission line. It means that the sound pressure varies along the length of the tube. In a tube closed at one end, the incoming sound wave will be reflected and creates a standing wave inside the tube. At the points where the incoming sound wave and the reflected sound wave interferes, a minimum occurs where the sound waves are 180° out of phase. A maximum occurs where the waves are in phase. Consequently, the sound pressure varies along the length axis of the tube.

Another important point to make is that a one-dimensional model assumes that the sound pressure is the same all over a cross-section. This is true in a tube as long as higher modes do not exist. The first higher mode occurs as a transverse resonance at the frequency [Rasmussen, 1991]:

$$f = \frac{1.84c}{2\pi a}; \quad [Hz] \tag{11.1}$$

where;
a = radius
c = 343 m/s at 20°C

Below this frequency only plane waves exist. In a uniform rigid tube of 7.1 mm in diameter, there will only be plane waves for frequencies under 28 kHz. The cut off frequency is lower in the real ear canal because the ear canal is not rigid.

Rabbitt and Holmes [1988] analyzed theoretically higher modes propagation in the ear canal in a cat and applied the results to the human ear. They found an infinite number of higher modes in the ear canal, but at low frequencies (f < 1 kHz) to moderate frequencies only the plane wave propagates. At higher frequencies only the lowest modes propagates. At 1 kHz the lowest modes are trapped at the eardrum. At 6 kHz the first higher mode becomes zero about 7 mm from the bottom of the eardrum in an ear canal with length 16 mm. The magnitude of the higher modes are 70% smaller than the magnitude of the plane wave. The occlusion effect is only effective up to 2 kHz, so a one dimensional model is adequate as long as the higher modes that exists at the eardrum are neglected.

The final model is drawn in **Figure 11.1**. The model shall be used to calculate the sound pressure in some point inside the ear canal relative to the reference point 6 mm out from the ear canal entrance.

Real ear data for the sound pressure difference between the reference point and the eardrum is obtained from the numerical data material provided by Bentler and Pavlovic [1989]. They took data from several studies to compose transfer functions from the free field (azimuth 0°) to various points in the ear for the purpose of hearing aid design. A transfer function from the center of the concha in an In-The-Ear hearing aid microphone location to the eardrum is derived from these data by subtracting the transfer function 'free field to eardrum' from the transfer function 'free field to In-The-Ear microphone location'. The data are taken from the table for third octave band frequencies. 'Free field to eardrum' is the same data as given in [Shaw and Vaillancourt, 1985].

The calculated transfer function from the reference point to the eardrum is in **Figure 11.2** compared to data from real ear measurements. The calculated curve follows the real ear data curve up to at least 8 kHz. The resonance around 2.6 kHz is the so called ear canal resonance and the resonance (seen as a bump on the curve) around 6 kHz is due to the depth resonance of the concha.

The greatest deviation from the measured data is -2 dB at 1-2 kHz and the reason is that the pinna, head and torso are not included in the model.

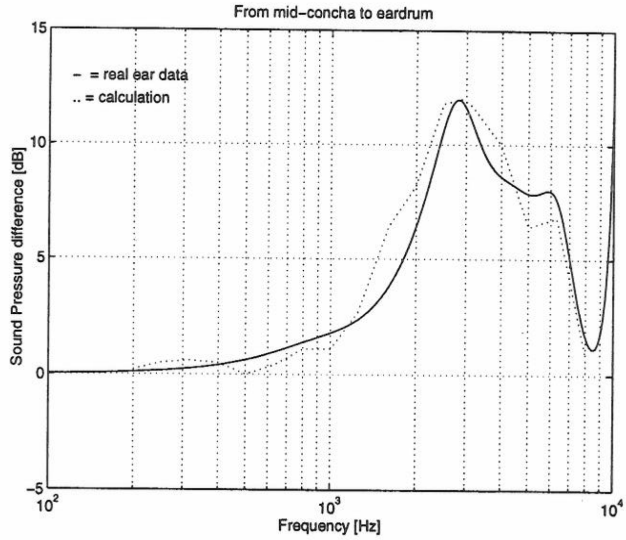


Figure 11.2 Transfer function from mid-concha to eardrum. Solid line: calculated with the model in **Figure 11.1.** Dotted line: Real ear data. From: [Bentler and Pavlovic, 1989].

11.3 MODEL OF THE OCCLUDED EAR

11.3.1 Effect of a non-uniform ear canal

The ear closed with a rigid termination is easier to model than the open ear because it is not necessary to account for the radiation effect. In the simple case with a rigid mould and no leakage, the only sound source is therefore the body conducted sound as illustrated in **Figure 11.3**. For illustration purposes, the internal sound source is placed at the position of the earmould. Note, that the piston does *not* symbolize that the earmould oscillates.

The produced sound pressure depends on the volume of the enclosed ear canal. In practice it is illustrated by varying the insertion depth of the earmould in the cartilaginous part of the ear canal. The characteristic of a closed tube has already been shown in the introduction to this report, see **Figure 1.2,** and the sound pressure at the eardrum becomes:

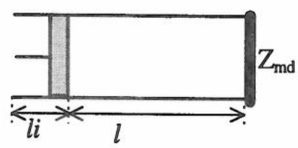


Figure 11.3 Model of the tightly closed ear canal. li = insertion depth.

$$p_{eardrum} = \frac{Z_{eardrum}}{j(S/\rho c)Z_{eardrum}\sin kl + \cos kl}q \tag{11.2}$$

where;
l = length of the occluded ear canal
q = volume velocity source
$p_{eardrum}$ = sound pressure at the eardrum
S = cross-section area

From (11.2) it is derived that both the occluded length and the cross-section affects the sound pressure in the ear canal. Instead of looking at the occluded length, the insertion depth, *id*, can be used.

The open ear could reasonably be well modeled as a uniform tube, but in the occluded ear it is more correct to change the diameter of the tube to account for the non-uniformity of the real ear canal whose cross-section varies along the canal length. In chapter 5 the anatomy of the ear canal was described and **Figure 5.4** shows that the diameter of the ear canal gets larger near the opening and very small towards the eardrum.

According to the data presented in chapter 5, different studies have resulted in slightly different data on the ear canal volume and length for an average adult person. Salvenilli et al. [1991] have made the largest study counting 280 cadaver ears, and their data have therefore been used in this report and the occlusion effect model. Unfortunately, their data study does not give data on the average shape of the ear canal. Therefore the data from Johansen [1975] is used to find the relation between the insertion depth of the earmould and the value of the enclosed volume.

Johansen's data for the ear canal volume as a function of the position from the eardrum as printed in **Figure 11.4**. Johansen's data are plotted with large dots and by the use of the program 'EXCEL', a trend line has been fitted to Johansen's data.

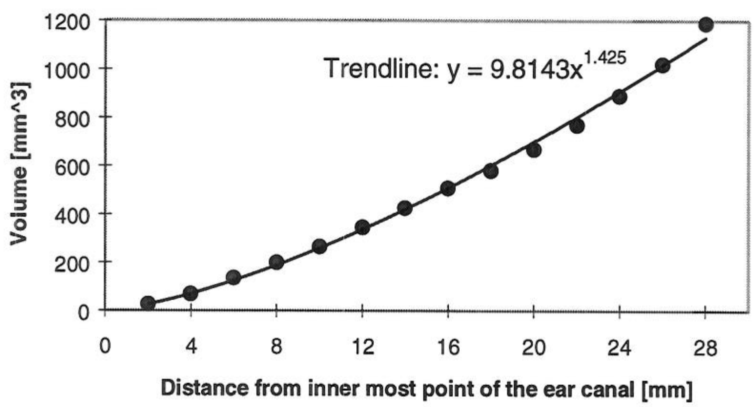


Figure 11.4 Volume as a function of the position in the ear canal seen from the point of junction between the eardrum and the inferior wall. Data points from: [Johansen, 1975, figure 3]. The trend line is derived by the author of the present report.

The position in the ear canal can be expressed by the length of the ear canal and the insertion depth which is a parameter in the occlusion effect model. The enclosed volume can be found by the trend line estimate:

$$V_{occluded\ ear} = 9.81(l_{ear\ canal} - l_{insertion})^{1.43} \qquad (11.3)$$

where;
V = volume in mm^3
l = length in mm

If the enclosed volume is assumed to have the shape of a cylindrical tube, then the diameter, $d_{occluded}$ for the specific insertion depth of the occluded ear canal is found by:

$$d_{occluded} = 2\sqrt{\frac{\tilde{V}_{occluded\ ear}}{\pi(l_{ear\ canal} - l_{insertion})}} \qquad (11.3\ )$$

where;
$\tilde{V}$ = estimated volume

The diameter based on Johansen's data uses the total physical volume of the ear canal. In the acoustic sense, the volume under the wedge shaped eardrum is not relevant. This volume can be neglected by calculating the volume from equation (11.2) with $l_{ear\ canal}$ = 25.5 mm and $l_{ear\ canal}$ = 2 mm, subtracting the two volumes and using the resulting volume (980.5 mm^3) in equation (11.3). This method gives a length of 23.5 mm, corresponding to the average length found by Salvenilli et al. [1991], but the diameter is a bit to large, namely 7.3 mm instead of Salvenilli et al's 7.1 mm. Therefore, Johansen's data will not be used, but equation (11.2) and (11.3) will be used and corrected. If the volume in equation (11.3) is set to be the volume found by equation (11.2) subtracted with the correction (50 mm^3) then equation (11.3) corresponds to Salvenilli et al's data.

The diameter, $d_{occluded}$, is plotted against the insertion depth, $l_{insertion}$ for a 23.5 mm long ear canal, see **Figure 11.5**. For the sake of comparison, the diameter of the uniform tube with no correction for decreasing volume is also plotted. At an insertion depth of 23.5 mm, $d_{occluded}$ must be 0 mm. For example, if the insertion depth is 10 mm, then the diameter for a uniform tube that corresponds to the correct volume is 6.5 mm, instead of 7.1 mm.

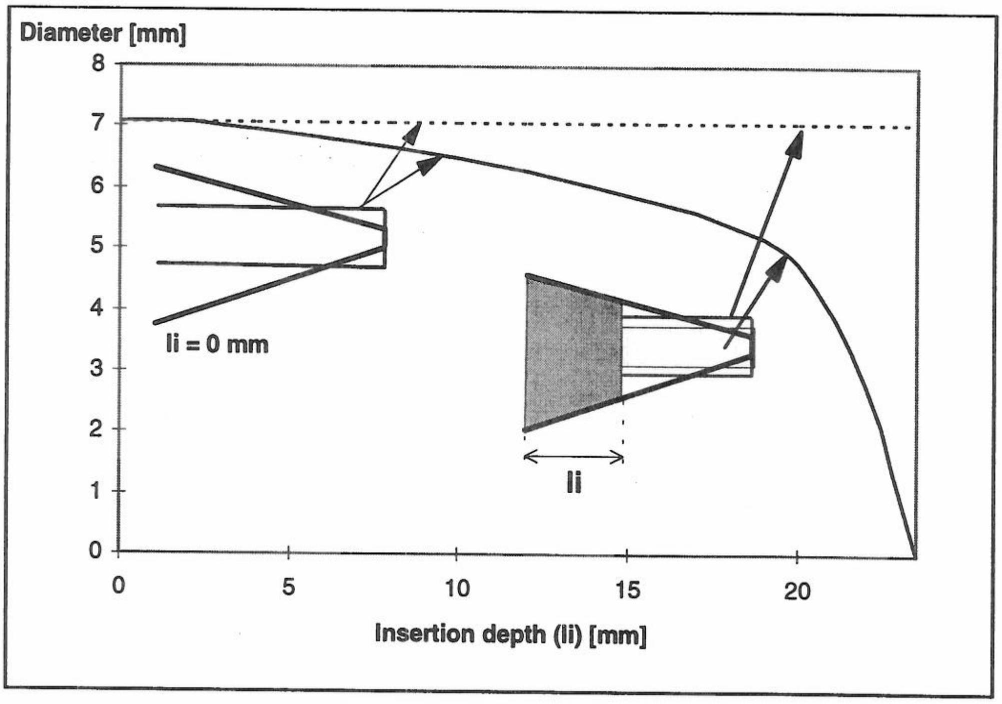


Figure 11.5 Diameter of a tube shaped ear canal as a function of insertion depth (li). The ear canal is 23.5 mm long. The diameter is calculated from equation (11.3).

Figure 11.5 shows that the diameter of a uniform tube model depends on the insertion depth. If a diameter of 7.1 mm is used for all insertion depths, then the occluded volume becomes to large compared to the occluded volume in a real ear. The acoustical consequence is illustrated in **Figure 11.6.**

The sound pressures in a tube with diameter 7.1 mm and in a tube with the corrected diameter is calculated in **Figure 11.6**, where the difference between the sound pressure in the occluded ear canal with diameter 7.1 mm and the corrected diameter are shown. Corrections to the occluded volume according to equation (11.2) are especially important above 1 kHz and with very deep inserted earmoulds. A standard hearing aid mould is inserted about 8 mm from the entrance and the difference is less than 1.5 dB for all frequencies.

Thus, in this case it is not strictly necessary to make corrections for decreasing the diameter of the tube model. The same conclusion is reached if only the response below 1 kHz is of interest.

The bony ear canal begins about 11 mm from the entrance. Deep inserted earmoulds such as Complete-In-the-ear-Canal hearing aids, extend into the bony ear canal and it will be necessary to account for the decreasing diameter and volume.

The present model of the ear canal serves the purpose of estimating the occlusion effect with a mould in the cartilaginous part of the ear canal only. The constant tube model is therefore adequate when an error of 1 dB at low frequencies and a maximum about 2.5 dB below 8 kHz are allowed.

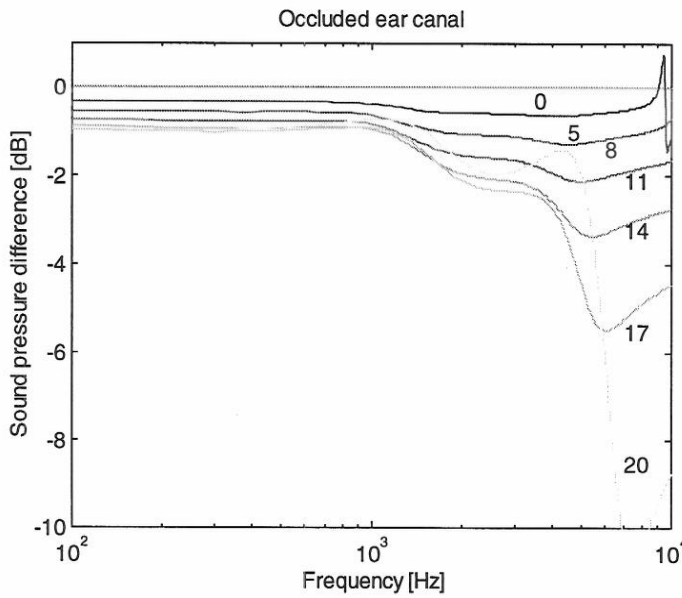


Figure 11.6 Difference between the sound pressure in a tube with diameter 7.1 mm and a tube with a corrected diameter according to the volume. The total ear canal was 26 mm long and terminated with the middle ear impedance. The numbers on the curves are the insertion depth in mm.

12. GENERAL DISCUSSION

The previous chapters discuss each of the main parameters for occlusion as a separate investigation. This chapter presents a discussion where the parameters are seen in relation to one another. The discussion is based on model estimations and tries to explain the specific limitations of the model and the problems connected to the prediction of the occlusion for the individual person.

12.1 ESTIMATIONS COMPARED TO AVERAGE DATA

The model of the open ear is well known (chapter 11). When the ear is occluded some new parameters become important. Some parameters can be determined precisely: probe location, earmould mass and earmould size. Other parameters are more difficult to measure, e.g. the exact size of the individual ear canal. The size of the occluded ear canal is defined to be 2/3 of the total ear canal, based on the assumption that a standard mould reaches 1/3 into the total ear canal length. The size of the leakage is determined upon fittings to the measured data. The tissue properties are only determined based on very few measurements, but cannot be estimated better with the available data.

The most uncertain parameter in the model (chapter 11) is the coupling between the tissue area covered by the earmould and the free tissue. Previously, this impedance was estimated based on one subject only and it is not necessarily the same as for the average person, keeping in mind that the occlusion effect varies up to 30 dB between subjects. This is illustrated in **Figure 12.1** where 16 subjects were fitted with acrylic earmoulds. The occlusion effect ranges from 0 to 35 dB at 250 Hz.

The estimated pressure in the occluded ear canal is also very sensitive to the value of the coupling between the free tissue and the tissue covered by the earmould in the ear canal, Z_{mt}, see chapter 8. If for example Z_{mt} is ¾ of the value used in **Figure 8.14,** then a problem with the insertion depth arises. An insertion depth of 11 mm causes 10 dB more occlusion effect than an insertion depth of 8 mm, and that is probably not what happens in a real ear. In a real ear the occlusion effect with a mould inserted 11 mm should theoretically be the same or less than the occlusion effect with 8 mm of insertion. The value of Z_{mt} used in **Figure 8.14** fulfilled this requirement. In **Figure 12.2** the occlusion effect for various values of Z_{mt} is shown.

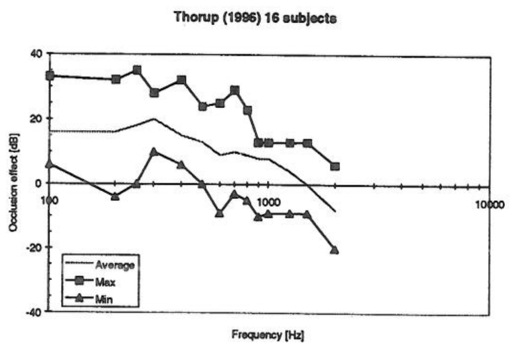


Figure 12.1 Occlusion effect from the subjects' own voice. 16 subjects. Tightly fitted mould. Data from [Thorup, 1996].

The estimate fits the data from Wimmer [1986] and May and Dillon [1992] between 300-2000 Hz best when the coupling impedance is $Z_{mt} = Z_c/4$. If this value is to be used the problem just described arises when the insertion depth is 11 mm.

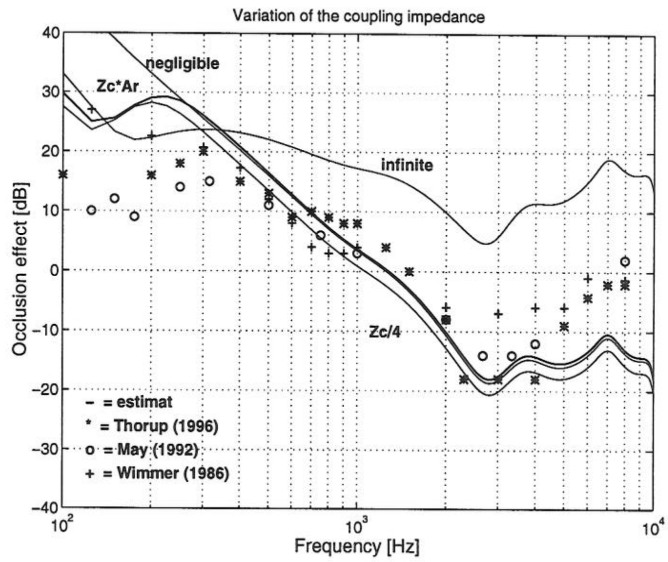


Figure 12.2 Occlusion effect from the subjects' own voice. Literature data (points) and estimated data (solid lines) with 4 different values of the coupling impedance. Z_c is the impedance of the cartilage. Ar = $(A_{covered}/A_{ear\ canal\ surface})$. The coupling impedance values are: negligible, Zc/4, Zc·Ar and infinite. Model default values. Earmould mass = 4 g. No leakage. p_c/p_a = -13 dB (see, **figure 12.3**).

Besides this problem the model can be used to estimate the measured data quite well, when the correct mechanical properties of the earmould and the leakage are included. Estimations and data from three studies are pictured in **Figure 12.3**. Below 1 kHz, the estimations agrees well with the literature data. The maximum deviation from the literature data is 6 dB but the most common deviation is not greater than ± 3 dB. This is fully acceptable because the uncertainty for a single measurement is about 5 dB and the standard deviation from the average data is about 6 dB for a foam plug [Bremmelgaard, 1997] and 10 dB for acrylic moulds [May and Dillon, 1992].

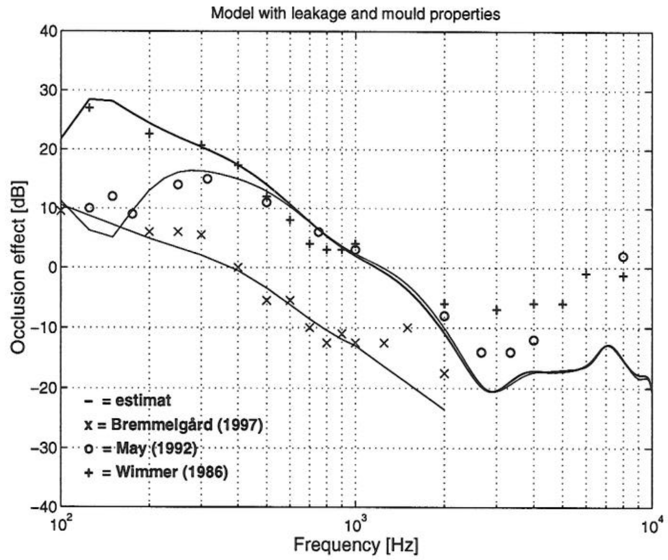


Figure 12.3 Occlusion effect from the subjects' own voice. Literature data (points) and estimated data (solid lines). Bremmelgaard [1997], 10 subjects: foam plug, inserted 12.7 mm, model slit: length = 14 mm, width = 4 mm, thickness = 0.25 mm. May and Dillon [1992], 10 subjects: acrylic plug (3 g), inserted 7.8 mm, model slit: length = 23.3 mm, width = 15 mm, thickness = 0.25 mm. Wimmer [1986], 4 subjects: impression material (10 g), inserted 7.8 mm, model slit: length = 23.3 mm, width 10 mm, thickness = 0.25 mm. Zmt = Zc/4. Otherwise default values.

12.2 PARAMETER VARIATIONS

12.2.1 Earmould fitting

A problem arises when including the earmould properties. The estimated occlusion effect gets too small. When there is no leakage, the sound pressure is 10 dB ($f < 200$ Hz) lower than if the mass of the mould was not included in the model. If the leakage is very small or very narrow, this is not so noticeable, but when the leakage becomes wider and more square like, the occlusion effect decreases very rapidly towards lower frequencies. And this is not seen in the measurements. As an example, look at subject 5 in **Figure 12.4**. In chapter 9, **Figure 9.14** the occlusion effect for subject 5 was estimated really well with a slit of width 2.5 mm and thickness 1 mm. However, if the same slit size is used in the new model including the earmould, the occlusion effect estimation is worse than the first estimate where the earmould was just a rigid wall, see **Figure 12.4**.

The occlusion effect cannot be estimated better by applying another leakage size and if the resonance peak should be matched, the occlusion effect will be too low below 200 Hz. Another possibility is to adjust some of the many other parameters in the model. It is evident that the fitting of the earmould must play a role. The fitting of the earmould is modeled with an impedance in series with the earmould, so if the earmould is very light but very tightly fitted, then the impedance will still be large. The fitting impedance is modeled as a pure compliance, $Z_{fit} = (K/j\omega)/A^2$ and a variation of K is illustrated in **Figure 12.4**. A value of $K = 6 \cdot 10^5$ N/m^3 provides the best fit to the measured data, but with this value of Z_{fit} the earmould mass becomes irrelevant.

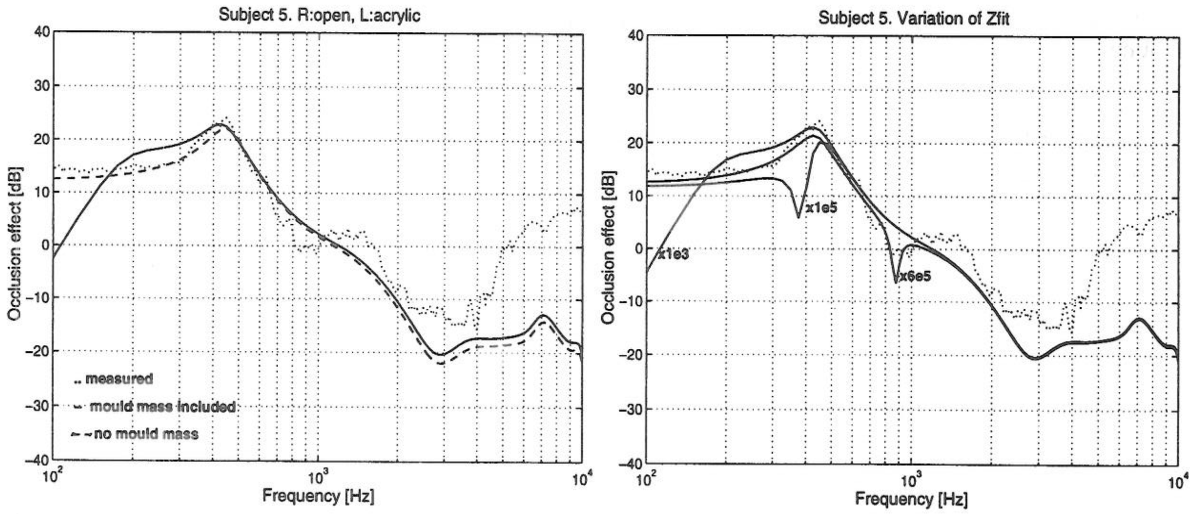


Figure 12.4 Measured and estimated occlusion effect. Slit leakage length = 21 mm, width = 2.5 mm, thickness = 1 mm. Left: Earmould mass 4.6 g included (solid line). No earmould, rigid wall (slashed line). Right: variation of the value K in the equation for $Z_{fit} = (K/j\omega)/A^2$.

12.2.2 Varying the length of the earmould

In chapter 8 and 11 the effect of varying the insertion depth was discussed theoretically. Based on the latter discussion there is no doubt that the individual fitting of the earmould is important for the occlusion effect. Therefore it cannot be expected that inserting an acrylic mould deeper into the ear canal will give a smaller occlusion effect.

An experiment was performed with the purpose of studying how well the theory holds. The same 4 subjects who participated in the acrylic-foam plug experiment had 3 acrylic moulds moulded for each ear. The three moulds were of different lengths. The smallest difference in length between two moulds for the same ear was 1 mm and the greatest difference was 4 mm. The average difference between the shortest and the longest mould for the same ear was 4 mm. The medium length mould was made as a standard mould, so in comparison with the model, the default insertion depth is 7.8 mm. The experiment and the results are given in appendix C.

Figure 12.5 shows results from the subject with the largest difference in occlusion effect and the subject with the smallest difference.

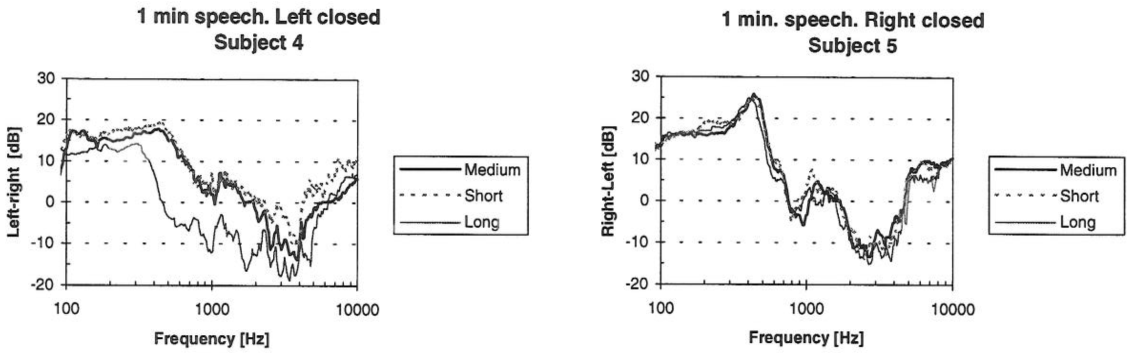


Figure 12.5 Occlusion effect with 3 earmoulds of varying length for the same ear. Subject 4: lengths: (between long and medium) = 1.5 mm, (between medium and short) = 2.5 mm. Subject 5: lengths: (between long and medium) = 2 mm, (between medium and short) = 3 mm.

Subject 5 has practically the same occlusion effect with a short, a medium and a long mould. In contrast, subject 4 has a smaller occlusion effect with a long earmould than a medium and a short, even though subject 4's long mould is only 1.5 mm longer than the medium mould whereas subject 5's long mould is 2 mm longer than the medium. The occlusion effect for subject 4 with the long mould is about the same as for the short and the medium mould from 100-300 Hz probably because of leakage. From 300-450 Hz the occlusion effect decreases rapidly. The same tendency is seen for the other subjects, if there is a difference. In general the difference is negligible between the short and the medium mould and largest between the long and the two other moulds.

In conclusion, the occlusion effect is not necessarily reduced because the mould is inserted deeper and if the occlusion effect is reduced it seems that the reduction is *not* effective at the lowest frequencies. This contradicts the general belief in the literature [e.g. Revit, 1992] but not in the clinical experience.

The reduction is individual and must mainly depend on the actual shape of the earmould and the exact shape of the ear canal. Therefore, it cannot be expected that the model can predict the effect of the different insertion depths very well for the individual subject, unless very accurate data for the individual subject is known. The theoretical estimated occlusion effects for insertion depths of a short, a medium and a long mould are pictured in **Figure 12.6**. The model with the earmould inertia and the $Z_{fit} = (6 \cdot 10^5\ N/m^3\ /j\omega)/A^2$ is used. Leakage is not included.

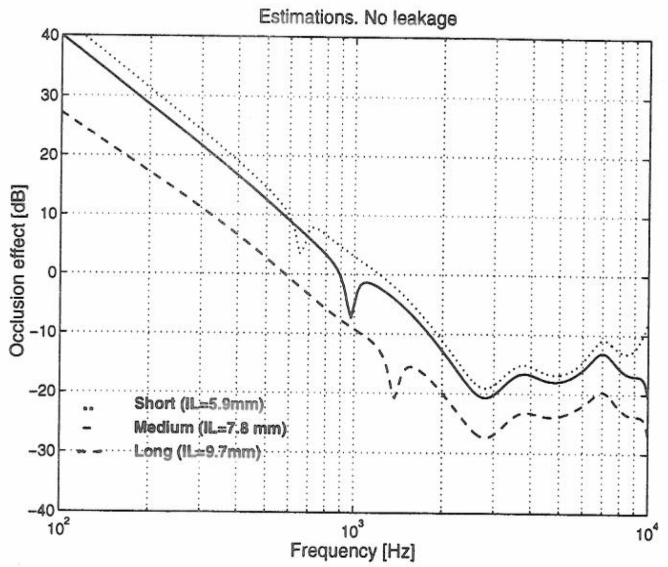


Figure 12.6 Estimated occlusion effect with variable insertion depths (IL). Earmould mass = 4g. Zfit included. $Z_{mt} = Z_c/4$.

At some points the predictions agree with the measured data. The difference between a short and a medium mould length is negligible and the occlusion effect with a long mould is smaller than the occlusion effect with the two other moulds. The difference is about 10 dB for f<1 kHz which corresponds with subjects 4's data. Below 300 Hz, the leakage is dominant over the effect of the insertion depth when the value of the tissue coupling is one quarter of the impedance of the cartilaginous tissue in the ear canal, $Z_{mt} = Z_c/4$. If Z_{mt} is 25% greater then the difference between all moulds are negligible.

12.3 MEASUREMENT REPRODUCIABILITY

Measurements on human beings are always more difficult to reproduce than measurements on 'dead' things. Making real ear measurements where the subjects sits passively in the same spot is used daily in the hearing clinics. The reproducibility seems to be ± 4 dB, [Hawkins, 1987]. When the subjects have to speak, the measurement is more complicated because it is not possible for the operator to control the speech signal and the reproducibility becomes worse. It is difficult to evaluate each measurement immediately after the measurement because the subject is waiting in the test room and gets tired if he/she has to sit passively after each performance. Of course, the subject needs some breaks, but in the breaks it is most polite to talk with the subject, and even more important, it might take a while to evaluate the measured data correctly. Hence, the measurement session has to be efficient and the subject must be 'entertained' all the time. Mistakes or, for example, a badly inserted earmould will not be detected until after the subject session. If something is wrong, then the subject must be called in again another day. For the same reasons, it is time consuming to do experiments with human subjects. This is the reason why many of the experiments in the present project are only performed as pilot studies with one or very few subjects.

12.3.1 Measurement method

The experimental method where one ear is measured at a time and a reference microphone is placed outside the ear was discussed in the first part of the Ph.D.-report [Hansen, 1997]. The repeatability of real ear measurements and the use of speech, vowels and chewing were commented upon. In most of the experiments used in the present report, the two ears were measured at the same time and it is relevant to make a few comments about this method because it does not seem to be described in the literature.

The principle of the measurement method was to measure the level in the right and left ear simultaneously. In preference to the more traditional method, where the occlusion effect is measured in one ear at a time, the new method has some advantages:

- Measurements of the right and left ear at the same time halves the measurement time.
- Simultaneous measurements makes it possible to obtain more reliable results especially for short duration sounds such as a sustained vowel. The speech signal is exactly the same in the two situations, e.g. occluded and open ear. If only one ear is measured at a time, the subject must repeat the same speech sound. It introduces an additional uncertainty because it is difficult to reproduce exactly the same sustained sound, especially when the ear is open in the first case and occluded in the next case, because one's own voice sounds differently in the two cases. Continuous speech is averaged over time and is therefore more reliable to repeat, but the subjects have a tendency to lower their voice when the ear is occluded [Hansen, 1997]. Hence, a reference microphone placed outside the ear is required.
- The phase difference between right and left ear (occluded and open ear) can be determined.
- The occlusion effect as a function of time is detected simultaneously. In contrast, if the occlusion effect is measured in one ear at a time, the occlusion effect does not become available as a function of time, because the subject does not repeat the spoken text with the same rhythm and speed.

The method has also some disadvantages:

- If the right and left ear are not reasonably anatomically symmetrical and the function of the right and left middle ear are not the same, then the method cannot be used.
- The subject might feel that it is more unpleasant to have earmoulds and probe microphones in both ears at the same time.
- This method requires 2 probe microphones that have the same frequency characteristics and if not, then a compensation must be made.

Here are some practical tips especially pointed to the bilateral method and the use of one's own voice:

- Try to only use the bilateral method if otoscopy confirms that the ears look reasonably symmetrical. If, for example, one ear has a sharp bend and the other has not, the method should not be used. The tympanogram should also be reasonably the same for the right and left ear.

- Make sure that the earmoulds are put in correctly.
- Make sure that the probe tube is at least 3 mm beyond the inner earmould tip.
- Make sure that the probe tube in the open ear does not point into the ear canal wall. A trick is to 'glue' the probe to the ear with a very tiny lump of impression material at the entrance to the concha. Then it is possible to adjust the tube to a point into the middle of the ear canal.
- Continuous speech should be averaged over at least 30 seconds. The text shall include various portions of unvoiced and voiced sounds.
- A sustained vowel or nasal sound should be averaged over 3 seconds.
- Chewing gum or crisp bread should be averaged over about 3 min. In the present project, a very short average time was used and the repeatability was not very good.

Always repeat the measurement because it does happened that the earmould is not inserted correctly or that the subject can make some strange noises during the recording.

12.3.2 Variation in individual fitting of the earmould

An example of how different acrylic moulds can fit into the right and left ear is illustrated in **Figure 12.7**.

If stimulus is one's own voice, the differences are generally larger and the difference does not agree with the difference for the loudspeaker measurements. This is seen by comparing **Figure 12.7** and **Figure 12.8**. For example, the real ear difference is negligible for subject 4 with loudspeaker stimulus but one's own voice gives a difference up to 8 dB. These measurements were made in the same session and the earmoulds were *not* touched between the two measurements. It illustrates very well that using the speech organ and the jaws gives quite different results than when the mouth is closed and the subject is passive.

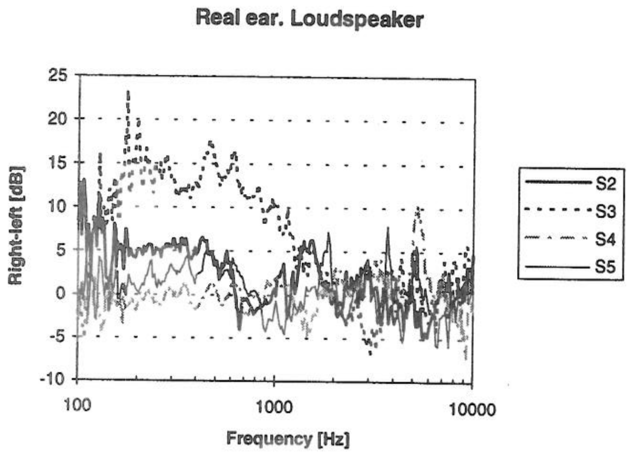


Figure 12.7 Difference between the sound pressure in the occluded right and left ear. Standard acrylic moulds. Stimulus: loudspeaker.

This is also another argument for using the subject's own voice to measure the occlusion effect and not a bone conductor placed on the skull as one could be tempted to do. The advantage with the bone conductor is that the signal is under nearly full control and that the subjects shall do nothing except sitting still. But as seen here, the situation can be quite different.

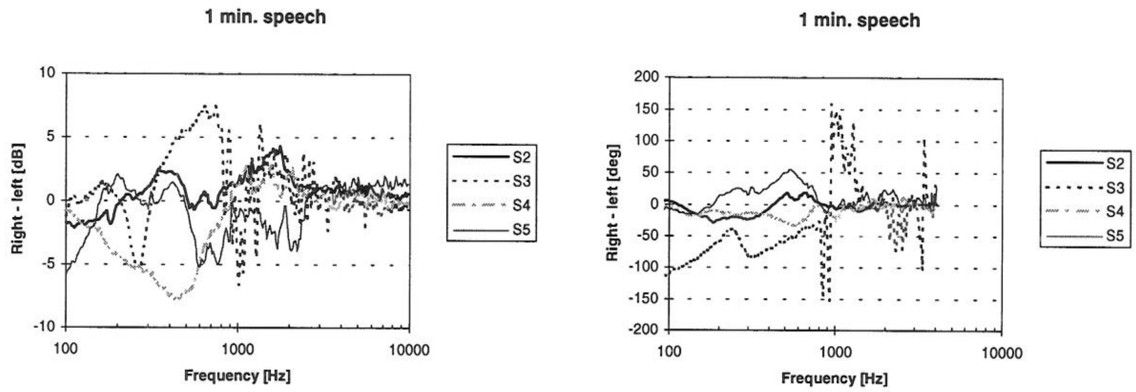


Figure 12.8 Difference between the sound pressure in the occluded right and left ear. Standard acrylic moulds. Subject's own voice.

The differences are most likely due to the fitting of the earmould and anatomical differences between the left and right ear. For example if the right mould is slightly more loose than the left mould, the dynamic leakage will be greater in the right ear.

- Based on the previous discussion and experience obtained during the project it is recommended that the bilateral method be used in experiments where the subject has to make the stimulus.

12.4 A HEARING AID

The occlusion effect is a problem for many hearing aid users and the influence of a hearing aid shall be analyzed briefly. The hearing aid introduces a third sound path to the occluded ear via the transducer placed in the inner tip of the earmould. The transducer output depends entirely on the signal processing and electronic components in the hearing aid. Furthermore, the electrical amplification in the hearing aid is adjusted to fit the hearing loss for the individual user. So for the present purpose it does not seem worth while to try and model a real hearing aid. Instead a very simple method is chosen. The transducer membrane is modeled as an ideal volume velocity source. If the hearing aid gain is 0 dB, the velocity source is defined to generate a sound pressure in the occluded ear canal equal to the sound pressure in the open ear canal when the sound source is airborne sound traveling into the ear canal. The reference point for the sound source is thought to be placed in the middle of the concha (see **Figure 11.1**)

If the hearing aid gain is 20 dB, then the hearing aid transducer shall produce a sound pressure in the occluded ear that is 20 dB higher than in the open ear, see **Figure 12.9**. The phase is the same as in the open ear. The curve for the occluded ear follows the open ear curve exactly at frequencies below 4 kHz. At higher frequencies there are some deviations. The reason is that the simulated hearing aid was designed to generate exactly the same sound pressure at the inner tip of the earmould whereas the measurement point is located 3 mm deeper into the ear canal than the earmould tip, see **Figure 11.1.**

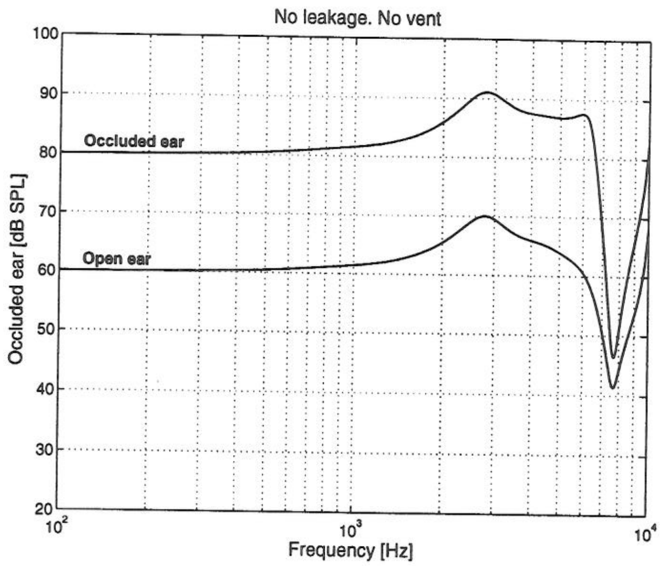


Figure 12.9 Estimated sound pressure in the open ear and the occluded ear with hearing aid. The sound pressure is estimated at 10.8 mm inside the ear canal. External sound source = 60 dB SPL. Hearing aid gain = 20 dB. Default values.

The occlusion effect from one's own voice is not the same when a hearing aid is switched on (HAon) and when there is only an earmould (the hearing aid is switched off, (HAoff)), see **Figure 12.10**. When the hearing aid is switched off, the sound pressure in the occluded ear is mainly generated by the body conducted sound. When the hearing aid is switched on, it is in principle no longer a pure occlusion effect because in addition to the body conducted sound, an electronic transmitted sound is present too. This situation is sometimes referred to as 'amplusion'. Some examples of the sound pressure difference between the situation with a hearing aid switched on and the hearing aid switched off when a person speaks is shown in **Figure 12.10**. Below 300 Hz the amplusion is only about 3 dB larger than the occlusion effect when the hearing aid amplifies 20 dB. At higher frequencies, the contribution of the body conducted sound decreases significantly but the hearing aid transmitted sound is the same or even higher at higher frequencies. The difference between Haon and Haoff becomes therefore greater towards higher frequencies.

The point to make is that at low frequencies (f < 500 Hz), the sound pressure in the occluded is about the same when the hearing aid is turned off and on, at least when the hearing aid amplifies up to 20 dB. This theory is in agreement with the results from the user survey of hearing aid users reviewed in chapter 2.

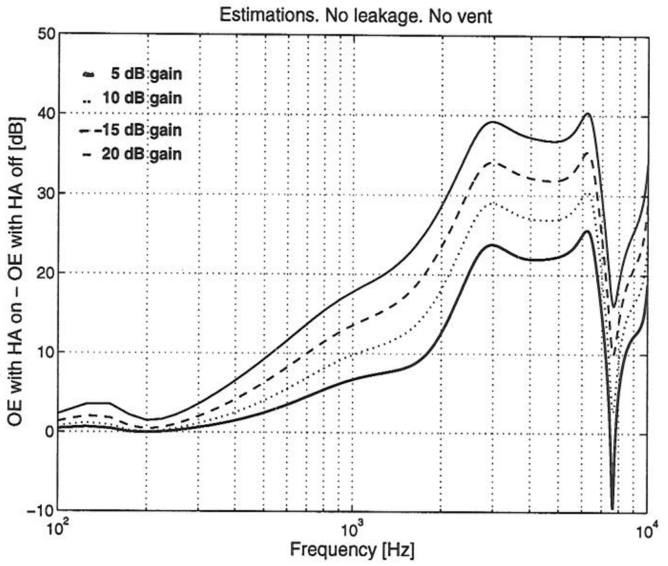


Figure 12.10 Estimated difference in occlusion effect (OE) from one's own voice with hearing aid switched on (HA on) and with the hearing aid switched off (HA off). Hearing aid gain is the same at all frequencies. Default model values. Mould mass = 3 g. $Z_{fit} = 0$, $Z_{mt} = Z_{flesh} \cdot (A2)/(A1+A2)$.

In practice the hearing aid gain is often small at low frequencies and one of the results from a user survey [Hansen, 1997] was that the 'amplusion' is often the same as the occlusion effect below 800 Hz. In these real situations the hearing aid moulds had leakage and vents, so the situation cannot be compared exactly with **Figure 12.10**. The influence of a vent in the mould will be analyzed in chapter 13.

12.5 SHORT TIME OCCLUSION EFFECT

Finally an important point to emphasize is that the mathematical model operates in the frequency domain. If the model should be useful in hearing aid design, it must be able to predict the occlusion effect also as a function of time. So, knowledge of the long term (30-60 seconds) average speech is not adequate. As shown previously, the occlusion effect is not the same for the sustained vowel /aaa/ and the vowel /eee/, and it is not the same for other vowels as well. Hence, it is necessary to look at the short time occlusion effect as well because a speech signal fluctuates in time.

The short time occlusion effect does not exactly equal the long time averaged occlusion effect because it depends on the relative sound pressure created by air conduction and body conduction, see chapter 10. This ratio is determined by how much sound is radiated from the mouth and the nose and therefore the position of the jaw and the lips influence the air-to-body-conduction ratio.

13. REDUCTION METHODS

Many people need to wear earmoulds every day and it may be expected that quite a large number of these people are annoyed by the occlusion effect. The user investigation (reported in the Occlusion effect Part I [Hansen, 1997]) showed that about 70% of the hearing users in the specific sample detected that their own voice sounded differently in the daily use of their hearing aids likely due to occlusion.

If the occlusion effect can be eliminated it might help many people. This chapter does not come up with a new and revolutionary method. The purpose here is to make a few comments on existing methods to reduce the occlusion effect and to sum up some of these less known attempts that have been done over the years. Direct reduction methods can be divided into 3 categories: ventilation control, deep insertion and electronic compensation. The advantages of the deeply inserted earmould has already been illustrated several times in this report, and it will not be commented further here.

13.1 EARMOULDS WITH VENT

13.1.1 "No mould" and open mould

The extreme situation is the "no mould" configuration, [Berland, 1975]. In a "no mould" hearing aid, the amplified sound is transmitted to the ear canal via a tube placed in the ear canal. It can though be rather difficult to keep the tube in place when the person speaks or chew. Another configuration is the open mould where the sound tube is kept in place with a minimum of a mould, like a ring. The really great problem with "no moulds" and open moulds is the risk of acoustic feedback. Acoustic feedback occurs when the feedback signal has a certain level and the feed-back signal is detected by the hearing aid microphone in phase with the original input signal. The feedback is heard as a pure tone (howl) around 2-4 kHz. However, it is possible to use an open mould and still be successful, [Curtois, 1988]. The open mould is not widely used partly due to feedback problems and partly due to the fact that most hearing aid users need some gain at low frequencies and a hearing aid with an open mould cannot provide enough gain.

13.1.2 Conventional vents

A conventional vent is by far the most prevalent method of reducing the occlusion effect. There are three types of conventional vents; parallel vent, diagonal vent and cavity vent. The parallel vent is a tube from the outer faceplate of the mould to the inner tip. Hearing aids are often produced with a vent before it is fitted to the user. The vent can then be adjusted either by a plug with a hole in the middle placed in the outer most part (Select-A-Vent (SAV)) or by insertion of PVC-tubes with various diameters in the outer part of the vent (Select-a-tube (SAT)).

13.1.3 The occlusion effect model used with a vented mould

In relation to the present Ph.D.-project, a master thesis project was made with the purpose of analyzing the occlusion effect from one's own voice as a function of vent size, [Thorup, 1996]. These data will be used to show how well the occlusion effect model performs with a vented earmould. A parallel vent is easily implemented in the model by a small tube put in parallel with the leakage.

First, it is necessary to look at the non-vented mould in order to find an estimate for the natural dynamic leakage. The best fit for the non-vented mould is obtained with a natural leakage of width 18 mm and thickness 0.2 mm. In **Figure 13.1** the estimation for the non-vented mould and three vent diameters are shown. The estimations are very good up to 4 kHz for the smallest vent and the non-vented mould.

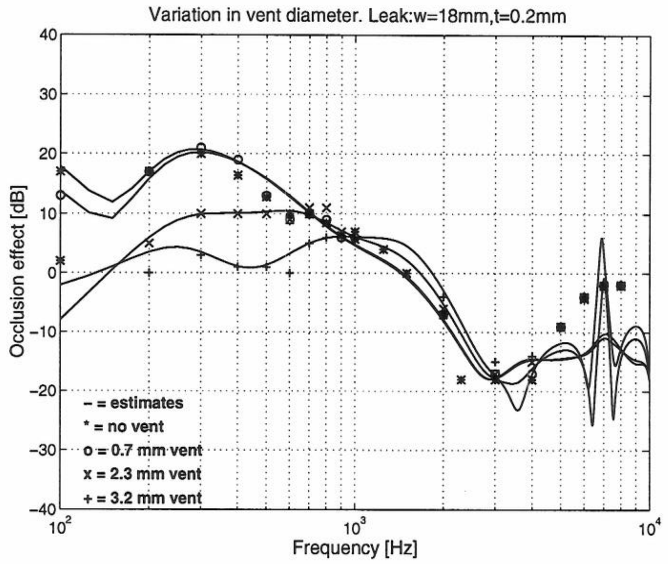


Figure 13.1 Effect of vent size. Estimations (solid lines) and measured data, subjects' own voice, continuous speech, [Thorup, 1996]. Natural leakage: w=18 mm, t=0.2 mm, length = BTE mould. Mass = 3 g. Default values, new pcpa. $Z_{fit} = 0$, $Z_{mt} = Zmt \cdot (Amct)/Ao$.

There is nothing surprising about the data printed in **Figure 13.1** but an important point to make is that the natural leakage should not be neglected. If the natural leakage is neglected (presuming a completely tight mould) the estimated occlusion effects will be too large compared to the actual measured occlusion effects as illustrated in **Figure 13.2**. This becomes especially clear if **Figure 13.1** and **Figure 13.2** are compared.

The leakage is not negligible with a 2.3 mm vent and with a 3.2 mm vent, the leakage gives 3 dB less occlusion effect at 400 Hz. Below 200 Hz, the occlusion effect is the same whether or not leakage is considered. These figures are only examples, and the results depends on the specific size of the leakage.

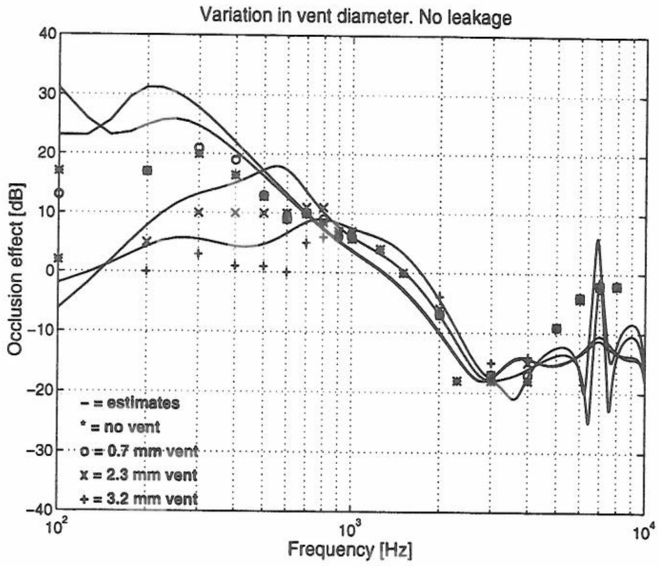


Figure 13.2 Effect of vent size. Estimations (solid lines) and measured data, subjects' own voice, continuous speech, [Thorup, 1996]. No natural leakage. Vent length = BTE mould. Mass = 3 g. Default values, new pcpa. $Z_{fit} = 0$, Z_{mt} = Zmt·(Amct)/Ao.

13.1.3.1 Occlusion effect as a function of vent size

A rule of thumb says that a vent should be in the order of 4 mm to eliminate the occlusion effect completely and it corresponds well with the data in **Figure 13.1**. According to the model estimates, the leakage cannot be neglected, not even for a large vent with a 3.2 mm diameter, but the influence of the leakage is, of course, smaller with a large vent than with a small vent. Consequently, the occlusion effect greatly depends on the leakage. This explains why there is no correlation between the occlusion effect and the vent size for moderately small vents.

In part I of this Ph.D.-project the occlusion effect was measured on a sample of hearing aid users, [Hansen, 1997]. The occlusion effect was measured with the hearing aids turned off and the occlusion effect was compared to the individual vent size[1]. The hearing aids represented vent sizes from 0 mm to 5 mm in diameter with various lengths. In order to be able to compare the vent diameters, they must be normalized with the vent length by the following formula, [Lundh, 1994]:

$$d_{normalised} = d\sqrt{\frac{l}{l_{reference}}} \qquad (13.1)$$

where;
d = vent diameter
l = vent length

If the vent length is assumed to be 23.3 mm as used for the default BTE mould, the occlusion effect plotted against the normalized vent diameter is as shown in **Figure 13.3** upper panel. In general, it looks like the occlusion effect at 125-250 Hz is reduced to 5 dB

[1] (page 6.19 and page 7.5 in [Hansen, 1997].

when the vent diameter is greater than 2.5 mm, but at 250-500 Hz, the occlusion effect is *not* reduced so much. In this case, it is possible that the occlusion effect is 10-15 dB even with vent diameters larger than 4 mm.

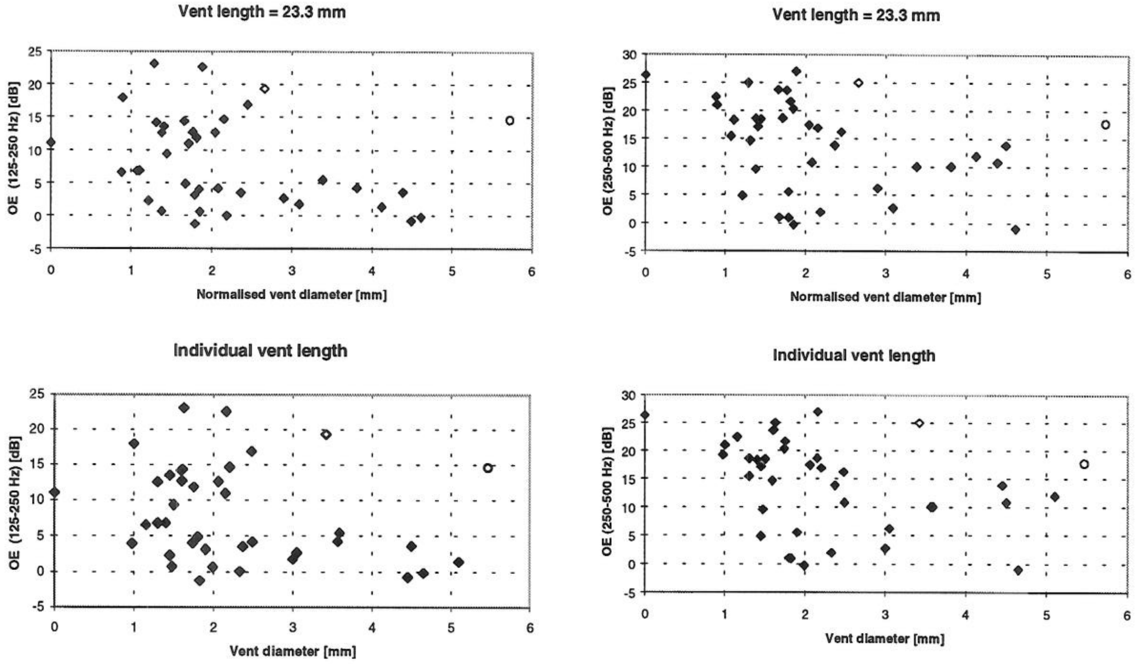


Figure 13.3 Measured occlusion effect against vent size. 38 subjects. Subjects' own voice. The non-filled points were described as outliners in [Hansen, 1997][2]. The occlusion effect is given as the average level within the specified frequency range. Upper panel: normalized vent diameters. Lower panel: individual vent diameters and lengths.

If the non-normalized vent diameters are plotted, the picture does not change much. The occlusion effect is still about 10 dB for very large vents (250-500 Hz). This observation is interesting because it shows that the occlusion effect is not necessarily eliminated with a large vent.

However, it does not mean that a 4 mm vent cannot reduce the subjective annoyance experienced by the hearing aid user. The situation is namely more complicated when the mould is used in a hearing aid, partly because the person wearing the hearing aid is hearing impaired and might not be so concerned about occlusion effect above a certain frequency and partly because the hearing aid amplifies the air borne sound of one's own voice.

13.1.4 'Occlusion effect' with vent and hearing aid

A hearing aid was introduced in chapter 12 and, as explained, the 'occlusion effect' when the hearing aid is switched on, then it is not a pure occlusion effect but a combination of the occlusion effect and electronically amplified sound. This situation could be termed 'amplusion'. In chapter 12 the amplusion was estimated for a very tightly fitted mould.

[2] Notice, that the other outliner in figure 6.7 in [Hansen, 1997] is not plotted as an outliner here, because it is not so distinct when the occlusion effect and vent is compared this way.

An example of the combination of hearing aid gain and vent influence on the sound pressure in the occluded ear is illustrated in **Figure 13.4**. The figure shows the difference in the occlusion effect when a hearing aid is swicthed on (OEHAon) and when the hearing aid is switched off (OEHAoff). In other words, it is the difference in the sound pressure in the occluded ear. The occlusion effect is only effective below 1-2 kHz and at the sound pressure at higher frequencies is due to hearing aid amplification.

At low frequencies (f < 500 Hz) the difference is practically independent of the vent diameter at least for diameters of 3 mm or smaller. For the 4 mm vent the resonance magnitudes become more noticeable. The phase difference grows with frequency until the first vent resonance where it shifts 180°.

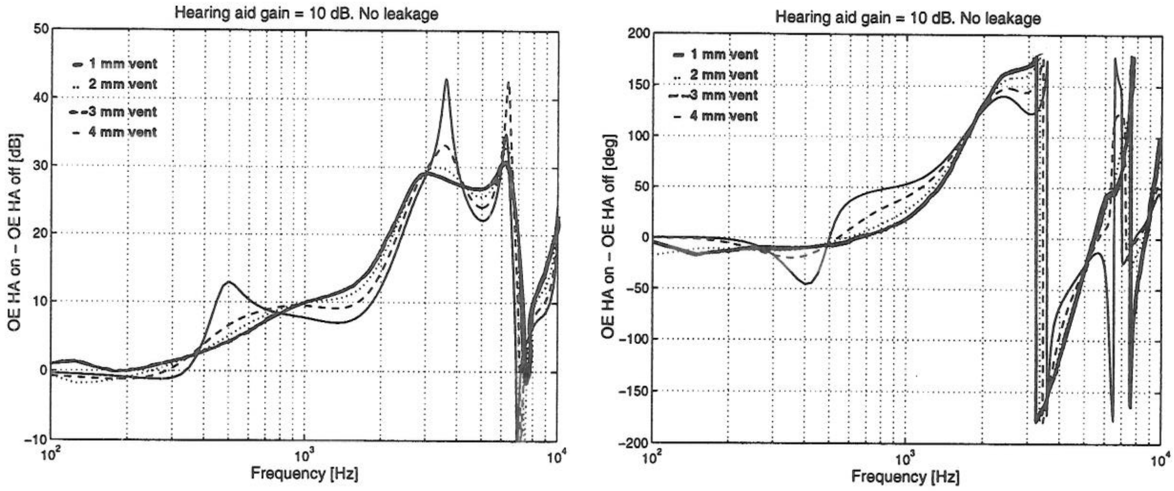


Figure 13.4 Estimated difference in occlusion effect with hearing aid switched on (OEHAon) and hearing aid switched off (OEHAoff). Hearing aid gain is 10 dB at all frequencies and flat in insertion gain. Variation in vent diameters. Vent length = 21.5 mm. Default model values. Mould mass = 3 g. Z_{fit} = 0, Z_{mt} = Zflesh·(A2)/(A1+A2).

Figure 13.4 illustrates also why there is a risk of acoustical feedback when the vent diameter is enlarged. The resonance peak between 3 and 4 kHz is considerably larger for a 4 mm vent than a 2 mm vent. Because of feedback, the vent cannot always be made large enough to reduce the occlusion effect. If the feedback could be eliminated, then the occlusion effect could also be reduced because it would be possible to make a larger vent. There are some methods that reduces the risk of feedback but there is no really efficient method to cancel the feedback, at least not without creating other problems.

One vent does not create feedback and that is the cavity, also called the 'Macrae' vent, [Macrae, 1985]. The vent is designed as a cavity cut in the earmould with a tube opening in the inner tip of the mould and a smaller tube opening to the outer faceplate of the mould. In principle this vent should reduce the occlusion effect. The problem with this vent is that it requires much space in the earmould.

Another solution is to place the hearing aid microphone on one ear and then transmit the sound to the other ear. This technique can be used with a very large vent without causing feedback. The disadvantage is that it is very difficult to localize a sound because a sound heard in the right ear will come from the left side and vise versa.

13.2 ELECTRONIC METHODS

The feedback problem has been studied quite a lot and development of electronic feedback cancellation has been of interest. In contrary, the use of an electronic method to reduce the occlusion effect has not been offered so much attention. However, in the literature it is possible to find at least two very different attempts to use an electronic method. Since it seems that they are not very well known, a short summary of the two methods will be given here.

13.2.1 Electronic vent

As described above, an acoustical vent will reduce the occlusion effect, provided that it can be made large enough. The problem is that it is sometimes impossible to do that. If the vent could be simulated electronically the acoustical feedback would not occur. Schweitzer and Smith [1992] have developed a so-called electronic vent. The vent is implemented in a circuit module called 'Active Vent 1' and inserted in the hearing aid between the microphone and the hearing aid amplifiers. The vent modifies only frequencies below 2 kHz.

In principle, the vent works by adjusting the phase of the microphone signal so that the hearing aid amplified sound in the ear canal is out of phase with the body conducted sound in the ear canal. Thereby, the body conducted sound is canceled out. It is done in practice by adjusting the phase of the vent until the subject feels that one's own voice sounds naturally. The electronic vent has been fitted to 2500 persons. The reference does not give any examples on how much the occlusion effect was reduced.

13.2.2 Electronic cancellation

The idea with electronic cancellation is, roughly speaking, to control the electronic output signal in the hearing aid so that the resulting sound pressure in the occluded ear canal equals the sound pressure generated by the hearing aid transducer only - this method was designed for unvented earmoulds. The contribution of the body conducted sound should be totally eliminated. The method used by Langberg et al. [1996] was to place an extra microphone next to the output transducer in the hearing aid. This internal microphone picks up the sound in the occluded ear canal and by means of an electronic feedback circuit, the signal to the output transducer was adjusted. Theoretically it should work, but a field test showed that in most cases, the occlusion effect was only reduced by a few dB and most of the users were still annoyed by occlusion from one's own voice.

13.3 CONCLUDING REMARKS

A considerable number of people have to live with the occlusion effect every day. Some of these people, but *not* all, can be helped by traditional methods, like a vent, or newer methods, with a deeply inserted hearing aid. Hence, there is no really efficient method that can be applied to all mould configurations if one wants to avoid the deeply inserted earmoulds and avoid acoustic feedback.

The present Ph.D.-project has inspired two further studies about how the occlusion effect may be reduced by electronic means. This work is still at a very preliminary stage and will not be reported here.

14. CONCLUSION

A result from the first part of this Ph.D.-report was, first of all, that hearing aid users have a problem with occlusion effect from one's own voice. In the study, 73% of the subjects answered that they experienced a change in one's own voice when they used their hearing aid and it is most likely that the change is caused by the occlusion effect. The study showed that the experienced occlusion effect generated by one's own voice is significantly correlated with the objective occlusion effect, which was measured in the ear canal. It is therefore worthwhile to try and eliminate the occlusion effect. In order to do that effectively, it is necessary to study the physical mechanism of the occlusion effect.

Therefore the aim of the study in the present report was to develop a better theory of the occlusion mechanism and to develop a mathematical model of the occlusion effect.

Literature on bone conducted sound was analyzed and applied to one's own voice. Almost all literature on bone conducted sound addresses the situation where a bone conductor is placed on the skull of the subject. This situation cannot uncritically be adapted to sounds produced with the speech organ because the bone conductor stimulates the skull more than one's own voice. The sound pressure in the ear in the two situations is also different because a bone conductor (properly shielded) only produces bone conducted sound in the ear canal.

A pilot study was performed during a stay at the National Research Council, Ottawa, Canada. This study served the purpose to find the relative amplitude between the sound transmitted through the body and the air borne sound radiated from the mouth and nose generated from one's own voice. The pilot study was followed up by a master thesis project, where 10 subjects attended. The contribution of the body conducted sound to the total sound pressure in the open ear canal is about 13 dB lower than the air conducted sound below 1 kHz. The body conducted sound lags behind the air conducted sound because the sound velocity is smaller in human tissue than in air. The present measurements seems to be the first published since Békésy's measurements in 1949. A more untraditional experimental method was used where the signal in the right and left ear was recorded simultaneously. This method can also be used to measure the occlusion effect and it is especially an advantage if sustained vowels are used or to determine the variation in time.

Furthermore, it was found that the ratio between the body and air conducted sound depends on the sound, for example the ratio is lower for sustained /aaa/ than /eee/. The ratio depends on the position of the tongue, lips and mouth opening. Further studies are needed to decide how the ratio varies with different speech sounds.

One's own voice sets the skull into vertical vibrations but it is mainly the soft cartilage in the ear canal that generates sound pressure. The vibrations of the cartilage was measured on 1 subject with a vibrometer placed on the soft ear canal wall. Simultaneously, the sound pressure in the occluded ear canal was detected while the subject talked. The signal picked up by the vibrometer corresponded well with the sound pressure in the occluded ear canal, except at the lowest frequencies, where the occluded ear sound pressure is lower, most likely because of leakage between the earmould and the ear canal wall.

The influence of the mechanical properties of the earmould was analyzed using an acrylic standard mould and a foam plug. An experiment was performed with 4 subjects. The sound pressure was measured in the ear canal occluded with an individually moulded acrylic mould and a pre-produced foam plug respectively. The acrylic mould gave up to 15 dB higher sound pressure than the foam plug both when the subject read a text aloud for 1 min., made a nasal sound with closed mouth and chewed gum. The difference in sound pressure was different for the 4 subjects but in all cases, the acrylic mould gave a higher sound pressure.

In addition to these measurements a pilot experiment was performed on 1 subject where a light (1 g) and a heavy (5 g) hard acrylic mould was compared. The heavy mould gave up to 15 dB higher pressure in the occluded ear than the light mould.

The conclusion from these measurements was that the mass and the elasticity of the mould does influence the occlusion effect. Several hypotheses can be made about the mechanical mechanism. One hypothesis was implemented in a mathematical model. The model is able to estimate the measured difference between an acrylic mould and a foam plug quite well.

The difference between the two mould types are also caused by the different fittings of the moulds, because the moulds provides different leakages. Natural leakage was implemented in the model as a narrow slit and the leakage is different for the situation where the person is doing nothing and when the person talks or just moves the jaws. The leakage can also explain the observed difference between the two mould types quite well.

A mathematical model based on acoustic models was developed. In comparison to previous models, this new model includes the mechanical properties of the earmould, a natural dynamic leakage and most important, it was designed especially to model one's own voice as a sound source. It means that an air conducted and a body conducted sound source is present at the same time, whereas previous models only operate with either air conducted or body conducted sound. The model is able to predict the average occlusion effect with an uncertainty of 5 dB. The individual occlusion effect is difficult to predict because there are many variable parameters, especially the fitting of the earmould.

The fitting of the earmould decides the amount of natural leakage. Especially the dynamic leakage will be very individual, but this area is not very well studied and further studies are needed to make a better model of dynamic leakage.

The overall conclusion is that this study contributes in some points to a better understanding of the mechanism of the occlusion effect from one's own voice and that a new mathematical model has been developed. The model is new compared to previous models, in the way that this model is designed especially to include the sound from one's own voice through the air and the body. Even though the aim of the study, in this sense, has been fulfilled, the study provides a basis for further studies in body conducted sound from one's own voice, dynamics of the ear canal tissue during speech, and earmould fitting.

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A. APPENDIX: MATLAB FILES AND 2 PORTS

A.1 MATLAB PROGRAM 'OESIM'

The program that calculates the occlusion effect is printed below. The program was written to be used for research purposes only, therefore nothing is done to make a good user interface.

```
% FILENAME: oesim.m
% Calculates the complex sound pressure in the open ear canal and in an ear canal
% occluded with a rigid earmould. Options are: slit leakage and parallel vent
%------------------------------------------------------------------------


clear;
tempcel=20; % Temperature in Celsius
rho=1.29230*(273.15/(273.15+tempcel))*(101325/101325); % air density
c=331.45+0.607*tempcel; % sound velocity
rhoc=rho*c;
rhoc2=rho*c^2;

% Inputs
%########
ccolor=input('Curve color: ','s');
constants=input('Do you want default values (1) or user values (2) ?');
if constants==1,
        dv=7.1e-3; % Average ear canal diameter
        lcd=23.5e-3; % Physical length of ear canal (eardrum: length = 0 )
        ladd=0.85*(dv/2); % Additional length of ear canal
        lbp=3e-3; % Length probe to earmould tip
else,
        rf=input('Enter the ear canal resonance frequency [kHz] :');
        rf=rf*1000;
        lfd=c/(4*rf);
        dv=input('Enter the average ear canal diameter [mm] :');
        dv=dv/1000;
        lcd=input('Enter the physical ear canal length [mm] :');
        lcd=lcd/1000;
        ladd=0.85*(dv/2); % Additional length of ear canal
        lbp=input('Enter the distance from the earmould tip to the probe [mm] :');
        if lbp==0
                lbp=0.01,
        end;
        lbp=lbp/1000;
end;

dc=22e-3; % Concha diameter
        lc=6e-3;  % Mid-concha depth
lcb=input('Enter insertion depth of the earmould (mm) :');
```

```
        lcb=lcb/1000;

mould=input('Enter the mould configuration BTE = 1, ITE = 2, ITEC = 3, your choice = 4 :');
        if mould == 1,
                lab=23.3e-3;
        elseif mould == 2,
                lab=21.5e-3;
        elseif mould == 3,
                lab=20.2e-3;
        elseif mould == 5,
                tmp=input('Enter mould length in mm:');
                lab=tmp/1000;
        end;

M=input('Enter the mass of the earmould [g] :');

% Leakage and vent
wd=input('Enter the width of the slit [mm] : ');
wd=wd/1000;
t=input('Enter the thickness of the slit [mm] :');
t=t/1000;
tdin=input('Enter the diameter of the vent [mm] :');
if tdin == 0
        td=0.01/1000;
else
        td=tdin/1000;
end;

% Frequency
%##########
f=(1:400)*25;
w=2*pi*f;
k=w./c;

% Ratio body conducted sound to air borne sound
%################################################
pcpaph=-0.1077.*f-150.73;
pcpam=10.^(-13/20);
pcpaph=pcpaph*(pi/180);
pcpa=pcpam.*exp(j*pcpaph);

% Lengths and areas
%##################
lbd=lcd-lcb; % Length tip of earmould to eardrum
lpd=lbd-lbp; % Length probe to ear eardrum
lfd=ladd+lcd; % acoustical length of ear canal
lfp=ladd+lcb+lbp; % Length entrance to probe open ear
lfb=lcb+ladd; % Length entrance to tip of earmould open ear
lct=lcd/2; % Length of cartilage tissue
lbt=lcd/2; % Length of bony tissue
```

```
Act=2*pi*(dv/2)*lct; % Surface of the cartilage tissue
Abt=2*pi*(dv/2)*lbt; % Surface of the bony tissue
if lcb > lct
        Amct=2*pi*(dv/2)*lct; % Cartilage covered by the earmould
        Ambt=2*pi*(dv/2)*(lcb-lct); % Bone covered by the earmould
else
        Amct=2*pi*(dv/2)*lcb;
        Ambt=0;
end;
if (lcb+lbp) > lct
        Apct=2*pi*(dv/2)*lct; % Outer Cartilage over probe
        Apbt=2*pi*(dv/2)*((lcb+lbp)-lct); % Outer Bone over probe
else
        Apct=2*pi*(dv/2)*(lcb+lbp);
        Apbt=0;
end;

Ao=2*pi*(dv/2)*lcd; % Surface of the ear canal
A=pi*(dv/2)^2;  % Cross-section area
Afree=Ao-Amct-Ambt;
Aopentot=2*pi*(dv/2)*lfd;
K=Afree/(Amct+Ambt);

% Tissue impedance
%################
% Wargowske, mastoid
Msw=2.58e-4;
Csw=1.09e-5;
Ksw=1./Csw;
Rsw=10.5;
Zfleshob=(Rsw+j*(w*Msw-Ksw./w))*((Abt)/175e-6)/A^2;  % Open ear canal

%Schroeter and Els 1982 ear canal, modified
Ms=0.5e-4;
Ks=5303;
Rs=3;
Zfleshoc=(Rs+j*(w*Ms-Ks./w))*((Act)/175e-6)/A^2; % Open ear canal

% Occluded ear
if lcb > lct
        Zfleshmc=(Rs+j*(w*Ms-Ks./w))*((Amct)/175e-6)/A^2; % Covered by the earmould
        Zfleshmb=(Rsw+j*(w*Msw-Ksw./w))*(Ambt/175e-6)/A^2;  % Covered by the earmould
        Zfleshc=0;
        Zfleshb=(Rsw+j*(w*Msw-Ksw./w))*((Ao/2-Ambt)/175e-6)/A^2; % Free bone
        Zfcover=parallel(Zfleshmc,Zfleshmb);
        Zffree=Zfleshb;
        Zmt=1e50; % Coupling impedance
else
        Zfleshmc=(Rs+j*(w*Ms-Ks./w))*(Amct/175e-6)/A^2; % Covered by the earmould
```

```
        Zfleshmb=0;
        Zfleshc=(Rs+j*(w*Ms-Ks./w))*((Ao/2-Amct)/175e-6)/A^2; % Free cartilage
        Zfleshb=(Rsw+j*(w*Msw-Ksw./w))*((Ao/2)/175e-6)/A^2; % Free bone
        Zfcover=Zfleshmc;
        Zffree=parallel(Zfleshc,Zfleshb);
        Zmt=Zfleshoc*(Amct/Ao); % Coupling impedance
end;

% Earmould
%#########
lm=K^2*(M/1000)/A^2;  % earmould mass
Zlm=j*w*lm;

% Standard impedances and 2 ports
%#################################
Zd=mdlear(f);  % Eardrum and middle ear
[Ad,Bd,Cd,Dd]=zend2p(Zd);
Zrad=zsphere(f,dc,rho,c); % Radiation impedance from a sphere
[Arend,Brend,Crend,Drend]=zend2p(Zrad); %Radiation from a sphere, end impedance
[Ar,Br,Cr,Dr]=zinser2p(Zrad); % Radiation from a sphere, in series

% Vent and leakage
%#################
% Vent
labt=lab+0.85*(td/2); % Acoustic length of leakage
alfa=absorp(f,lab,td); % Absorption from walls
katt=k-j*alfa;
[Atb,Btb,Ctb,Dtb]=tube2p(katt,rho,c,labt,td); % Leakage as a narrow tube
Zrt=((k*td/2).^2)*rho*c./(2*pi*(td/2).^2);
[Atr,Btr,Ctr,Dtr]=zend2p(Zrt);

% Slit leakage
labs=lab+0.85*sqrt((wd*t)/pi); % Acoustic length of leakage
Zs=zslit(labs,wd,t,f,rho); % Slit from natural leakage
[Aes,Bes,Ces,Des]=zinser2p(Zs);
Zrs=((k*t/2).^2)*rho*c./(2*pi*(t/2).^2); % Radiation from a flanged slit
[Asr,Bsr,Csr,Dsr]=zend2p(Zrs);

% Two port transmission
%######################
% Ear canal
[Apd,Bpd,Cpd,Dpd]=tube2p(k,rho,c,lpd,dv); % From probe to eardrum
[Abd,Bbd,Cbd,Dbd]=tube2p(k,rho,c,lbd,dv); % From tip of earmould to eardrum
[Afd,Bfd,Cfd,Dfd]=tube2p(k,rho,c,lfd,dv); % From entrance to eardrum, acoustic length
[Abp,Bbp,Cbp,Dbp]=tube2p(k,rho,c,lbp,dv); % From earmould to probe
[Afp,Bfp,Cfp,Dfp]=tube2p(k,rho,c,lfp,dv); % From entrance to probe, open ear
[Arc,Brc,Crc,Drc]=tube2p(k,rho,c,lc,dc); % Concha
% Open ear
[Aec,Bec,Cec,Dec]=mult2p(Ar,Br,Cr,Dr,Arc,Brc,Crc,Drc);  %From external to concha
[Aed,Bed,Ced,Ded]=mult2p(Aec,Bec,Cec,Dec,Afd,Bfd,Cfd,Dfd); % From external to eardrum
```

```
[Aedd,Bedd,Cedd,Dedd]=mult2p(Aed,Bed,Ced,Ded,Ad,Bd,Cd,Dd); % From external to eardrum end
[Apdd,Bpdd,Cpdd,Dpdd]=mult2p(Apd,Bpd,Cpd,Dpd,Ad,Bd,Cd,Dd); % From probe to eardrum impedance
[Abdd,Bbdd,Cbdd,Dbdd]=mult2p(Abd,Bbd,Cbd,Dbd,Ad,Bd,Cd,Dd);
                                             % From earmould plane to eardrum impedance
[Apr,Bpr,Cpr,Dpr]=mult2p(Afp,Bfp,Cfp,Dfp,Arc,Brc,Crc,Drc); % From probe to radiation
[Apee,Bpee,Cpee,Dpee]=mult2p(Apr,Bpr,Cpr,Dpr,Arend,Brend,Crend,Drend);
                                             %From probe to radiation end

% Occluded ear
if td == 0
      % Only slit leakage
      [Aesd,Besd,Cesd,Desd]=mult2p(Aes,Bes,Ces,Des,Abd,Bbd,Cbd,Dbd);
                                             % From outer end of slit to eardrum
      [Aesdd,Besbdd,Cesdd,Desdd]=mult2p(Aesd,Besd,Cesd,Desd,Ad,Bd,Cd,Dd);
                                             % From outer end of slit to eardrum impedance
      [Apaa,Bpaa,Cpaa,Dpaa]=mult2p(Abp,Bbp,Cbp,Dbp,Aes,Bes,Ces,Des);
                                             %From probe to outer end of slit in end
      [Apaar,Bpaar,Cpaar,Dpaar]=mult2p(Apaa,Bpaa,Cpaa,Dpaa,Asr,Bsr,Csr,Dsr);
                                             % From outer end of slit to radiation
else
      % Occluded ear with slit leakage and tube vent
      [Asout,Bsout,Csout,Dsout]=mult2p(Aes,Bes,Ces,Des,Asr,Bsr,Csr,Dsr);
                                             %From inner slit to radiation
      [Avout,Bvout,Cvout,Dvout]=mult2p(Atb,Btb,Ctb,Dtb,Atr,Btr,Ctr,Dtr);
                                             %From inner vent to radiation
      Zinout=parallel((Asout./Csout),(Avout./Cvout));
      [Asvr,Bsvr,Csvr,Dsvr]=zend2p(Zinout); %From inner to radiation
      [Apaar,Bpaar,Cpaar,Dpaar]=mult2p(Abp,Bbp,Cbp,Dbp,Asvr,Bsvr,Csvr,Dsvr);
                                             % From outer end of slit to radiation
      [Asvpa,Bsvpa,Csvpa,Dsvpa]=par2p(Aes,Bes,Ces,Des,Atb,Btb,Ctb,Dtb);
                                             % Vent and leakage in parallel
      [Aesd,Besd,Cesd,Desd]=mult2p(Asvpa,Bsvpa,Csvpa,Dsvpa,Abd,Bbd,Cbd,Dbd);
                                             % From outer end of slit to eardrum
      [Aesdd,Besdd,Cesdd,Desdd]=mult2p(Aesd,Besd,Cesd,Desd,Ad,Bd,Cd,Dd);
                                    % From outer end of slit to eardrum impedance
end;

% Air conduction source
%######################
Pea=10^(60/20);  % Pa/20e-6 Pa, reference phase is zero

% Sound pressure in the open ear canal
%#####################################
% Air conduction source
Ppao=Pea.*(Apdd./Aedd);

% Cartilage source
Ppco=pcpa.*Ppao; % Sound pressure created by cartilage in plane p
Popen=Ppao+Ppco;
```

```
% Estimation of volume velocity source when Pc/Pa is known
%###############################################################
Zpee=Apee./Cpee; % Impedance from Probe to Radiation
Zpdd=Apdd./Cpdd; % Impedance from Probe to eardrum
Zopen=parallel(Zpee,Zpdd);
qc=Ppco./Zopen;

% Sound pressure in the occluded ear canal
%#############################################
% Air conduction source
Ppcas=Pea.*Apdd./Aesdd;

% Impedance of ear canal occluded with a mould with leakage (and vent)
Zpaar=Apaar./Cpaar; % Rigid wall with slit
Zpdd=Apdd./Cpdd;
Zears=parallel(Zpaar,Zpdd);

% Cartilage source - mould inertia - source and probe at same point
Zfit=0;
Z1=Zears+Zffree;
Z2=Zfcover+Zlm+Zfit;
Ztot=(Z2.*(Zmt+Z1)+K^2.*Zmt.*Z1)./(Z1+Z2+(1+K)^2*Zmt);
Ptot=qc.*Ztot;
q0=Ptot.*(1-1/K)./(K*Zmt-(Zmt+Z1)/K);
P11=Ptot-q0.*Zmt;
Ppccs=P11.*Zears./Z1;
Pocc=Ppccs+Ppcas;

% Occlusion effect
%##################
% Mould with slit leakage and vent
Poe=Pocc./Popen;
OE=20*log10(abs(Poe));
OEp=angle(Poe)*180/pi;

% Figures
%########
subplot(121),h=semilogx(f,OE,ccolor);
set(h(1),'linewidth',1);
axis([100 10000 -40 40]);
grid;
xlabel('Frequency [Hz]','FontSize',14);
ylabel('Occlusion effect [dB]','FontSize',14);

subplot(122),h=semilogx(f,OEp,ccolor);
set(h(1),'linewidth',1);
axis([100 10000 -200 200]);
grid;
xlabel('Frequency [Hz]','FontSize',14);
ylabel('Occlusion effect [deg]','FontSize',14);
```

The program calls the matlab functions printed here below and in section A.2 and A.3.

```
% FILENAME: zslit.m

% Calculates the impdance of a narrow slit
% INPUT: length (l), width (wd), thickness (t), frequency (f)
% OUTPUT: impedance of split
%--------------------------------------------------------------

function z=zslit(l,wd,t,f,rho)

vis=1.83e-5;
w=2*pi*f;
z=(12*vis*l./(t^3.*wd))+j*w.*(6*rho*l./(5*t*wd));
```

```
function [Zar]=zsphere(f,dv,rho,c);

% Calculates the radiation impedance for a piston in a sphere
%  (Ref: Pierce, 1981)
% All values is given in standard units: kg,m,s
% The impedances are in acoustic units
% INPUT : Frequency (f), rho, c, ear canal diameter (dv)
% OUTPUT: Complex radiation impedance (Zar)
% Revised by Mie Oe. Hansen, 05-01-98
%-----------------------------------------------------------------

% Constants
av=dv/2;          % Ear canal radius. Used until the 18-09-96

% Variables
w=2*pi*f;
k=w/c;
kav=k*av;

% Radiation impedance
tmp=2*kav;
R=(1.5/2)*tmp.^2/(4*2);
X=kav*0.8;
Zar=rho*c/(pi*av^2)*(R+X*i);
```

A.2 TWO PORT THEORY

The model is based on two port representation of all acoustic transmission lines and impedances. The two port method makes it possible to construct a logical and elegant calculation structure and it is easy to add new or remove elements.

A.2.1 Transmission line

An acoustic transmission line is drawn in **Figure A.1**. The input is a volume velocity, q_i, and sound pressure, p_i, and the output is q_o, and sound pressure, p_o.

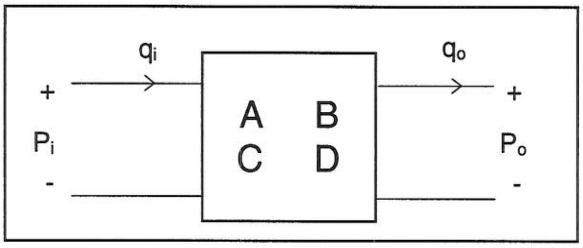


Figure A.1. Two port representation of an acoustic transmission line.

A uniform tube of length, l, and cross-section area, S, is an acoustic transmission line, see **Figure A.2**.

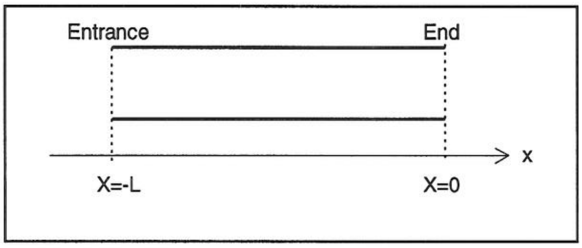


Figure A.2. Open uniform tube.

Seen from the entrance at x=-l, the two port parameters are:

$$A = \left.\frac{P_i}{P_o}\right|_{q_o=0} = \cos kl \quad ;$$

$$B = \left.\frac{P_i}{q_o}\right|_{P_o=0} = j\frac{\rho c}{S}\sin kl \quad ;$$

$$C = \left.\frac{q_i}{P_o}\right|_{q_o=0} = j\frac{S}{\rho c}\sin kl \quad ;$$

$$D = \left.\frac{q_i}{q_o}\right|_{P_o=0} = \cos kl \quad ;$$

<u>Matlab m-file:</u>

```
% FILENAME: tube2p.m

% Calculates the two port elements for a rigid uniform tube
% INPUT: frequency (k), rho (rho),speed of sound (c);
% length (l), diameter (dt)
% OUTPUT: two ports parameters (A,B,C,D)
%----------------------------------------------------------------

function [A,B,C,D]=tube2p(k,rho,c,l,dv);
```

```
S=pi*(dv/2)^2;
Z0=rho*c/S;
A=cos(k*l);
B=j*Z0*sin(k*l);
C=j*sin(k*l)/Z0;
D=cos(k*l);
```

A.2.2 Terminating impedance

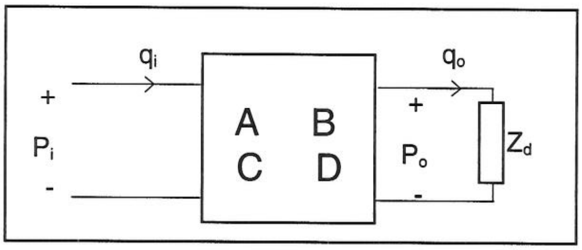


Figure A.3. Two port terminated with an impedance.

If a system is terminated with an impedance as in **Figure A.3**, e.g. the eardrum at the end of the ear canal, the two port representation of the impedance will be:

$$\begin{bmatrix} 1 & 0 \\ \dfrac{1}{Z_d} & 0 \end{bmatrix}$$

Matlab m-file:

```
% FILENAME: zend2p.m

% Calculates the two port parameters for a parallel impedance at the end of two port chain
% INPUT: lumped element impedance (z)
% OUTPUT: Two port parameters (A,B,C,D)
%--------------------------------------------------------------------------------------------

function [A,B,C,D]=zend2p(z)

A=ones(size(z));
B=zeros(size(z));
C=1./z;
D=zeros(size(z));
```

A.2.3 Impedance in series with pressure source

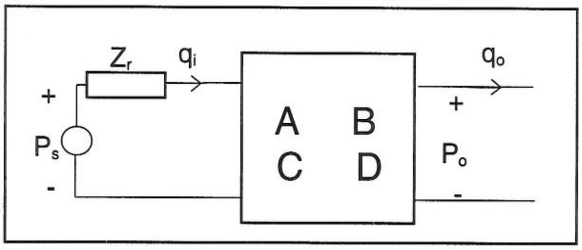


Figure A.4. Impedance in serie with pressure source.

The impedance in **Figure A.4** has the two port parameters:

$$\begin{bmatrix} 1 & Z_r \\ 0 & 1 \end{bmatrix}$$

Matlab m-file:

```
% FILENAME: Zinser2p.m

% Calculates the two port parameters for an entrance impedance
% in series with a pressure source
% INPUT: Impedance (z)
% OUTPUT: Two port parameters (A,B,C,D)
% -----------------------------------------------------------------

function [A,B,C,D]=Zinser2p(z)

A=ones(size(z));
B=z;
C=zeros(size(z));
D=ones(size(z));
```

A.2.4 Multiplikation

The version of Matlab which was used can only operate in two dimensions. The parameters A,B,C and D in a two port is an array with a number of elements equal to the number of frequency samples. The two port becomes then a three dimensional matrix, which Matlab cannot handle. It was therefore necessary to manually write matlab functions to handle two port operations. A multiplication is used when to find the total two port of two ports in series.

The total two port is:

$$\begin{bmatrix} A_1 & B_1 \\ C_1 & D \end{bmatrix}\begin{bmatrix} A_2 & B_2 \\ C_2 & D_2 \end{bmatrix} = \begin{bmatrix} A_1A_2 + B_1C_2 & A_1B_2 + B_1D2 \\ C_1A_2 + D_1C_2 & C_1B_2 + D_1D_2 \end{bmatrix}$$

Matlab m-file:

```
% FILENAME: mult2p.m

% Perform a two port parameter multiplication
% INPUT: Two port parameters of 2 two ports
% OUTPUT: The two port parameters of the multiplied two ports
%---------------------------------------------------------------------------

function [A,B,C,D]=mult2p(A1,B1,C1,D1,A2,B2,C2,D2)

A=A1.*A2+C2.*B1;
B=A1.*B2+B1.*D2;
C=C1.*A2+D1.*C2;
D=C1.*B2+D1.*D2;
```

A.3 BASIC CALCULATIONS

The most used basic calculations are derived here. Other calculations can be derived easily with the same principles as shown here.

A.3.1 Source and pressure measured at the entrance

A piston placed in one end of a tube generates sound inside the tube. The pressure at the piston, P_i, is the volume velocity source multiplied with the acoustic impedance seen from the entrance:

$$P_i = q_i Z_i$$

If the tube is terminated with an impedance, Z_D, the system looks like the one in **Figure A.3** and we get:

$$\begin{bmatrix} P_i \\ q_i \end{bmatrix} = \begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} P_o \\ q_o \end{bmatrix} \Rightarrow \begin{matrix} P_i = AP_o + Bq_o \\ q_i = CP_o + Dq_o \end{matrix}$$

The volume velocity through Z_d is:

$$q_o = P_o / Z_d$$

This is implemented in the expression above:

$$\left. \begin{matrix} P_i = (A + B/Z_d)P_o \\ q_i = (C + D/Z_d)P_o \end{matrix} \right\} \Rightarrow P_i = q_i \frac{A + B/Z_d}{C + D/Z_d} \Leftrightarrow P_i = q_i Z_i$$

The two port representation of the whole system is:

$$\begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} 1 & 0 \\ 1/Z_d & 0 \end{bmatrix} = \begin{bmatrix} A + B/Z_d & 0 \\ C + D/Z_d & 0 \end{bmatrix} = \begin{bmatrix} A' & B' \\ C' & D' \end{bmatrix}$$

It is seen that Z_i becomes:

$$Z_i = \frac{A'}{C'}$$

and the two port calculation is therefore:

$$P_i = q_i \frac{A'}{C'}$$

A.3.2 Source and pressure measured at the entrance with rigid termination

A two port representation of for example a closed tube with rigid walls is shown in **Figure A.5.**

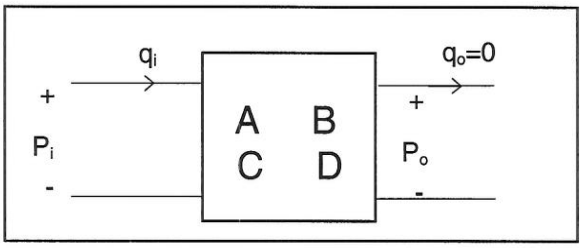


Figure A.5. Two port with acoustic rigid termination.

The volume velocity is zero on the rigid wall and tbe two port calculation becomes:

$$\begin{bmatrix} P_i \\ q_i \end{bmatrix} = \begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} P_o \\ q_o \end{bmatrix} \Rightarrow \left. \begin{array}{l} P_i = AP_o \\ q_i = CP_o \end{array} \right\} \Rightarrow Z_i = \frac{A}{C}$$

The same result could have been determined by using the two port representation of a terminating impedance and setting $Z_d = \infty$:

$$\begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} 1 & 0 \\ 1/Z_d & 0 \end{bmatrix} = \begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} 1 & 0 \\ 0 & 0 \end{bmatrix} = \begin{bmatrix} A & 0 \\ C & 0 \end{bmatrix}$$

A.3.3 Source and pressure measured at the entrance with soft termination

A two port representation of for example a tube closed with a very absorpbant material is shown in **Figure A.A.6.**

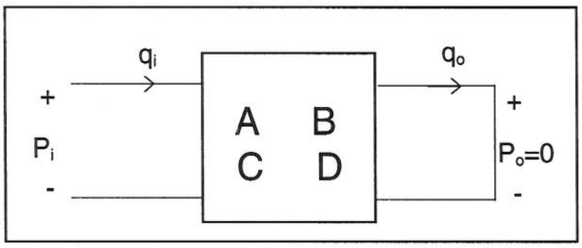


Figure A.A.6. Two port with acoustic soft termination.

The pressure across the soft wall is zero because the wall absorps all the sound waves and tbe two port calculation becomes:

$$\begin{bmatrix} P_i \\ q_i \end{bmatrix} = \begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} P_o \\ q_o \end{bmatrix} \Rightarrow \left. \begin{matrix} P_i = Bq_o \\ q_i = Dq_o \end{matrix} \right\} \Rightarrow Z_i = \frac{B}{D}$$

The same result could have been determined by using the two port representation of a terminating impedance and setting $Z_d = 0$ and using legal matrix operation:

$$\begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} 1 & 0 \\ 1/Z_d & 0 \end{bmatrix} = \begin{bmatrix} A + B/Z_d & 0 \\ C + D/Z_d & 0 \end{bmatrix} = \begin{bmatrix} AZ_d + B & 0 \\ CZ_d + D & 0 \end{bmatrix} = \begin{bmatrix} B & 0 \\ D & 0 \end{bmatrix}$$

A.3.4 Source at entrance and pressure measured inside tube

The system in **Figure A.7** could represent a piston in a tube, which could be the earmould moving in the ear canal. The point X is somewhere inside the tube.

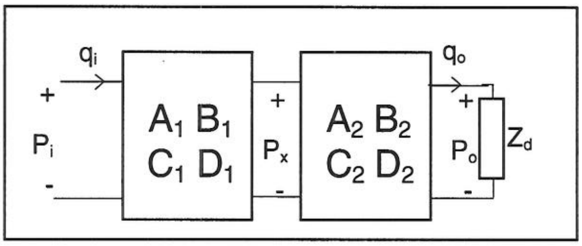


Figure A.7. Volume velocity source. Pressure measured in point X.

The two port of the whole system is:

$$\begin{bmatrix} A_1 & B_1 \\ C_1 & D_1 \end{bmatrix} \begin{bmatrix} A_2 & B_2 \\ C_2 & D_2 \end{bmatrix} \begin{bmatrix} 1 & 0 \\ 1/Z_d & 0 \end{bmatrix} = \begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} 1 & 0 \\ 1/Z_d & 0 \end{bmatrix} = \begin{bmatrix} A + B/Z_d & 0 \\ C + D/Z_d & 0 \end{bmatrix} = \begin{bmatrix} A' & B' \\ C' & D' \end{bmatrix}$$

The two port of the system from point X to Z_D is givenby A'',B'',C'',D'' and from the two port equations we get:

$$\left. \begin{matrix} P_i = \dfrac{A'}{C'} q_i = A' P_o \\ P_x = A'' P_o \end{matrix} \right\} \Rightarrow P_x = \frac{A''}{A'} P_i \Leftrightarrow P_x = \frac{A''}{A'} \frac{A'}{C'} q_i \Leftrightarrow P_x = \frac{A''}{C'} q_i$$

If the point x is moved to the entrance then A''=A' and we get $P_x=P_i=(A'/C')q_i$ which is the same equation as before.

If the point x is moved the end to Z_d then A''=1 and we get $P_x=P_o=q_i/C'$.

A.3.5 Source and pressure measured inside a tube

A volume velocity source placed inside a system and the pressure measured in the same plane is illustrated in **Figure A.8**. This source could be the vibration of the ear canal wall.

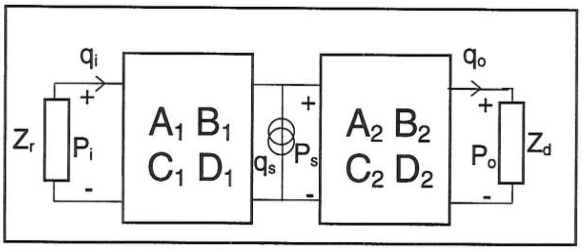


Figure A.8. Volume velocity source inside the tube.

Looking into Z_d the impedance is:

$$Z''=\frac{A''}{C''}=\frac{A_2+B_2/Z_d}{C_2+D_2/Z_d}$$

Looking out of the system towards Z_r, the impedance is :

$$Z'''=\frac{A'''}{C'''}=\frac{A_1+B_1/Z_r}{C_1+D_1/Z_r};$$

The system in **Figure A.8** represents a source in parallel with two 2-ports. Seen from the source point, qs, we look into two tubes with terminating impedances and the pressure P_s is found by:

$$P_S=q_s\frac{Z''Z'''}{Z''+Z'''}=q_S\frac{(A''/C'')(A'''/C''')}{(A''/C'')+(A'''/C''')}$$

The m-file to calculate two impedances in parallel is called parallel.m

Matlab m-file:

```
% FILENAME: parallel.m

% Calculates the impedance of two impedances in parallel
% INPUT : two impedances (r1), (r2)
% OUTPUT: the parallel impedance (rp)
%----------------------------------------------------------

function [rp]=parallel(r1,r2)

rp=r1.*r2./(r1+r2);
end;
```

A.3.6 Two 2-ports in parallel

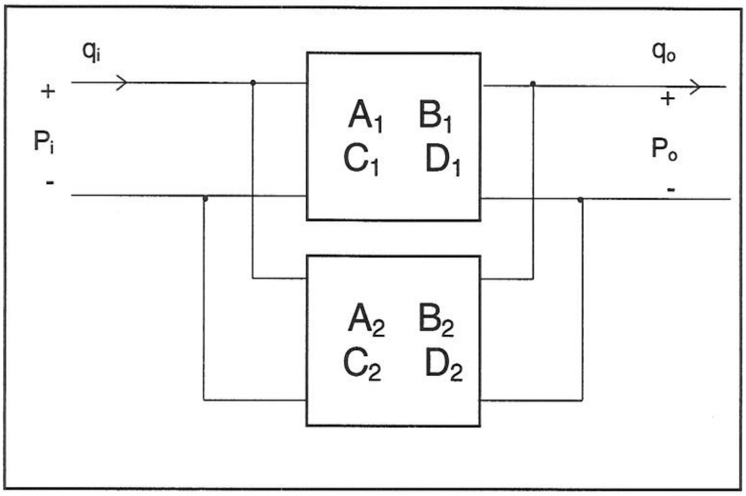


Figure A.9 Two 2-ports parallel coupled.

Looking from left to right in **Figure A.9** the equivalent two ports to the two parallel coupled 2-ports is determined as printed in the matlab file below.

Matlab m-file:

```
% FILENAME: par2p

% Calculates the 2 port of two parallel 2 ports
% INPUT: Two port parameters for port 1 and 2
% OUTPUT: The parallel two port parameters
%--------------------------------------------------

function [Ap,Bp,Cp,Dp]=par2p(A1,B1,C1,D1,A2,B2,C2,D2)

N=B1+B2;
Ap=(B1.*A2+A1.*B2)./N;
Bp=B1.*B2./N;
Cp=((B1+B2).*(C1+C2)-(D1-D2).*(A2-A1))./N;
Dp=(D1.*B2+B1.*D2)./N;
end;
```

B. APPENDIX: EXPERIMENT II

IMPEDANCE OF EARMOULD MATERIALS

B.1 LIST OF EQUIPMENT

Transducer	Tesla MA1 Electroacoustic transducer for measuring acoustical impedance
Coupler	Laboratory construction
Microphone	½" Brüel&Kjær, type 4134
Audio Analyzer	Brüel&Kjær, type 2012 Steady State response, Log ISO R40 sweep, complex averaging
Amplifier	

The earmoulds were put into the coupler and the coupler was screwed on the Tesla MA1 transducer. The sound pressure in the coupler cavity and the velocity of the piston in the transducer were measured with the Audio Analyzer Brüel&Kjær 2012 which has two inputs and outputs.

B.1.1 Earmould coupler

The earmould was inserted into a special designed device that could be screwed on top of the transducer. The coupler is a tube made of metal. In the end nearest to the Tesla transducer there is a cylindrical cavity with diameter 24.01 mm and depth 3.10 mm. On top of the volume is a cylindrical tube with diameter 9.52 mm and length 10.01 mm. In order to detect when the earmould is placed correctly a small ring was cut at the reference plane leaving a cylindrical hole with diameter 7.5 mm and depth of 1.47 mm between the cavity and the tube. The ring prevents that the earmould is pushed in beyond the reference plane. The system is calibrated by closing the cavity at the reference plane with a solid metal tube. The reference volume is in total 2.1 cm^3, see **Figure B.1**.

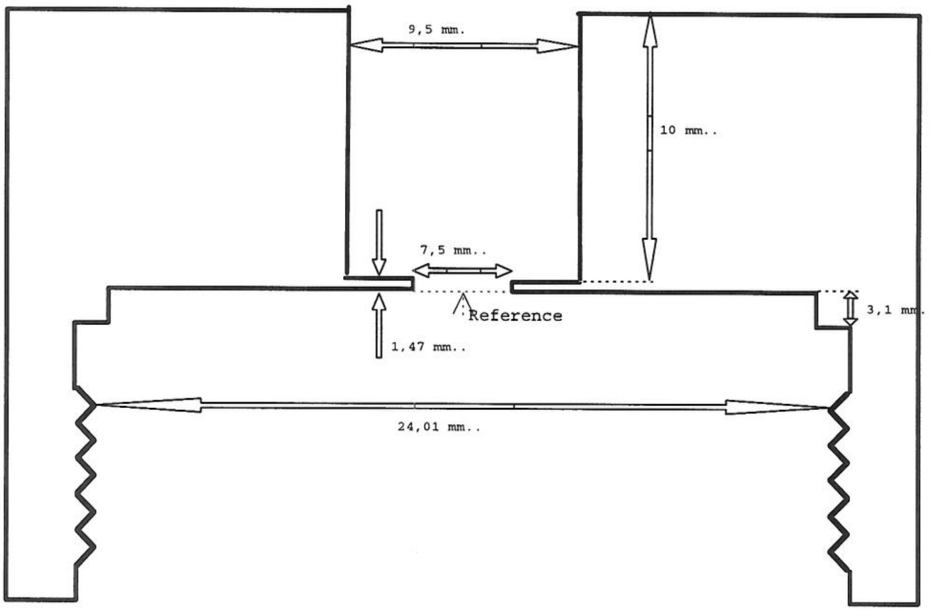


Figure B.1. The special designed coupler to hold the earmould and the Tesla reference volume.

B.2 DETERMINATION OF THE EARMOULD IMPEDANCE

The acoustic impedance of a cavity, Z_{ca}, with rigid walls at low frequencies is given by:

$$Z_{ca} = \frac{\rho c^2}{j\omega V} \tag{B.1}$$

where;
V= volume of the cavity

If one wall or a piece of a wall is exchanged with some non-rigid material the impedance will change. In a lumped element network the impedance of the rigid walled cavity is parallel to the material impedance when the driving source is a volume velocity generator. This means that if the material is rigid the pressure inside the cavity will equal the pressure in a rigid cavity and if the material is very soft the pressure will in the theory be equal to zero.

The result from this experiment was that a standard earmould made in foam or acrylic material works like an acoustical rigid wall in the occluded ear canal. If the earmould is made of a very porous material or the mould is not fitted tightly, it has a highpass-filter effect. The pressure inside the occluded ear canal is less than in a rigid cavity.

C. <u>APPENDIX:</u> EXPERIMENTAL SET UP II

A COMPARISON OF FOAM PLUGS AND ACRYLIC MOULDS AND THE INFLUENCE OF THE INSERTION DEPTH

Earmoulds and earplugs are commonly used in hearing aids and as hearing protection. One purpose of this experiment was to compare the occlusion effect with acrylic and foam plugs fitted to the same persons. The other purpose was to study the effect of different insertion depths.

C.1 LIST OF EQUIPMENT

The set up is illustrated and described in chapter 8.1.1. Data measured during the preparation of the present experiment are written in small fonts.

Probe microphones	2 Knowles electret microphones Type 3024 Rastronics, PVC tubes, outer diameter 1.5 mm
Amplifiers	2 Electrect microphone amplifiers, home build
Attenuators	2 Hewlett Packard 350 D Precision ±0.1 dB up to 40 dB at least
Signal Analyzer	Brüel&Kjær Signal Analyzer Unit Type 2035
DAT-recorder	Sony TCD- D10 Pro used with the record level control in a locked position
Loudspeaker	Vifa 5” unit in a 2.5 l box
Power amplifier	50 W
Reference microphone	½” Brüel&Kjær type 4134
Pistonphone	Brüel&Kjær type 4220
Ear Simulator (IEC 711)	Brüel&Kjær type 4157 used with adaptor DB 2012
Measuring Amplifier	Brüel&Kjær type 2607
Oscilloscope	

C.1.1 Probe microphones and tubes

The probe microphones consist of a PVC probe tube attached to a hearing aid microphone. The microphone membrane has access to the field via a cylindrical inlet. The probe tube (from Rexton Danplex) with outer diameter 1.5 mm and length 85 mm was - with some handy experience - fitted tightly over the tip. These tubes are normally used with standard clinical hearing testing instruments. In order to avoid metallic contact to the skin, the microphones were encapsulated in plastic. These microphones (Knowles EK 3024) are developed to be used in hearing aids and is therefore very small and light (weight = 0.13 g nominal).

C.1.1.1 Probe microphone correction

The nominal sensitivity of the electret microphones is 25mV/Pa but the two probe microphones did not have exactly the same sensitivity.

The microphones were compared by measuring the response in the anechoic room. The microphones were placed as closed as possible without touching each other. The microphone amplifiers for the probe microphones had previously been tested with a frequency response measured with the same microphone in both channels. The amplifiers were set to the same in left and right channel. A loudspeaker was placed 186 cm away, 105.5 cm above the ground grid. A multisine signal of 57 dB rms was used and recorded with the signal analyzer unit (Brüel&Kjaer 2035) averaging in 1 min. Since the two microphones had been measured within 3 months before in a coupler by Bremmelgaard [1997], the measurement was only meant as a control.

The microphone in the right channel (mic. 2) gave between 0.8-1.4 dB less output at 100-4000 Hz than the left channel (mic. 3), the average of the difference was -1.03 dB. The difference between the right and left channel are shown in **Figure C.1**. The phases are the same up to 2 kHz and decrease to about -20° at 5 kHz, see **Figure C.1**. The deviation was in practice applied as a correction factor of +1 dB to the mic. 2 data. . The shape and magnitudes of the curves in **Figure C.1** corresponds with the deviation measured previously in a coupler, [Bremmelgaard, 1997].

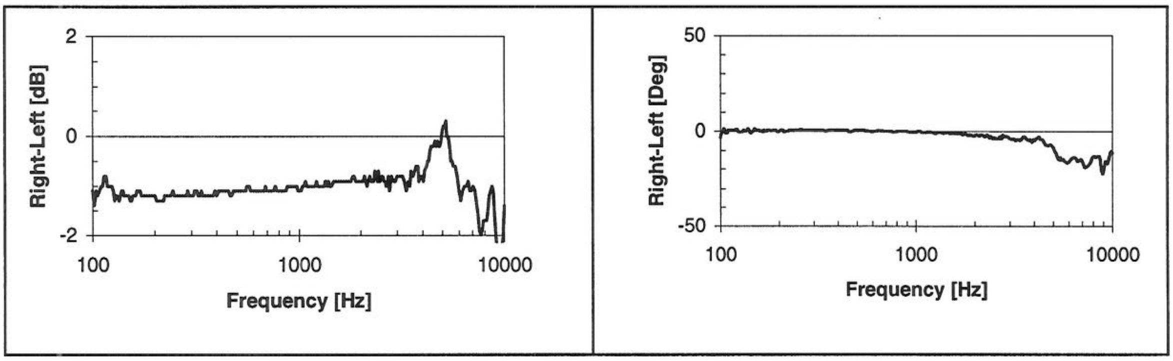


Figure C.1 Sound pressure level difference between the right and the left probe microphone.

C.1.1.2 Probe tube

It was known from other experiments that the tubes from the same sample do not always have the same frequency response. Bremmelgaard [1997] detected a difference of up to 4 dB between two tubes, thus the difference might be significant. For the purpose of the present experiment 10 tubes were required for each subject, one for the open ear and one to put through each of the 3 acrylic earmoulds and one through the foam plug. Therefore, the relative frequency characteristics for all probe tubes were tested in a coupler with the set-up illustrated in **Figure C.2**. The coupler was a small cavity (height 2 mm) screwed on a 1" pressure microphone (Brüel&Kjaer Type 4142) which functioned as sound source. The coupler cavity was covered with a Plexiglas plate in which a tiny hole was made where the tube could be inserted. The opening into the coupler had a diameter equal to the inner diameter of the tube. The tube was put on the probe microphone and the signal send to the Signal Analyzer (Brüel&Kjaer 2012) through the amplifier. The output from the signal analyzer was send via a 20 dB preamp and a calibrator apparatus to the 1" microphone.

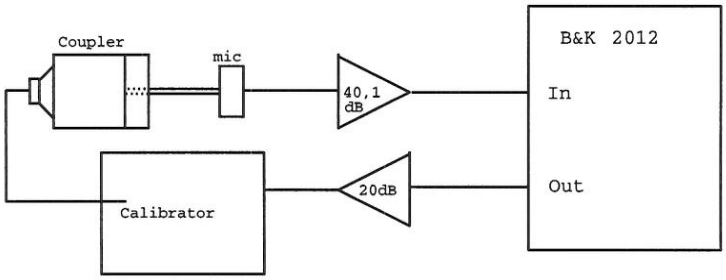


Figure C.2. Set up used to measure the probe microphone characteristics.

The present sample was a good one. Out of 52 tubes, only 1 tube had to be disqualified. The remaining tubes fell into 3 groups as shown in **Figure C.C.3**. A tube belongs to a group if the tube characteristic is exactly the same up to 3 kHz and less than 0.5 dB above that. There is only 1 dB difference in the 3 groups and it is not strictly necessary to sort the tubes, but now that it was done anyway, then a subject would only be measured with probe tubes from one of the groups. The repeatability of taking off and on the same probe was measured 4 times. The results were the same up to 4,5 kHz and deviated maximum 0,8 dB at 8,5 kHz.

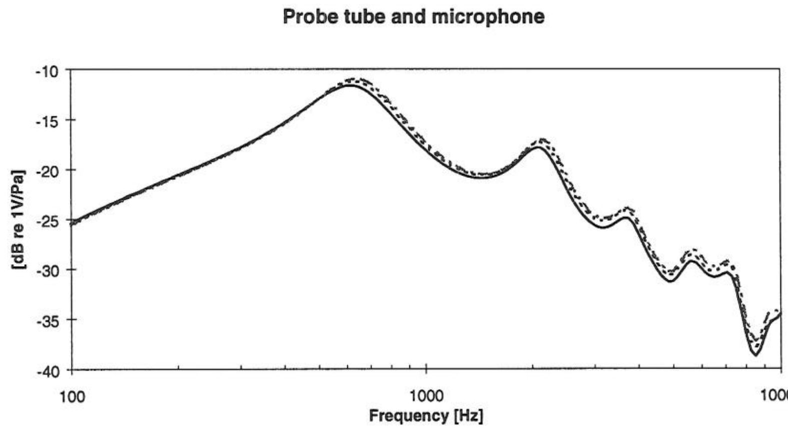


Figure C.C.3. Probe tube characteristics. One example from each of the 3 groups. The differences are negligible.

C.1.1.3 Earmould arrangement

The probe tubes were put through a drilled hole in the acrylic moulds. The foam plugs had already a plastic tube inserted and the probe tube fitted nearly perfectly into that tube. A third tube was put over the probe tube and the plastic tube. Hereby, the leakage between the probe tube and the plastic tube should be avoided. In order to minimize the influence of reflections from the inner tip of the earmould, the tip of the probe tube was extended 3 mm beyond the earmould tip.

C.1.1.4 Mould mechanic properties

The impedance of the mass is given by:

$$Z_{Mm} = j\omega M \tag{C.1}$$

where;
$\omega = 2\pi f$
M = mass of earmould (standard foam earplug = 0.4 g, average of 14 acrylic Full concha plugs = 3.6 g)

The complex stiffness is:

$$s_m = s(1 + j\eta); \quad s = \frac{EA}{l} \tag{C.2}$$

- and the impedance of the stiffness is:

$$Z_{Csm} = \frac{1}{(j\omega / s_m)} \tag{C.3}$$

where;
s = stiffness parameter
s_m = complex stiffness
S_{chism} = impedance of the stiffness
A = cross-section area
l = length of material piece
E = dynamic Young's modulus
η = Dynamic material loss factor

The impedance of the earmould is Z_M in parallel with Z_C. In the case with a rigid acrylic mould, $Z_C = \infty$ and the impedance is a pure mass. In the case of a very light or elastic mould, the impedance becomes infinite, like the mould was not there. A very stiff and heavy mould act as a rigid wall. The mechanical properties of the foam plug is provided by Berger [1990] measured from 10-1000 Hz.

C.1.1.5 Verification of tight tube insertion

The risk for leakage is minimized with the used earmould arrangement, but just a very small and non-visible leakage would be enough to make the measurements useless. It was therefore verified, that the probe tube did not cause a leakage.

The attenuation of the earmoulds was measured in the anechoic box at Oticon. The open ear was simulated using an IEC 711 coupler with the adaptor DB2012, which is cone shaped in order to simulate the outer most part of the ear canal. The coupler was placed 30 cm from the center of a spherical loudspeaker and by means of two laser beams it was possible to put the coupler in the same position every time it was moved. The loudspeaker was fed from an audio analyzer and the coupler signal was connected to the preamp input on the audio analyzer. The input signal to the analyzer from the coupler was calibrated with a Pistonphone. The signal analyzer was used in Steady State mode performing a Log ISO R80 series with complex linear averring in the frequency range 100-2000 Hz. It is adequate to measure the response up to only 2 kHz because if there is a leakage, it occurs at low frequencies.

Each of the medium sized earmoulds was inserted into the coupler in the adaptor DB2012. The real ear moulds have an arbitrary shape and does not fit perfectly into the cone-formed adaptor. It was essential that the moulds were acoustical tightly fitted into the adaptor because otherwise it would be impossible to distinguish any other leakage. This problem was fixed by using some earmould material between the mould and the adaptor while it still was wet. A probe tube was put through the hole in the earmould and the outer end of the tube was attached to a dummy probe microphone such that the system simulated the real ear situation.

The results are shown in **Figure C.4**. The pressure in the coupler is at least 40 dB lower when it is closed with an acrylic mould. The foam plug attenuates less than the acrylic moulds but still considerable. Conclusive, the probe tube does not cause leakage neither in the foam plug nor in the acrylic moulds.

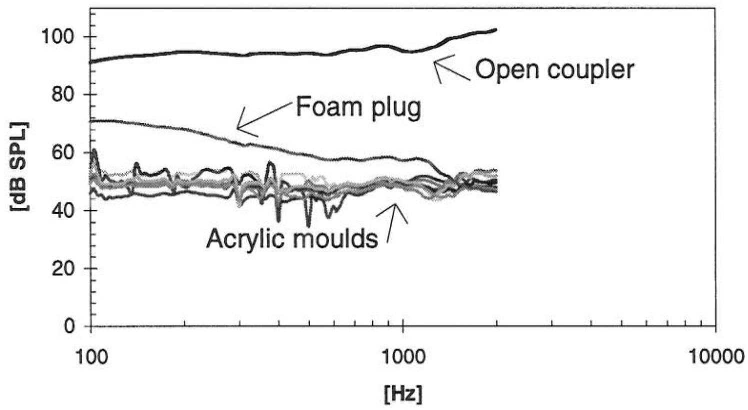


Figure C.4. Sound pressure in the coupler with open coupler, foam plugs and 8 medium sized acrylic moulds.

C.1.2 Loudspeaker

A Vifa 5" Woofer in a box made of 2 cm plywood plates with the inner volume length·height·depth=16·18·10 cm^3 = 2880 cm^3. The loudspeaker response was measured in an anechoic chamber with a ½" B&K microphone Type 4134 and a Signal analyzer unit B&K 2035. The loudspeaker was placed with the center of the membrane 105.5 cm above the floor grid, approximately the same height as the ear of a sitting person. The reference position for the microphone was 181 cm away from the center of the loudspeaker in the centerline and in same height. The sound pressure was 67 dB SPL rms in the reference point. The frequency response was then measured with a microphone in the reference position as well as 15 cm to the left and right, 15 cm and 15 cm upwards. The greatest deviation occurred by moving the microphone backwards, which was excepted. In the other positions the response deviate less than 1 dB from the reference at frequencies below 6 kHz, see **Figure C.C.5**. The error introduced by measuring the real ear response in both ears at the same time, when the loudspeaker is placed in front of the subject, is therefore about 2 dB.

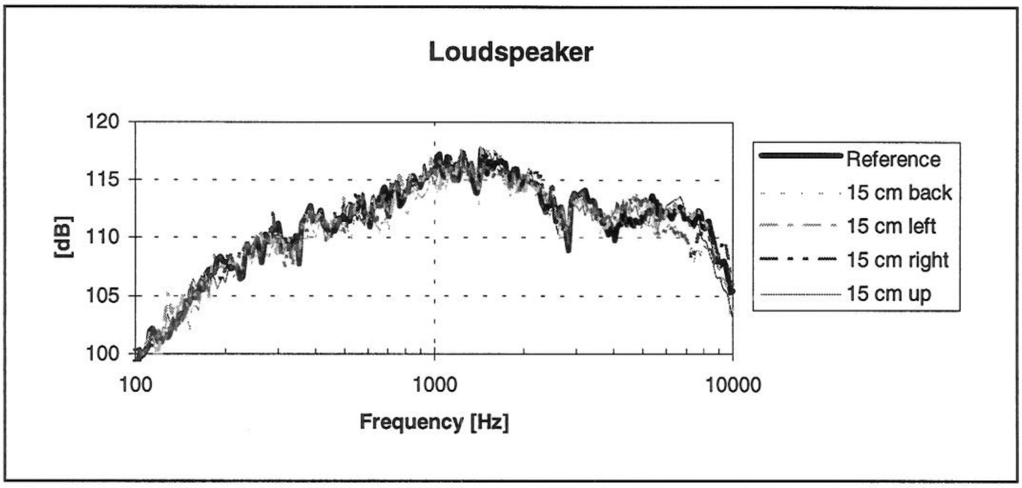


Figure C.C.5. Loudspeaker frequency characteristic in different positions. Arbitrary dB levels.

C.1.3 Calibration

Figure C.6 shows the set up to calibrate the loudspeaker. All the measurements are relative, it was not necessary to make an absolute calibration. The only control in dB SPL was the level of the loudspeaker. In relation to each measurement session, the level of the loudspeaker at the subjects position was checked.

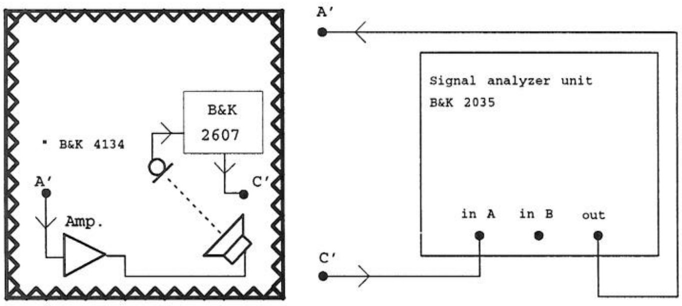


Figure C.6 Set up to check the sound pressure level generated by the loudspeaker.

C.2 MEASUREMENT PROCEDURE

The measurement set up and method was tested on one subject. A try-out was especially important, because the experiment involved subjects and the measurements are hard to do all over again if a mistake is found later on. Relevant adjustment of the attenuators was obtained during this session and a testing of the data analysis was performed.

C.2.1 Subject characteristic

It was required that the subjects did not have problems with eczema, cerumen or other abnormalities in the ear canal. In fact two subjects had to have a wash out of cerumen before the experiment could take place.

Each subject was carefully inspected with a video otoscope by an audiologist from Oticon. This was done in connection with earmould impression taking. The subjects were inspected again during each measurement session.

Audiometric testing took place at the facilities at the Department of Acoustic Technology, unless the subject had their audiogram taken within the last 6 months by an audiologist. Subject 3-5 had perfectly normal hearing (0-15 dB HL at all frequencies). Subject 2 had a bit lower threshold at 250-500 Hz (10-25 dB HL) but nothing that would have been caused by a dysfunction in the middle ear. As a child, subject 2 had an inflammation of the middle ear, and the tympanogram deviated a bit from the normal, but it was judged to still be acceptable. The other subjects had normal tympanograms.

C.2.2 Earmould impressions

Acrylic earmoulds are fitted individually and an impression of each ear were made. The impression, were made a bit longer than normal because three earmoulds should be made: a very short one, a medium standard length and a longer one for each ear. These earmoulds were to be used in the study of the insertion depth and only the medium one was compared with foam plugs. The earmoulds were cast at the Oticon earmould laboratory.

The weight of each earmould is written in **Table C.1**.

	Right			Left		
	Short [g]	Medium [g]	Long [g]	Short [g]	Medium [g]	Long [g]
Subject 2	5.2	5.3	5.5	5.5	5.7	5.4
Subject 3	4.2	4.0	4.2	4.0	3.8	3.9
Subject 4	3.3	3.4	3.5	3.2	3.2	3.4
Subject 5	4.5	4.7	5.2	4.5	4.6	4.9
Average	4.3	4.4	4.6	4.3	4.3	4.4

Table C.1 Weights of earmoulds.

C.2.3 Session m: Occlusion effect with foam and acrylic plugs

The aim of this experiment was to determine the difference in the occlusion effect between the ear occluded with a hard acrylic mould and the ear occluded with a soft foam plug.

The first thing to do was to check the system calibration and make the earmoulds ready for use. Then the subject was placed by the supervisor to sit in the right position to the loudspeaker used for insertion loss measurements. Insertion loss with the acrylic moulds were made with one ear open and one ear occluded at the same time. Occluded ear spectra with the acrylic mould in one ear and the foam plug in the other ear were also measured. The right ear level was determined with microphone 2 and the left ear with microphone 3 at the same time.

Occlusion effect was measured as the sound pressure difference between the occluded ear and the open ear. Only the medium acrylic mould was used for that.

The levels in the right and left ear were obtained simultaneously when one ear was occluded with an acrylic mould and the other with a foam plug. It was necessary to try to assure that the acrylic and the foam plug was inserted in the same depth into the ear canal. The insertion depth of the acrylic mould was estimated by defining the entrance of the ear canal as the point with the largest increase of diameter. The insertion depth was then measured and the foam plug was marked, so when inserting the foam plug, the mark should be exactly at the entrance to the ear canal. With some experience it was found that the most precise way to do this was to define the sharp bend where the ear canal opens out in concha.

The measurements were obtained with three stimuli: Continuos speech during 60 s, nasal sound during 1 s and chewing gum during 10 s. The subjects were instructed to chew until the sugar coating were gone, because the sugar is crunchy and causes a scratching sound. After that, the subject had to divide the gum into 2 equal sized pieces (as well as possible), and chew with a piece in each side of the mouth until the recording was finished.

Open ear level was determined by putting the probe tube carefully into the ear canal. The insertion length was measured to correspond the insertion length of the tube inserted in the earmoulds, so that the tip of the probes was placed in the same point. When the probe was inserted into the open ear, the position of the tip was inserted with an otoscope (without touching the tube). It was important to check that the tip of the tube pointed towards the eardrum free of the ear canal wall. If the tube touches the ear canal wall, there is a risk of creating scratching sounds. In practice the probe was 'glued' to the ear with a very small piece of impression material placed on the outside of pinna just below anti-tragus so it did not disturb the sound field in the ear canal. In this way the probe tube tip could better be kept free of the ear canal wall.

C.2.4 Session g: Insertion depth of acrylic plugs

The aim of this data series was to compare three insertion depths with in principle the same earmould.

These measurements were in the principle performed the same way as just described. Thus, here the insertion loss was determined by first measuring the open ear and the occluded ear level. It could not be recorded at the same time, because the subject were sitting with the ear towards the center of the loudspeaker.

C.3 RELIABILITY

C.3.1 Right and left ear open

The sound pressure in the open ear was measured while the subject read aloud during 60 sec. and vocalized an /aaa/ and an /iii/ in 1 sec. A single vowel sound is shown in **Figure C.7**. At 200-2,000 Hz the difference between right and left ear is less than 4 dB except from subject 3 for 250-270 Hz. The phase difference is less than approximately 80° at all frequencies and less than 50° in most cases.

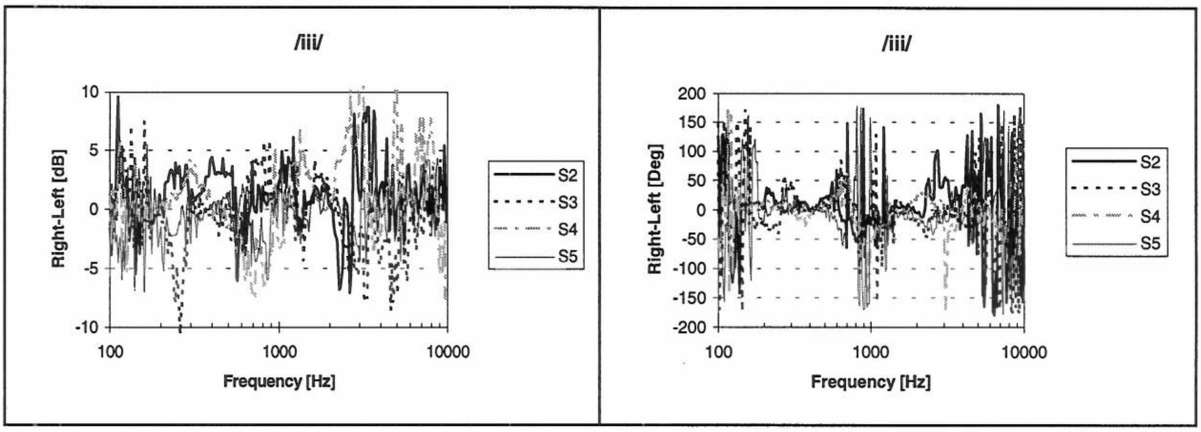


Figure C.7. Ear canal sound pressure differences. Right - left open ear /iii/. 4 subjects.

Continuos speech averaged in 60 sec. Does not fluctuate as much, see **Figure C.8**. The difference between right and left ear lies within ±4 dB at f < 1 kHz and ±5 dB at f<3 kHz. The phase differences are insignificant up to 2 kHz.

The deviation between right and left ear is mainly due to uncertainty in position of the probe tube, because in practice it is not possible to place the probe tube exactly at the same point twice. Another reason could be anatomical differences between left and right ear.

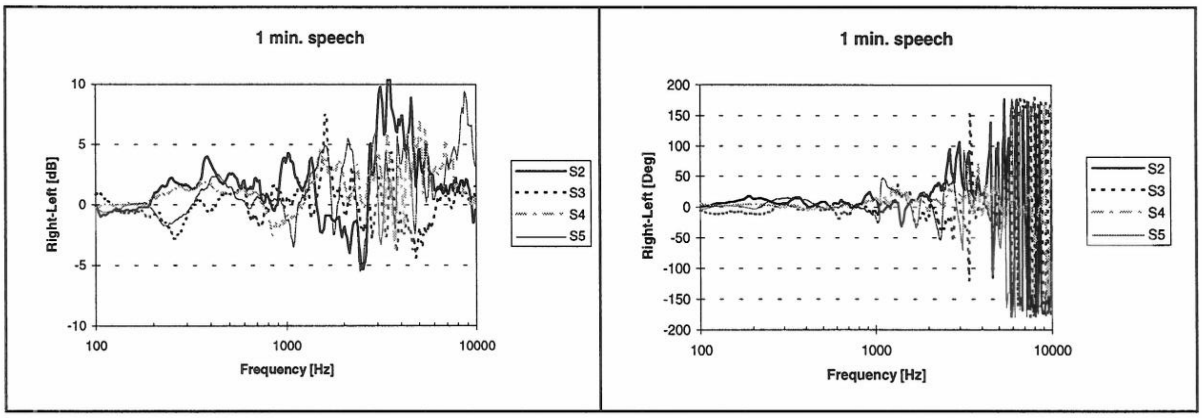


Figure C.8. Ear canal sound pressure difference between the right and the left open ear. Continuos speech. 4 subjects.

Another way to analyze the reliability between right and left ear measurements is to look at the coherence. The coherence, γ, between 2 signals (one from each ear) A and B, is a statistical value given by [Randall, 1987]:

$$0 \leq \gamma_{AB}^2(k) = \frac{\left|G_{AB}(k)\right|^2}{G_{AA}(k)G_{BB}(k)} \leq 1 \qquad \text{(C.4)}$$

where;
k = frequency component number k in the digital spectrum, $0 \leq k \leq N$
G = cross-spectrum = $(A(k) \cdot B(k))/N$

If the signal from channel A is perfect linear with the signal from channel B, then $\gamma^2 = 1$. The smaller γ^2 becomes the poorer is the linear relationship between A and B.

The coherence between right and left ear is better for 1 min. speech than for a sustained vowel. The coherence coefficient for the speech is approximately 1 up to 1 kHz. and more than 0.5 up to 2-3 kHz which is acceptable. The coherence is worst when the energy in the signals are low, i.e. the signal-to-noise ratio is low. For example in the /iii/ sound, there will be most energy at the fundamental and the first formats and at the ear canal resonance.

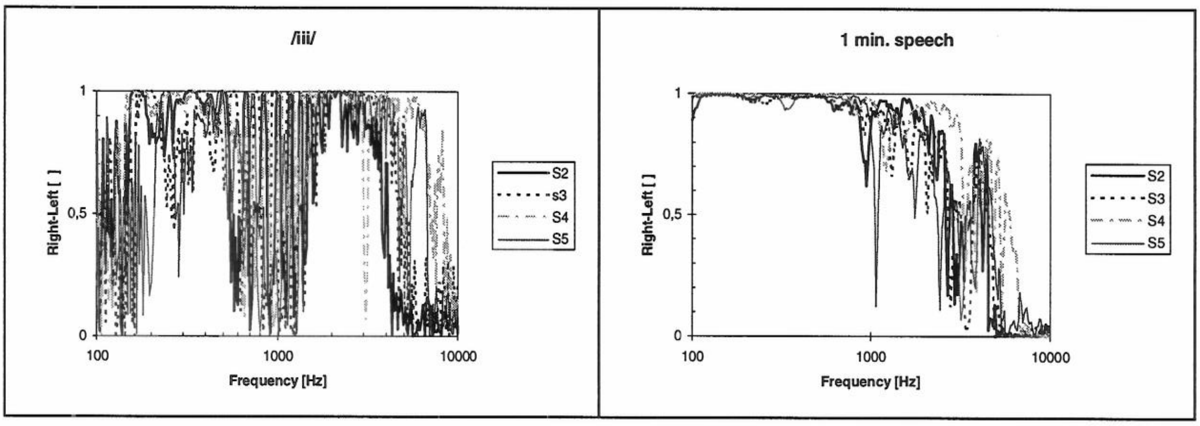


Figure C.9. Coherence between right and left open ear. Left: /iii/. Right: Continuos speech. 4 subjects.

The other purpose for this analysis was to check for systematic differences between right and left ear. One way to check that, is to perform a statistical t-test on the differences. But first of all, it is not reliable to perform statistical analyses on only 4 subjects and secondly, the curves in **Figure C.7** and **Figure C.8** do not indicate systematic errors as the curves lies randomly around 0 dB.

C.3.2 Right and left ear occluded

It must be expected that the difference between the sound pressure in the right and left ear is larger when the ears are occluded than when the ears are open, because the sound pressure in an occluded ear depends on the shape of the earmould. The sound pressures with an acrylic mould in both ears are plotted in **Figure C.10**. The results are very good for subject 2 and 5 where the maximum deviation between right and left ear is 2.5 dB for 200-1000 Hz when the stimulus was the subject's own voice (right panel). Subject 3 and 4 shows a very distinct difference. For subject 3 it can be explained by the fact that the right mould fitted more tightly than the left mould. This is clear from the occluded real ear measurements, left **Figure C.10**. Also, the subject himself commented (or rather complained) that the left mould did not fit very well.

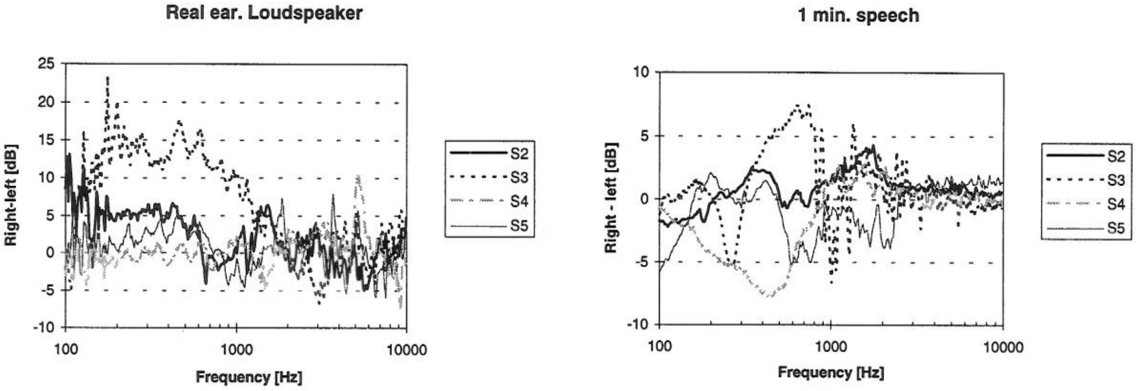


Figure C.10. Ear canal sound pressure differences. Right - left occluded ear. Acrylic mould. Left: stimulus, loudspeaker. Right: stimulus, continuos speech. 4 subjects.

C.3.3 Repeatability of the occlusion effect

C.3.3.1 Occluded ear, acrylic moulds

Occlusion effect caused by the acrylic medium earmoulds was measured twice on each ear. The measurement was repeated at the two test session over 2 days. The occlusion effect was measured as the difference of the sound pressure in the open right ear and the occluded left ear or the opposite. Only continuos speech was used as sound source. The repeatability is good below 400 Hz, where the measured occlusion in most cases deviates within 3 dB (except subject 2). From 400-1,000 Hz the deviation is within 5 dB and up to 10 kHz the deviation is mostly within 10 dB.

The right ear data for subject 2 from session g and session m agree very well. The left ear data do not agree and they deviate more than the deviations found for all other subjects, so it indicates an error in the recording, and the data on occlusion effect on the left ear in session 1 (session g) will be left out of the data analysis.

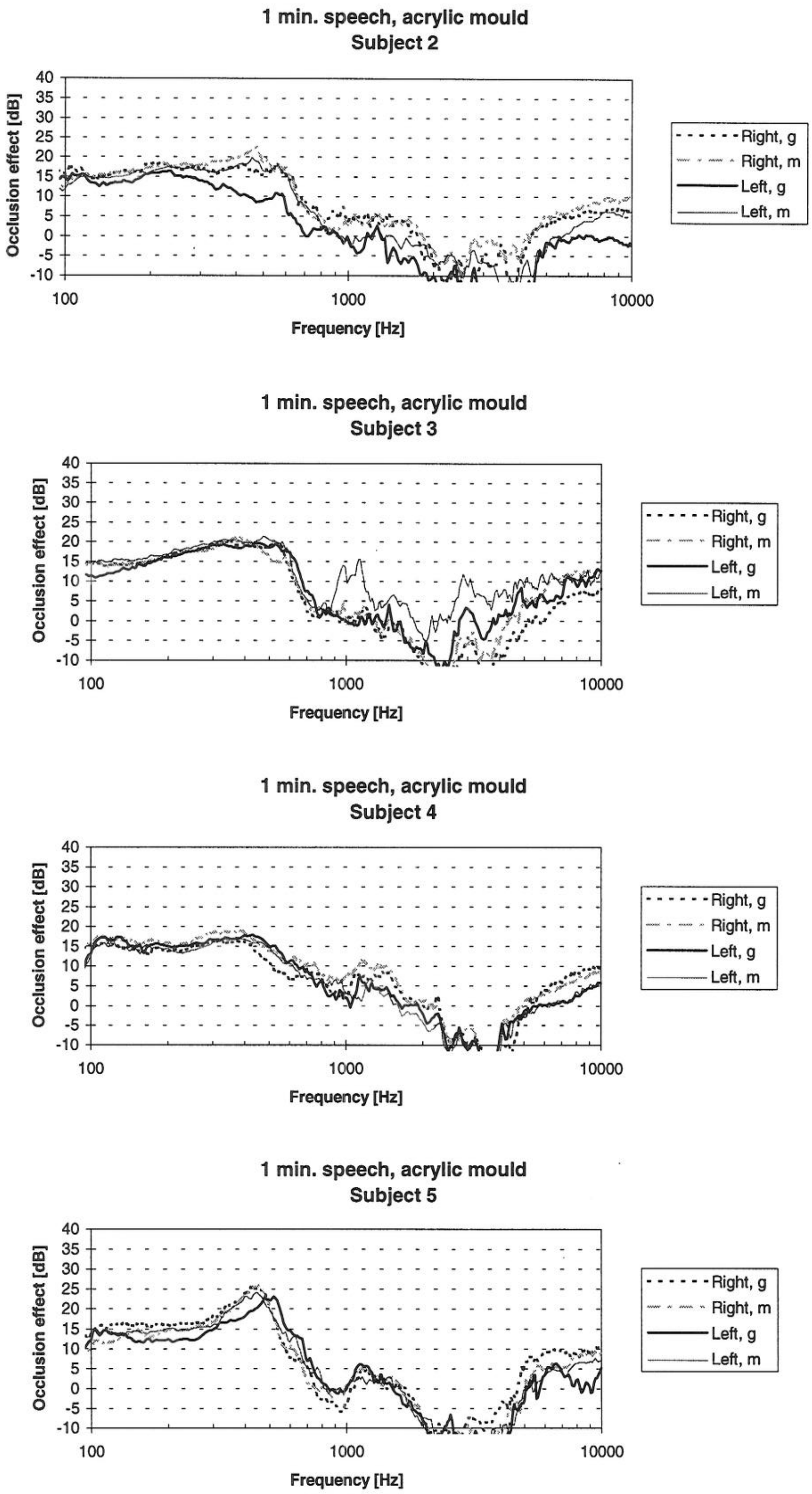
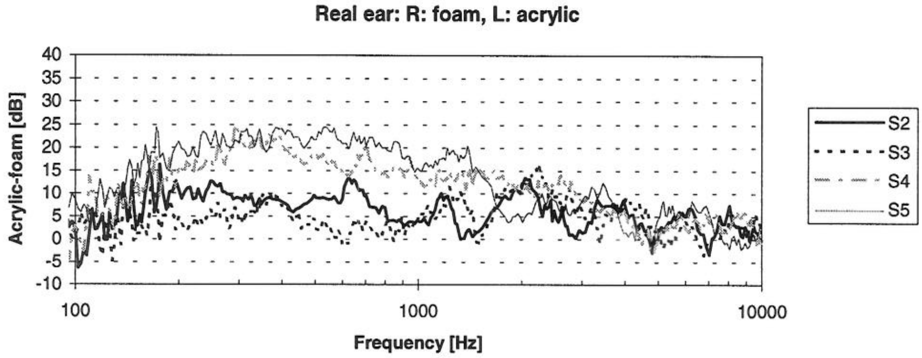


Figure C.11. Occlusion effect repeated twice in two days. Continuos speech. Label right = right ear occluded, left ear open. Left = left ear occluded, right ear open. Suffix: g = 1'st test session, m = 2'nd test session.

C.4 SESSION M: FOAM PLUGS CONTRA ACRYLIC MOULDS

Only the primary data will be presented here, see main report chapter 8 for further interpretation of the data.



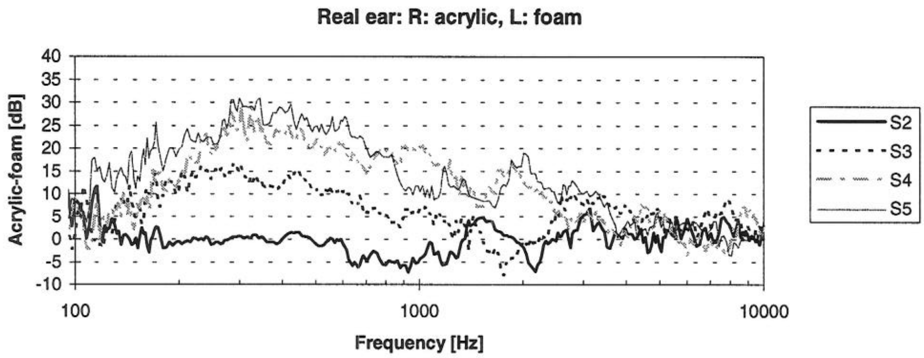
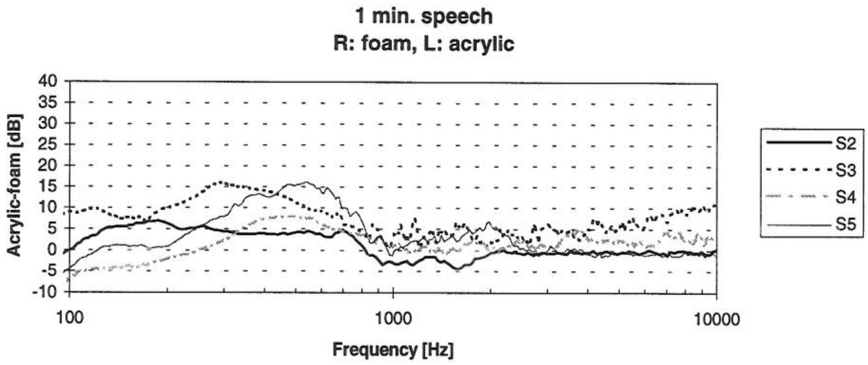
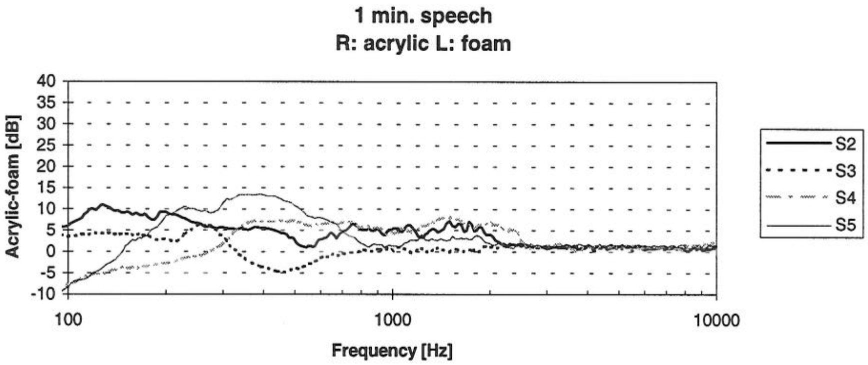
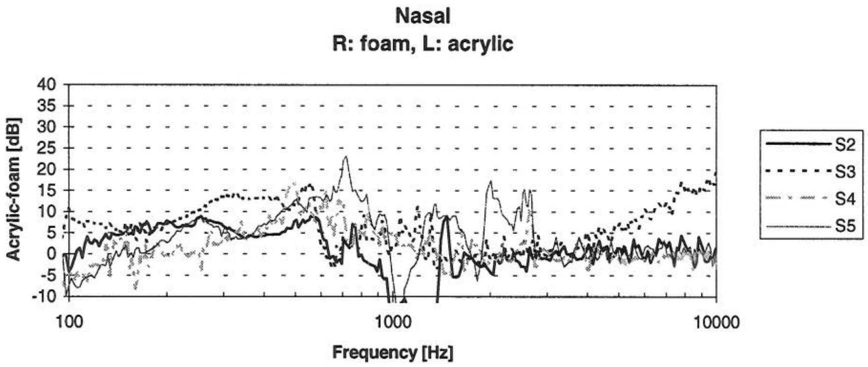
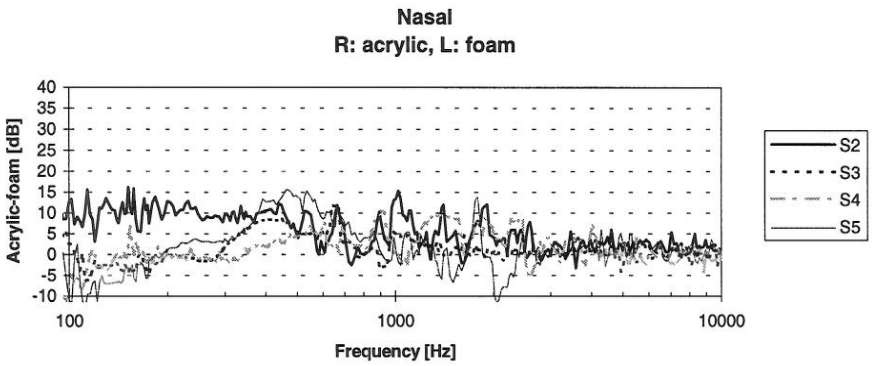


Figure C.12. Difference of the sound pressure between the ear occluded with an acrylic mould and the ear occluded with a foam plug. Stimulus: loudspeaker.









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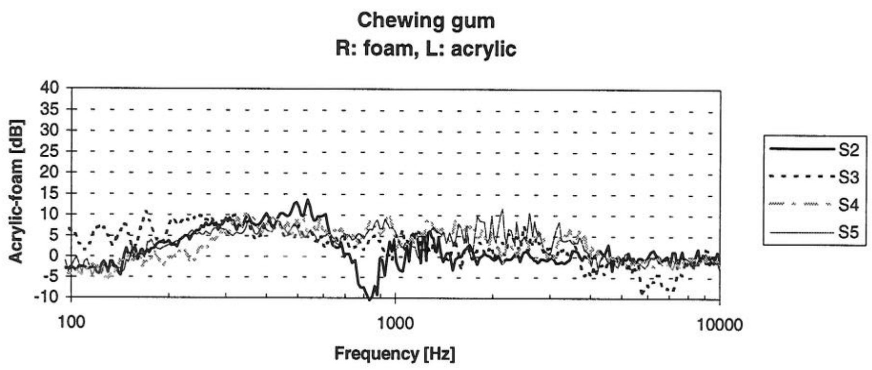


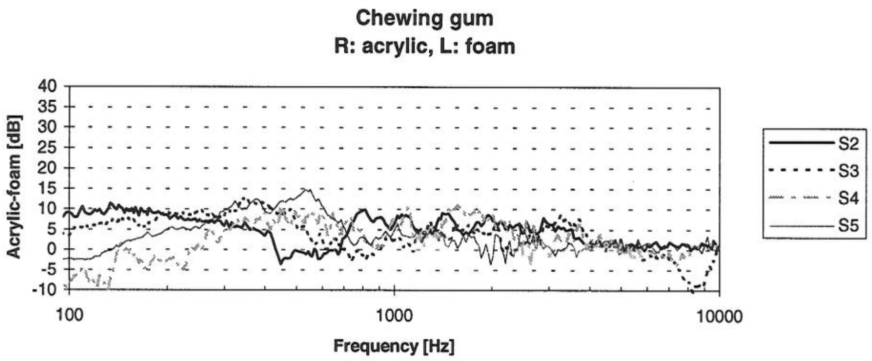
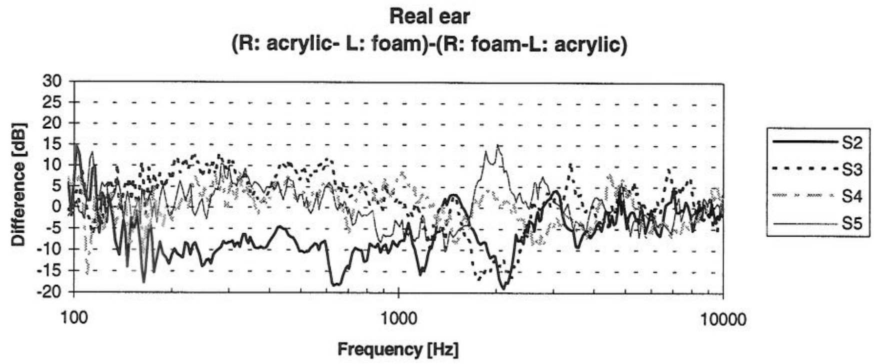


Figure C.13. Level difference between the ear occluded with an acrylic mould and the ear occluded with a foam plug. Stimuli: subject's own voice: speech, nasal sound and chewing gum.

The difference between the two measurements on each subject is shown in **Figure C.14** for external sound from a loudspeaker and 60 seconds own speech. Subject 2 and subject 5 are good between 200-800 Hz. Subject 3 is distinct different. The difference for subject 3 is very clear from **Figure C.13** (1 min speech, R: acrylic, L: foam). The differences are not similar to the differences in the occluded real ear measurements, which lies within ±10 dB.



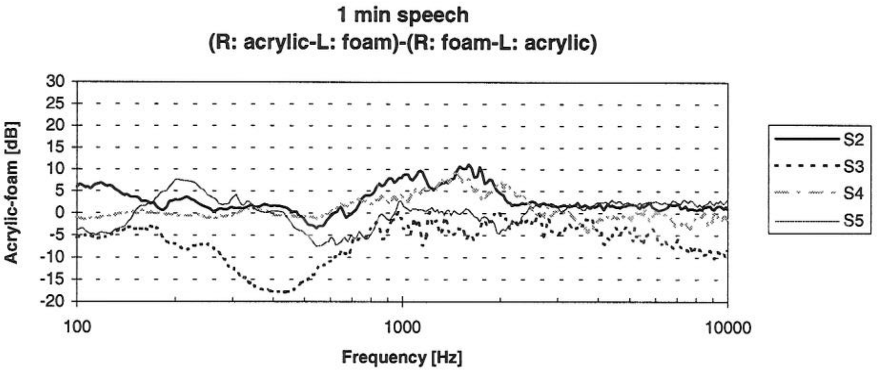


Figure C.14. Difference between the difference between acrylic mould and foam plug in right and left ear and contrary. Top: stimulus, loudspeaker. Bottom: stimulus, 60 sec.. speech.

C.5 SESSION G: ACRYLIC MOULDS OF DIFFERENT LENGTHS

Three acrylic moulds were made for each ear to each of the 4 subjects. The moulds had the same design but the length of the earmould tips were varied. The lengths were measured in a straight line from a defined point on the outer side of the mould to the inner tip of the mould. The lengths tabled in **Table C.2** can only be used as a relative measure to the same ear.

	Lengths [mm]	Subject 2	Subject 3	Subject 4	Subject 5
Right	Short	14.0	16.0	15.0	15.0
	Medium	15.5	19.0	16.5	18.0
	Long	19.5	18.0	17.5	20.0
Left	Short	14.5	15.4	16.0	17.0
	Medium	17.5	16.0	18.5	17.0
	Long	19.5	19.0	20.0	19.5

Table C.2 Length of earmoulds, measured on a straight axis.

The occlusion effect was measured for each mould by detecting the sound pressure in the occluded ear and the open ear simultaneously. Each mould was measured with continuous speech and the sustained vowels /eee/ and /aaa/. Only the occlusion effects with speech are shown in **Figure C.15**. The measured data for the vowels do not add any extra information.

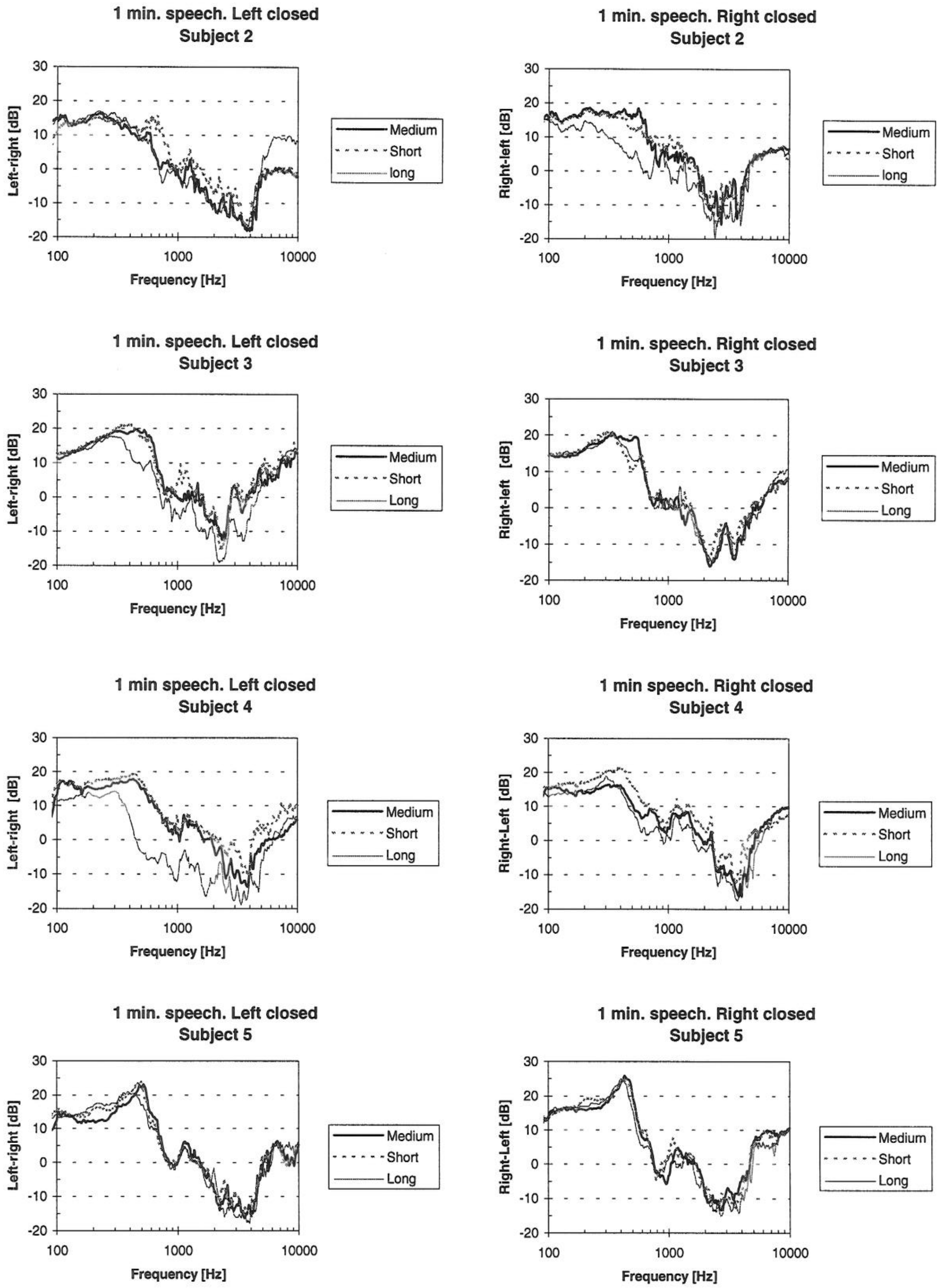


Figure C.15 Occlusion effect with varying length of acrylic moulds. subject's own voice, continuous speech.

D. APPENDIX: EXPERIMENTAL SET UP III

A HEAVY AND A LIGHT EARMOULD

D.1 EXPERIMENTAL SET UP

The set up and the method is the same as described in appendix C, but the experiment did no take place in an anechoic room. The experiment were only done with occluded ears, and the primary sound source was body conducted sound, thus the need of an anechoic room was not evident.

The occluded ear levels were recorded with the Brüel&Kjær Type 2035 Analyzer and the right and left channel was recorded simultaneously. The recordings were done from 50-2000 Hz with maximum overlapping Hanning-windows.

One subject participated and costume-made earmoulds were used. The earmoulds were designed as an ITEC mould, where the faceplate of the mould reached a few mm out from the ear canal entrance. The moulds were made hollow. One mould was filled with foam and the other mould was filled with 3 steel balls and some impression material. The weight of the moulds were:

	Heavy Steel ball fill	Light Foam fill
Right	5.38 g	1.08 g
Left	5.10 g	1.09 g

The probe tubes were inserted through a drilled hole in the moulds. The sound pressure in the occluded ear was measured while the subject either read aloud in 30 s, sustained a nasal sound in 5 s or chewed gum in 30 s.

D.2 ATTENUATION OF THE EARMOULDS

The attenuation of the earmoulds were measured with exactly the same technique as described in chapter C.1.1.5. The difference between the sound pressure in the IEC 711 coupler closed with the earmould and the sound pressure in an open coupler is shown in **Figure D.1**. The results shows that the moulds are tight. If there was a considerable leakage, then the difference would about 0 dB at low frequencies (at least below 200 Hz).

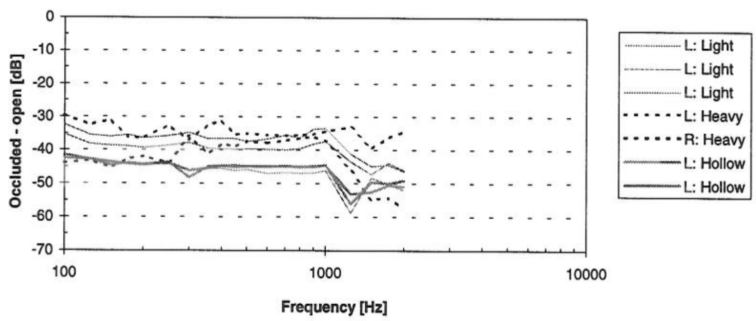
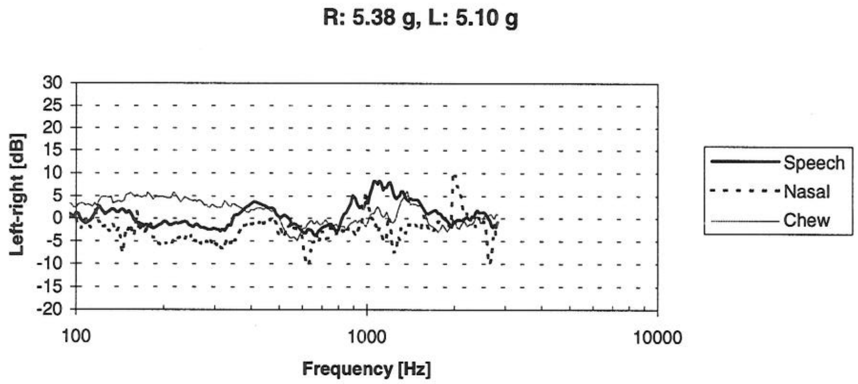


Figure D.1 Sound pressure difference between an occluded and an open coupler. The earmould shells were made of hard acrylic and filled with foam material (light), steel balls (heavy) or kept empty (hollow). L: left ear, R: right ear.

D.2.1 Right and left ear

The sound level in the right and left ear with light moulds and heavy moulds inserted are shown in **Figure D.2**. The sound pressure between the right and left ear for running speech deviates maximum 5 dB below 700 Hz and in average 2.5 dB. The nasal sound and chew is less reliable, first of all because the differences does not lie randomly around 0 dB but for example for the heavy mould, the chew in the left ear is 5 dB higher than the right ear below 700 Hz.



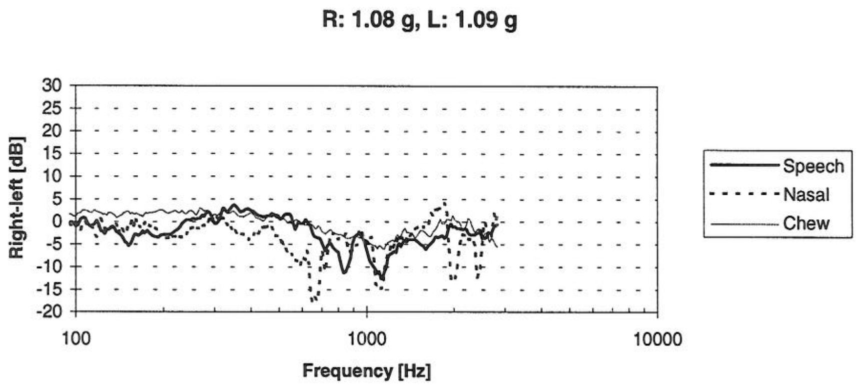


Figure D.2. Difference in occluded ear sound pressure levels between right (R) and left (L) ear on the same subject. Top: a light mould. Bottom: a heavy mould. Subject's own voice.

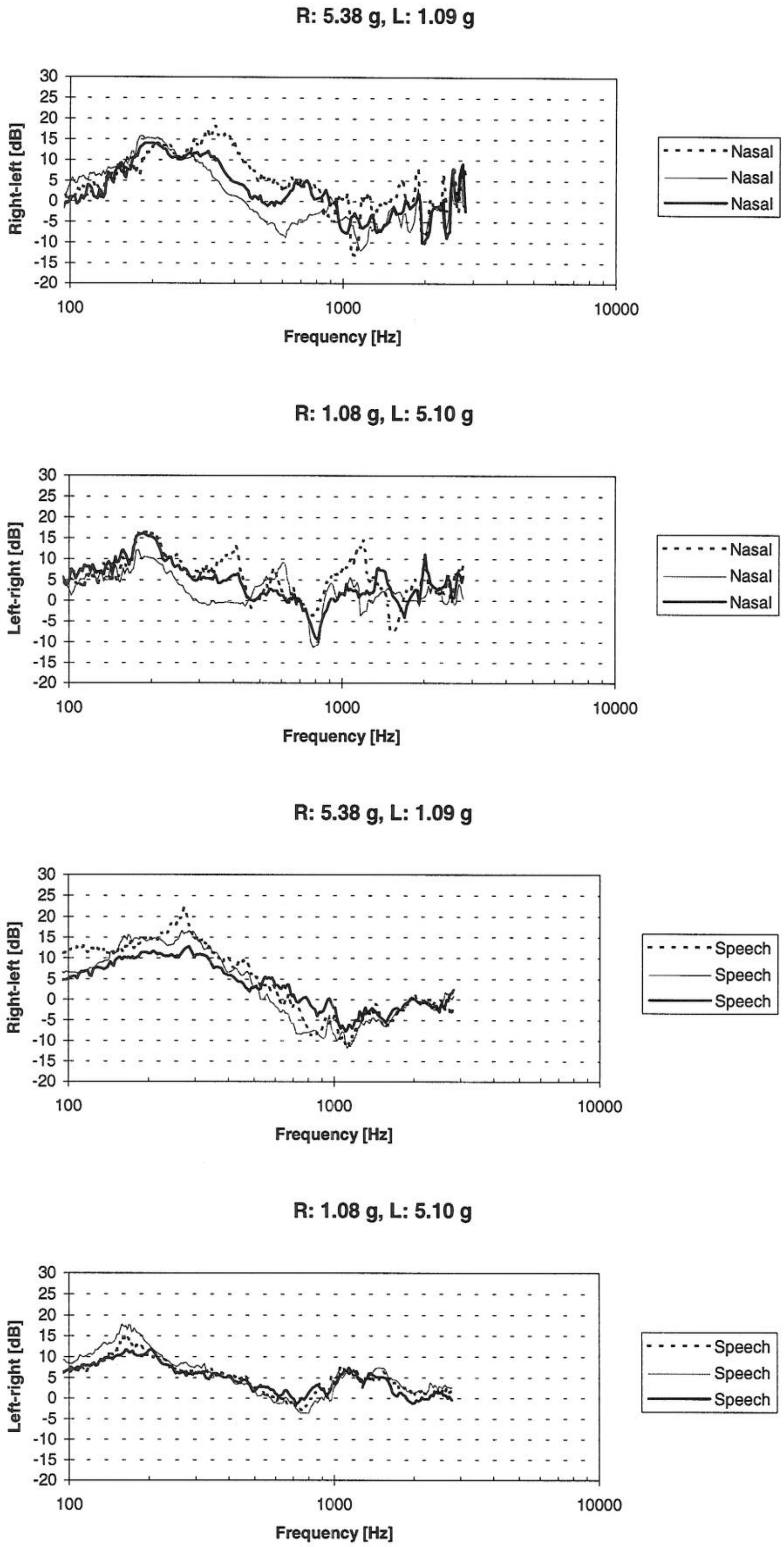


Figure D.3. Difference in occluded ear sound pressure levels between an ear occluded with a light mould and the other ear occluded with a heavy mould. The curves show the measurement performed 3 times on the same subject. Top: Nasal sound. Bottom: Continuos speech. Subject's own voice.

The speech and nasal sound were measured three times, see **Figure D.3**. Between each time, both moulds were completely taken out of the ear and inserted again,. Each measurement has the same signature in all graphs. Speech is the easiest to repeat but still a difference of 10 dB can occur. The nasal sound is also repeated acceptably. The standard deviation below 1 kHz is up to 5 dB for speech, but mostly below 3 dB. Between 300-1000 Hz the standard deviation for the nasal sound is 5-7 dB. The maximum difference in row 3 and 4 in **Figure D.3** does not occur at the same frequency.

The phase between right and left ear when both ears are occluded a light mould is plotted in **Figure D.4**. The phases for the chewing is very unstable and shifts all the time. Speech and nasal sound 'behaves' better and the phases are nearly flat.

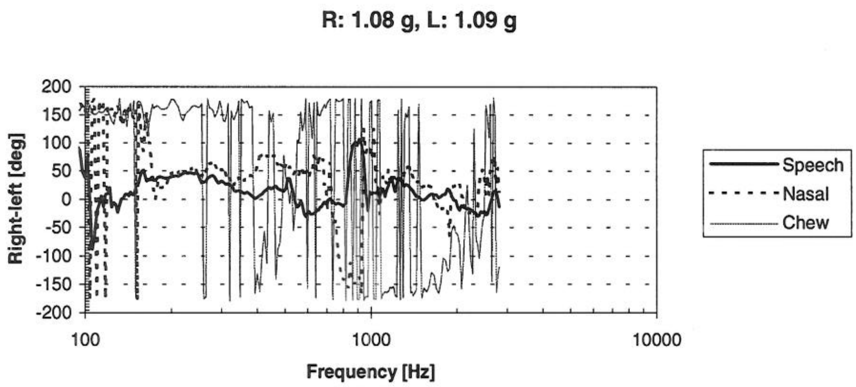


Figure D.4. Difference of the phases between right and left ear occluded with light moulds.

An important point to make, based on these measurements, is that real ear measurements with the subject's own voice are extremely difficult to reproduce and deviations of up to ±5dB are to be expected.

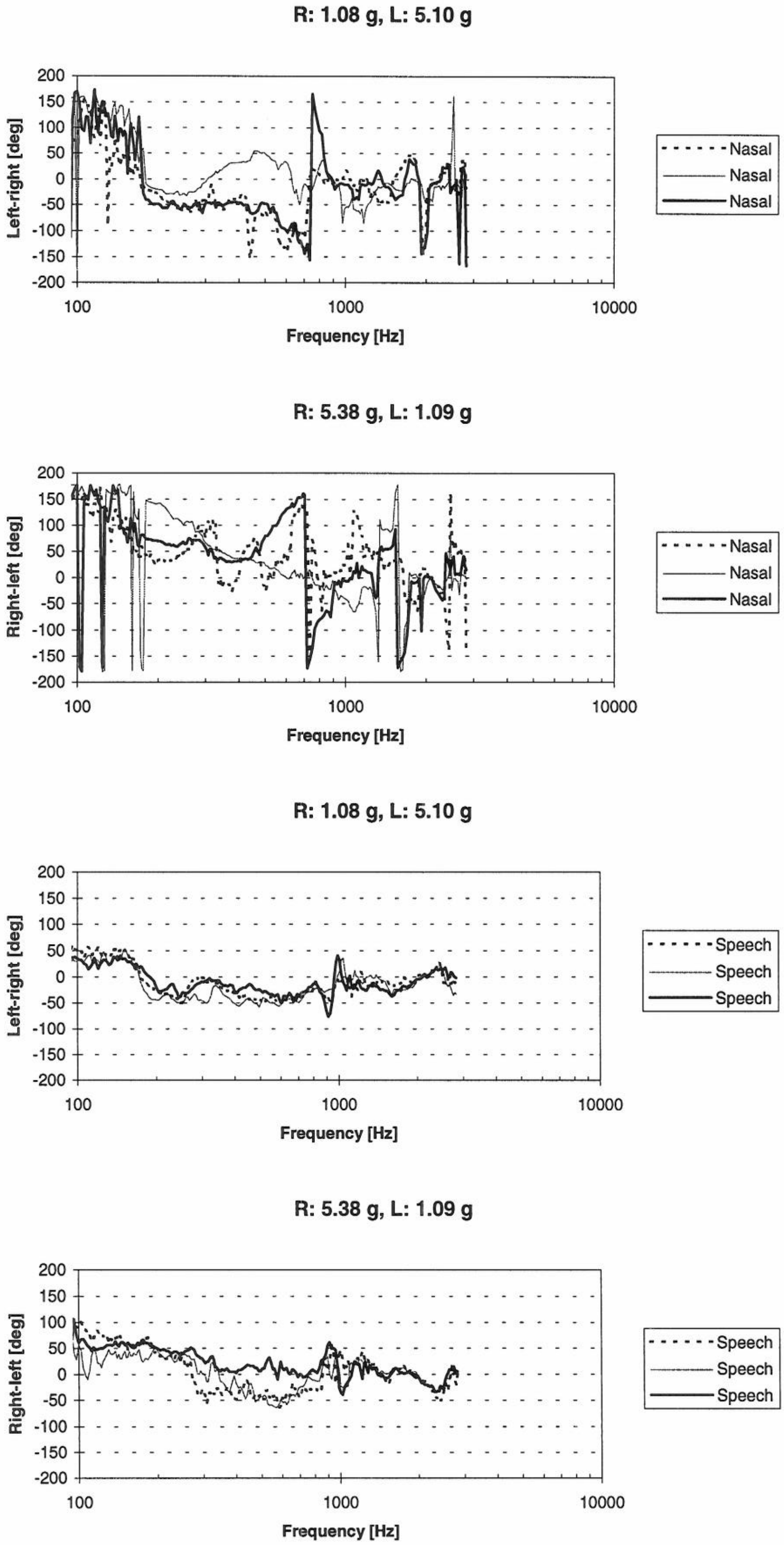


Figure D.5. Difference of the phases between an ear occluded with a light mould and the other ear occluded with a heavy mould. The same measurement performed 3 times on the same subject. Top: Nasal sound. Bottom: Continuos speech. Subject's own voice. R: right, L: left.

E. APPENDIX: EXPERIMENTAL SET UP VI:

Ratio of air borne and body conducted sound

This experiment was performed at the acoustic laboratory at the National Research Council, Ottawa, Canada. The measured data are shown in chapter 10.

E.1 EXPERIMENTAL EQUIPMENT

This is a list of the equipment used in the experiment. Nominal data from the manufacturer's data sheets are written in normal fonts. Data measured during the preparation of the present experiment are written in small fonts.

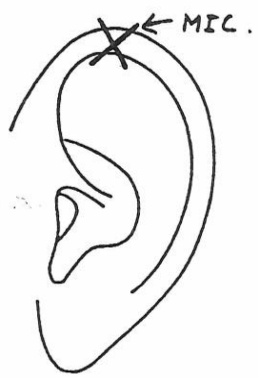
Voltage generator (+/-) 12 V

Lowpass filter and amplifier

Stanford Research Systems, Inc.
Dual Channel low-pass filter, Model SR640
Amplifier in steps of 10 dB.
The lowpass filters cut-off frequency can be set in steps of 1 Hz.
9 kHz lowpass: $f_{-3dB} = 9750$ Hz
$f_{-40dB} = 12000$ Hz

Reference microphone

1/2" Brüel&Kjaer, type 4134
Positioned right outside the top of pinna without touching the skin pointing upwards



Amplifier

Brüel&Kjaer Measuring amplifier, Type 2610
Input gain from 0-50 dB provided a very accurate gain from 50 Hz up to at least 10 kHz.

Probe microphone

Probe microphone originally used with standard clinical equipment
Inserted 8+3 mm in the ear canal

Pre-amplifier

Laboratory construction

DAT-recorder

Panasonic, Model SV-3700
f_s=44.1 kHz, analog input, 2 channels

Spectrum analyzer	Stanford Research Systems, Inc. FFT Spectrum Analyzer, Model SR760
Loudspeaker	JBL, Control Micro, Ser.no. J338-033736 B 4.5 cm off-axis error = ±1dB
Measuring system	Made at NRC. Measuring system including PC, preamplier and controller The system uses a program named 'measure'
Calibrator	Brüel&Kjaer Type 4231 94 dB SPL at 1.000 Hz
Headphones	
Oscilloscope	
Attenuating box	Height=104 cm, width=54 cm and depth=54 cm. The outside and inside of the box were covered with 10.16 cm (4") fiberglass
Anechoic room	Facility at the Acoustics and Signal Processing group, National Research Council, Ottawa, Canada - not anechoic below 200 Hz. For some microphone positions a reflection at 187 Hz was observed.

E.2 BOX DESIGN

E.2.1 Box attenuation

The most realistic measure of the box attenuation was obtained with the subject attached to the box. Since, the subject will be speaking in the experiment, a sound source similar to the subject's own voice would be the best to use. Unfortunately, it was not possible to attach the box to a Head-And-Torso-Simulator. Without a simulator it is difficult to make a sound source that will create the same radiation pattern as the radiation from the mouth and nose during vocalization. In order to get as close as possible to the experimental situation, where the subject will be vocalizing, a small loudspeaker was placed close to the mouth and chin of the subject. The loudspeaker was placed in a 0° and in a 45° angle to the subjects nose and the loudspeaker center was raised to mouth height.

The sound pressure in the subjects ear canal was measured. The box was then placed in the room and the test ear attached to the box. The box influence was calculated as the difference in sound pressure level between the free field condition and the boxed ear condition. The test stimulus was a white noise signal of 74 dB SPL rms measured at the position of the subject.

The average of the attenuation with the loudspeaker in the 45° position and the 0° position will be used. The box attenuation for both angles are plotted in **Figure E.1**. The measurements shows that the box attenuates less at higher frequencies, which is probably not true. The reason to this curve must be found in the set up used to measure the box attenuation, which was far optimal. The measured data seems to be reliable up to 2 kHz and this is good enough for the purpose of this experiment because the ration of body conducted sound relative to the air borne sound.

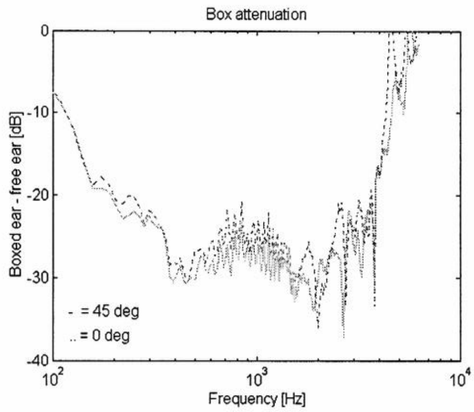


Figure E.1 Box attenuation. Speaker position 45° and 0°.

E.2.2 Disturbance from the box

The disturbance in the anechoic room caused by the box was estimated by comparing the signal from the reference microphone with and without the box installed in the room. A loudspeaker was placed in front of the subject's mouth. The reference microphone (½" Brüel&Kjær Type 4134) was placed in the reference position right outside the tragus of the ear. The difference is shown in **Figure E.2**.

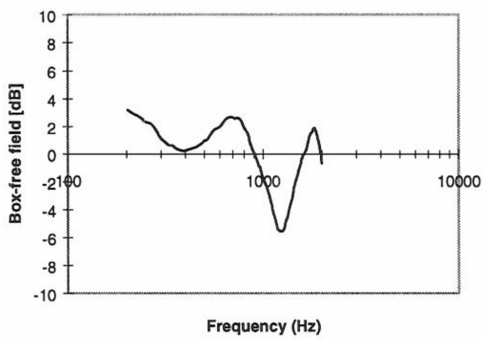


Figure E.2 Box disturbance of the free field. Reference microphone. 1/3 octave smoothed.

Determination of the real ear levels for continuous speech is normalized with the difference in rms-value of the reference microphone signal (200-2000 Hz). It would be most correct to account for the box influence, but the difference in rms-value (200-2000 Hz) due to box disturbance is exactly 0 dB, thus there is no need to account for the box disturbance regarding continuous speech.

The vowels produced in the open ear and the boxed ear were normalized with the levels at every single harmonic frequency in the reference microphone signal. Hence, the box disturbance cannot be disregarded.

E.2.3 Effect of cushion pressure

The subject can be attached less or more tight to the front plate. It is possible for the subject to press hard against the cushion, and the skull might couple mechanically to the front plate of the box, and this mechanical coupling would affect the natural vibrations of the skull. The effect of the cushion pressure was measured with the subject sitting in the anechoic room. The front plate was detached from the box and strapped on the subjects head. The sound pressure in the ear canal produced by the subject's own voice was measured first with the front plate very loose attached and then extremely tight attached. The difference between the two sound pressures is a measure for the effect of cushion pressure. The effect was measured both for the open ear and when the ear was occluded with an earmould.

When the ear was open, the detected difference was ±2.5 dB or less below 2 kHz. This in the range of repeatability error for probe measurements and therefore it can not be stated for sure, that it is an effect of the cushion and not a measurement uncertainty.

E.2.4 Reference microphone

The reference microphone was positioned at the non-boxed ear at the top of pinna and 1 cm out so it did not touch the skin. The effect of moving the reference microphone a few centimeters upwards, downwards and sideways was checked using a Kemar mannequin in an empty anechoic room. A speaker was positioned 3 cm in front of the mouth of the mannequin with the centerline at mouth height.

The response from the reference microphone was measured with a PC-program called 'meas2' using an input voltage of 2,38 V. The system output is a frequency sweep (up to 20 kHz) and the input is the microphone signal, through a Brüel&Kjaer measuring amplifier Type 2610.

The effect of moving the reference microphone a few centimeters is in most cases less than 1 dB as shown in **Table E.1**. Displacement of the microphone 2.5 cm downwards and 1 cm backwards gave the largest measured difference of +1.25 dB at f < 2 kHz.

Microphone position	MAX of $Lp_{,reference\ pos.} - Lp_{,new\ pos.}$	
	f < 2.000 Hz	2.000 Hz < f < 5.000 Hz
2.5 cm out from the ear	+ 0.25 dB	± 5 dB
1.5 cm moved forwards	- 0.75 dB	± 5 dB
- 1.5 cm moved downwards	- 0.75 dB	± 5 dB
2.5 cm moved downwards and 1 cm backwards	+ 1.25 dB	± 6 dB

Table E.1 Effect of displacing the reference microphone relative to the reference-position.

E.2.5 Probe microphone

The frequency response of the probe microphone system was determined by comparing the probe microphone with a ½" Brüel&Kjær microphone in a free field. The reference microphone has a flat frequency response. The microphones were placed 50 cm from a loudspeaker in the anechoic room.

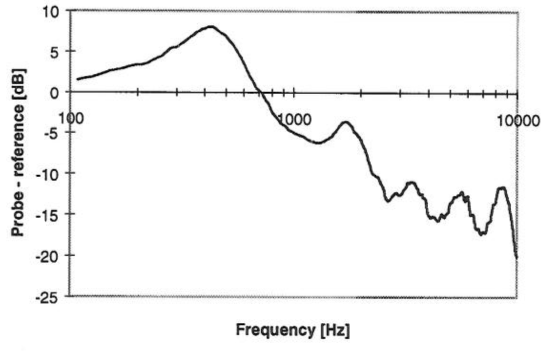


Figure E.3 Probe microphone and tube system characteristic.

Experience from previous measurements with this probe microphone have shown that the probe tube frequency response can vary up to a few dB. A procedure to select probe tubes with the same frequency response was performed. Each tube was measured in the anechoic room.

E.2.6 Calibration

The system was calibrated using a Brüel&Kjær calibrator Type 4231. The calibrator tone (94.0 dB ± 0.2 dB at 1 kHz) was measured with the 1/2" microphone, followed by a 70 Hz highpass filter and a 9.0 kHz lowpass filter and into a Brüel&Kjær measuring amplifier Type 2610, with an amplification of 30 dB.

Digital signals were then calibrated by calculating the rms-value of the calibration signal in the time domain and setting this value equal to 94.0 dB. Hereby, a calibration coefficient was computed.

As the preamp was set to 30 dB the correct level of a certain spectrum, Lp_s can be calibrated by:

$$Lp_s = 10\log\left(\frac{S(m)}{M-1}\gamma^2\right) - (preamp - 30) \qquad \text{(E.1)}$$

where;
M = number of frequency components in the spectrum
S = spectrum
γ = calibration coefficient
preamp = amplifier setting for the recording of S
m = frequency component number m

E.3 DATA TREATMENT

E.3.1 Data conversion

The microphone signals were recorded onto a 16 bit DAT-tape. The data were then transferred digitally to a computer using a program called "Fast Eddie", which saves the data as Windows wave-files. For the vowels, the most stationary 2,5 seconds of each measurement were selected and stored. The selected wave-files were then converted into Matlab-files and at the same time the right and left channel were split into two files and down sampled from 44.100 Hz to 22.050 Hz.

E.3.2 Spectrum analysis

The spectra were computed using the Matlab command PSD = Power Spectral Density. The spectral density is estimated with the Welch method. This method basically divides the time signal into a certain number of overlapping windows. It then multiply the time signal with the first window and performs a FFT. This FFT is then added to the FFT of the next window and so on. In order to unbias the spectrum, the final sum of all the FFT's are squared and normalized with the number of windows and the norm of the window. The theoretical power spectral density, PSD, is given by the auto-correlation:

$$PSD = P_{xx}(\omega) = \sum_{m=-\infty}^{\infty} R_{xx}(m)e^{-j\omega m} \tag{E.2}$$

The Welch estimate of the spectrum can be written as:

$$P_{xx}(\omega) = \frac{1}{K}\sum_{i=1}^{K}\left|\sum_{n=0}^{M-1} x^{(i)}(n)w(n)e^{-j\omega n}\right|^2 \quad ;i = 1,2,....,K \tag{E.3}$$

where;
K = number of windows
M = window length
w = window
n = frequency sample number n
$x^{(i)}$ = time signal in frame number I

The Matlab command is: Pxx=psd(s,Nfft,Fs,windowlength,overlap)

Nfft = length of spectrum
A better resolution requires a longer window, which gives a poorer information of a stochastic time signal. Therefore a comprise must be taken and here is chosen a length of 1024, with a sampling frequency = 22 kHz, the window is 47 ms. If the length of X is a power of two, a fast radix-2 fast-Fourier transform algorithm is used.

Window length
A window length of 30-50 ms is normally appropriate for a speech signal, [Randall, 1987]. If f_s = 22 kHz, a window of 1024 samples corresponds to 46 ms.

Overlap

If the Hanning windows are not overlapped, about half of the data information are lost. A good statistical reliability is obtained with 50% overlapping Hanning windows. A greater overlapping provides no effective improvement in statistical error but the ripple in the spectrum becomes smaller. A nearly perfectly flat time weighting is obtained with 2/3 overlapping, which in practice is applied as 75% overlap using a radix 2 FFT, [Randall, 1987]. A 75% time overlap takes of course longer time to compute than a 50% overlap and if the signal is long it pays off in time to decrease the overlapping. The improvement of the spectrum from 50% to 75% overlapping depends on the signal. For a speech signal, the improvement is not worth to waste time on for this purpose. An example with 50% and 75% overlap is shown in **Figure E.4**. The signal is the time-recorded sound pressure in an occluded ear while the person says 'ice'. The difference between the two spectra is vanishing.

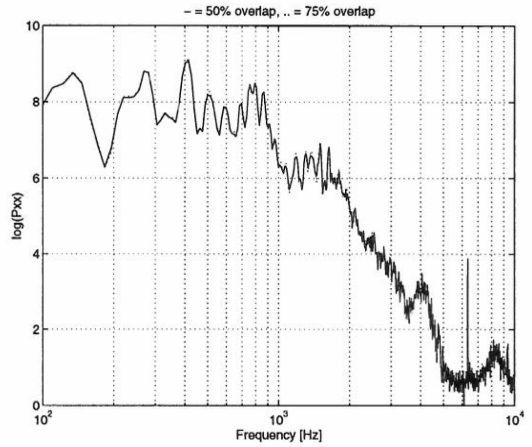


Figure E.4 Spectrum of a time signal (the word 'ice') computed with 50% and 75% overlapping Hanning windows. FFT-length=2048, f_s=25.000, window-length=2048.

E.3.3 Computing of P_c/P_a

E.3.3.1 Continuous speech

Each signal was calibrated according to the preamp setting and from the calibrated signals the difference was computed:

$$\Delta L_p(f)=10\log(p^2_{p,box}(f)/p^2_{p,ff}(f)) - 10\log(rms^2_{r,box}/rms^2_{r,ff})$$

where;
subscript p = probe microphone
subscript r = reference microphone
subscript ff = free field (no box in the anechoic room)
f =frequency

E.3.3.2 Vowels

Difference for the vowels were determined for each single harmonics. It was not always possible to reproduce exactly the same fundamental and harmonics of the vowel, it was therefore necessary to fit the two harmonics to the same frequency. In each measurement it was assured that the fundamental frequency agreed within ± 30 Hz. A Matlab routine was written to fit the harmonics of the spectrum of the free field recording (sp1) to the spectrum of the box recording (sp2).

```
% Move sp2 to the same fundamental freq as sp1
[sp2corr,ifq]=fitsp(sp1,sp2,freq1);
pause

        function [scorr,ifq]=fitsp(sp1,sp2,freq)

            ifq=input("Upper limit index for the frequency of the frequency :');
            [max1,i1]=max(sp1(6:ifq)); % Finds the peak between index 6 to ifq
            [max2,i2]=max(sp2(6:ifq));
            dif=i2-i1;
            tmp=ones([abs(dif),1]);
            if dif >= 0
                    scorr=([[sp2((dif+1):513)];[tmp]]);
            else
                    scorr=([[tmp];[sp2(1:(513-abs(dif)))]]);
            end;
        end;

f2=input('Upper limit index for the frequency of the 2nd. harmonic :');
f3=input('Upper limit index for the frequency of the 3rd. harmonic :');
f4=input('Upper limit index for the frequency of the 4th. harmonic :');
f5=input('Upper limit index for the frequency of the 5th. harmonic :');
f6=input('Upper limit index for the frequency of the 6th. harmonic :');

sp1f=sp1;
sp2corrf=sp2corr;

% Calibration to dB SPL
sp1fc=scalib(sp1f,calcoeff,30,ampsp1);
sp2fc=scalib(sp2corrf,calcoeff,30,ampsp2);

% Difference in dB
h1dif=max(sp2fc(6:ifq))-max(sp1fc(6:ifq));
h2dif=max(sp2fc(ifq:f2))-max(sp1fc(ifq:f2));
h3dif=max(sp2fc(f2:f3))-max(sp1fc(f2:f3));
h4dif=max(sp2fc(f3:f4))-max(sp1fc(f3:f4));
h5dif=max(sp2fc(f4:f5))-max(sp1fc(f4:f5));
h6dif=max(sp2fc(f5:f6))-max(sp1fc(f5:f6));
```

Each signal was calibrated according to the preamp setting and from the calibrated signals the difference was computed:

$$\Delta L_p=10\log(p^2_{p,box}(m)/p^2_{p,ff}(m)) - 10\log(p^2_{r,box}(m)/p^2_{r,ff}(m)) + 10\log(p^2_{l,box}(m)/p^2_{l,ff}(m))$$

where;
subscript l = reference microphone with loudspeaker stimulus
m = m'th harmonic of the vowel

The last part compensates for the disturbance of the box.

F. APPENDIX: EXPERIMENTAL SET UP V:

VIBRATIONS OF TISSUE AND EARMOULD

F.1 LIST OF EQUIPMENT

1 V voltage generator

Microphones	Lectret, model 2141 Knowles, EM 3356 1" Brüel&Kjær, type 4144
Pre-amplifier	Brüel&Kjær, type 2639
Signal processor card	Ariel DSP-16 TMS320C25
Amplifiers	Laboratory construction
Measuring Amplifier	Brüel&Kjær type 2607
Pistonphone	Brüel&Kjær, Type 4220
Software	Hypersignal and Matlab

The set up is shown in **Figure F.1**. The microphone output signals were stored as binary files on a hard disk by the mean of a signal processor card DSP 16+ TMS210C25 and the program 'Hypersignal'. The amplified analogue microphone signal outputs were lowpass filtered by the internal filter in the measurement system. The system was designed such that using a sampling frequency of 50 kHz, aliasing is avoided. The recorded time signals were stored and analyzed with the software program 'Matlab'. In 'Matlab' the average spectrum of a signal was calculated as the power spectrum with 50% overlapping Hanning windows.

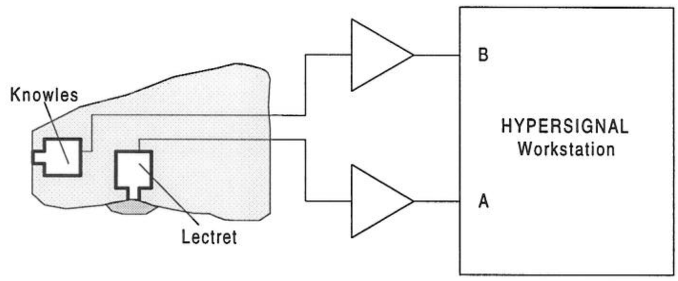


Figure F.1. Set up used to detect movements in the soft part of the ear canal.

F.1.1 Insulation of the rubber enclosure

The rubber enclosure was placed between the earmould and the ear canal wall. The acoustic insulation of the rubber enclosure was measured to be sure that the Lectret2141 microphone response was caused by vibrations and not by air transmitted sound. The insulation of the rubber membrane was 22 dB at 100 Hz and increasing to 36 dB around 4 kHz. At higher frequencies the insulation decreases again to 14 dB at 10 kHz. Hence the insulation of the rubber membrane is adequate in the speech frequency range.

F.1.2 Calibration to dB SPL

In order to calibrate the digitally stored signals to dB SPL it was necessary to make three transformations; one for the microphone sensitivity, Mic_{corr}, one for the values of the sampled binary signal and one for the preamp-settings.

The system was calibrated with a ½" microphone which has the same sensitivity in the whole frequency range 100 - 7000 Hz. It is therefore possible to calculate the compensation for the Lectret 2141 and EM3356 sensitivities by calculating the sensitivity relative to the ½" microphone sensitivity. The ½" microphone sensitivity is shown together with the Lectret 2141 and EM3356 sensitivities in **Figure F.2**. The correction factor, Mic_{corr} is then:

$$Mic_{corr}(f) = 20Log\left(\frac{Sensitivity_{Lectret2141}(f)}{Sensitivity_{B\&K4134}(f)}\right) \quad [dB] \qquad (F.1)$$

The correction factor for EM3356 is calculated in the same way.

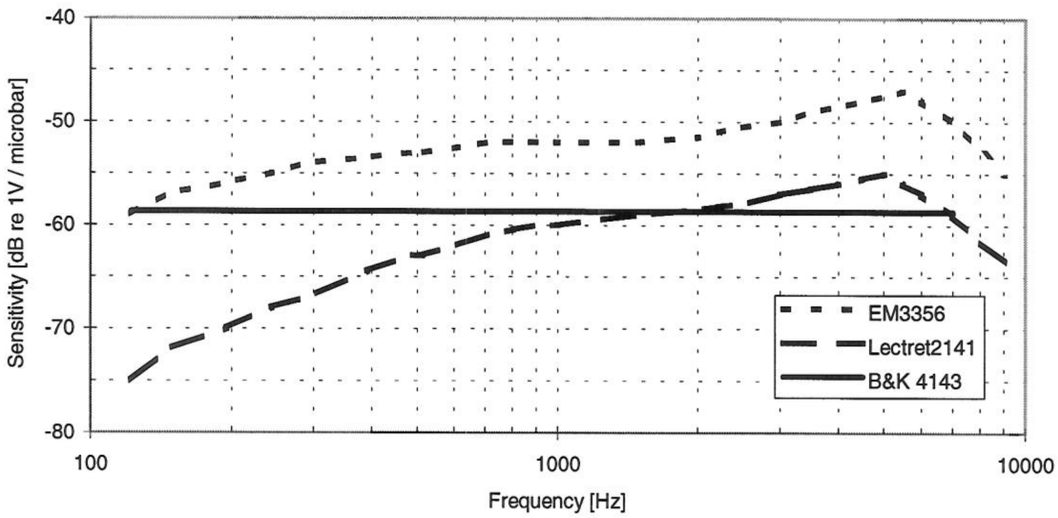


Figure F.2. Sensitivity of ½" microphone and the microphones in the earmould.

The absolute sound pressure in dB SPL of the signal from a microphone was then:

$$p(f) = 20\log\left(\frac{binary\ value(f)}{20\mu Pa}\right) - Mic_{corr}(f) - correction\ for\ binary\ value(f) - preamp$$

G. APPENDIX: EXPERIMENTAL SET UP IV: Earmould insulation

G.1 EXPERIMENTAL EQUIPMENT

Ear simulator	IEC-711 coupler Brüel&Kjær Type 4157 with adaptor DB 2012
Microphone	½" Brüel&Kjær type 4131
Preamp	Brüel&Kjær type 2619
Audio analyzer	Brüel&Kjær tube 2012 Steady State mode, Log ISO R-80, 50-10.000 Hz, complex linear averaging, max. detection 80ms
Pistonphone	Brüel&Kjær type 4220
Loudspeaker	KEF unit in a spherical enclosure
Anechoic box	Oticon box 2

G.2 EXPERIMENTAL METHOD

The attenuation of a foam mould, an acrylic plug and a plug of impression material was measured in an ear simulator B&K 4157 (IEC-711) with a ½" microphone B&K 4131 and a preamp B&K 2619 The earmoulds were inserted into an adaptor (DB 2012) that was screwed on the coupler such that the end of the earmould was placed in the coupler reference plane. The adaptor dB 2012 simulates the most outer part of the ear canal.

The ear simulator was placed about 30 cm under the center of a loudspeaker hanging in an anechoic box. The loudspeaker and coupler were connected to an audio analyzer B&K 2012. The output provided 84 dB SPL at 1,000 Hz. The system was calibrated with a pistonphone (124 dB - 0.7 dB, 250 Hz). The 0.7 dB is the correction according to the adaptor DB 2012.

G.3 RESULTS

Three materials were investigated: and EAR foam plug, Oticon impression material and acrylic. The impression and acrylic was cast in the adaptor DB 2012. The EAR foam plug is larger than the adaptor diameter, so the foam had to be compressed, just as it is when the plug is inserted onto a real ear. There was no leakage other than through the foam material itself. The impression material plug is also tight because the material was put wet into the adaptor and then dries.

The attenuation provided by the moulds were calculated as the closed adaptor response relative to the open adaptor response, see **Figure G.1.** The characteristic shapes of the curves corresponds with data reported in the literature. For example the anti-resonance and resonance peaks for the acrylic curve are similar to Schroeter and Els' [1982] who measured the attenuation of earplugs in an artificial head. The shape of the curve for the foam plug corresponds to data measured on a HATS Brüel&Kjær Type 4128 by Wargowske et al. [1995], for example, the characteristic increase at 6-8 kHz fits perfectly with the literature data.

The attenuations are not exactly the same primarily because the artificial head used by Schroeter and Els also simulates human flesh in the ear canal and damps low-frequency vibrations of the earmould. The ear simulator and the HATS does not simulate flesh.

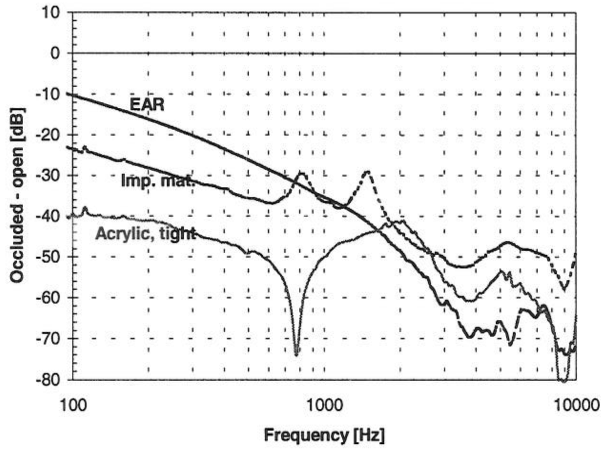


Figure G.1. Difference between occluded and open ear simulator with adaptor DB 2012. The acrylic plug was coated with Vaseline.

The impression material attenuates better than the foam plug and the acrylic material even better. It was necessary to coat the acrylic plug with Vaseline to avoid leakage. In **Figure G.2** the attenuation is shown for three acrylic plugs without Vaseline. It is very easy to see that the plugs provide less attenuation with no Vaseline, now the acrylic plug attenuates less than the EAR plug.

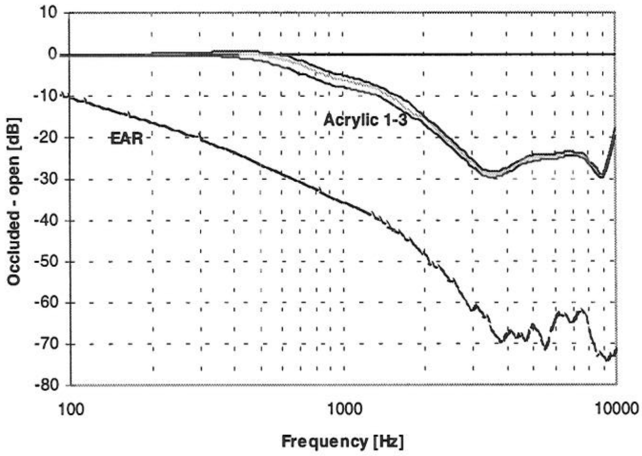


Figure G.2. Measured attenuation in an ear simulator B&K 4157 with an adaptor dB 2012. The acrylic plugs was *not* coated with Vaseline.

H. APPENDX: ACOUSTIC MODEL OF THE EAR

H.1 THE MIDDLE EAR

The acoustics and models of the eardrum and the middle ear is discussed here. The anatomy of the middle ear is described in chapter 5.

H.1.1 Acoustic impedance of the middle ear

The middle ear performs a linear sound transmission as long as the incoming sound is less than 85 dB SPL. At this sound level the stapedius reflex begins to work. The stapedius reflex reduces the sound transmission and it is there to protect the inner ear against high level input. Another non-linear effect arises when there is a sudden change in static air pressure. This is in effect when the Eustachian tube cannot equalize the pressure in the middle ear cavity to the pressure in the ear canal. The result is a lower acoustic sensitivity, it is for example common to be 'hearing impaired' when you have a cold.

Several investigators have studied the impedance of the eardrum and the middle ear seen from the ear canal, [Hudde, 1983], [Stinson and Shaw, 1982] and [Voss and Allen, 1994]. The behavior of the eardrum and the middle ear is simple at frequencies below 1 kHz. The eardrum makes piston-like movements. At higher frequencies the eardrum moves more complicated because of higher vibration modes [Stinson and Shaw, 1982]. Below 300 Hz, the middle ear cavities are dominant and the measured impedance acts like an acoustic compliance, [Peake and Rosowski, 1997]. The middle ear then behaves more like a resistance around 1-3 kHz. At even higher frequencies the behavior is more complicated and may vary a lot between individuals [Voss and Allen, 1994].

H.1.2 Model of the eardrum and middle ear

The well-known Zwislocki model of the middle ear (details in [Zwislocki, 1962]) will be used here in a version modified by Shaw [1977], where the eardrum model is modified. In Zwislocki's model the part of the eardrum that is coupled to malleus can move independently of the rest part of the eardrum. In Shaw's model the eardrum is modeled as two rigid pistons. The model presented by Shaw in 1977 used trial values only. The parameter values has later been adjusted to make the best fit [Stinson and Shaw, 1996]. More details about the modified Zwislocki model is given in [Shaw and Stinson, 1983]. The modified middle ear network are shown in **Figure H.1**

In the lumped parameter model in **Figure H.1**, the bone vibrations is considered as a movement of a mechanical mass and is modeled as an inductance. Compression or stretch of ligaments and tendons follow Hookes law as long as the displacements are small. This means that the restoring forces can be considered as a capacitor. Resistance occurs from friction between tissue and viscous and thermal losses.

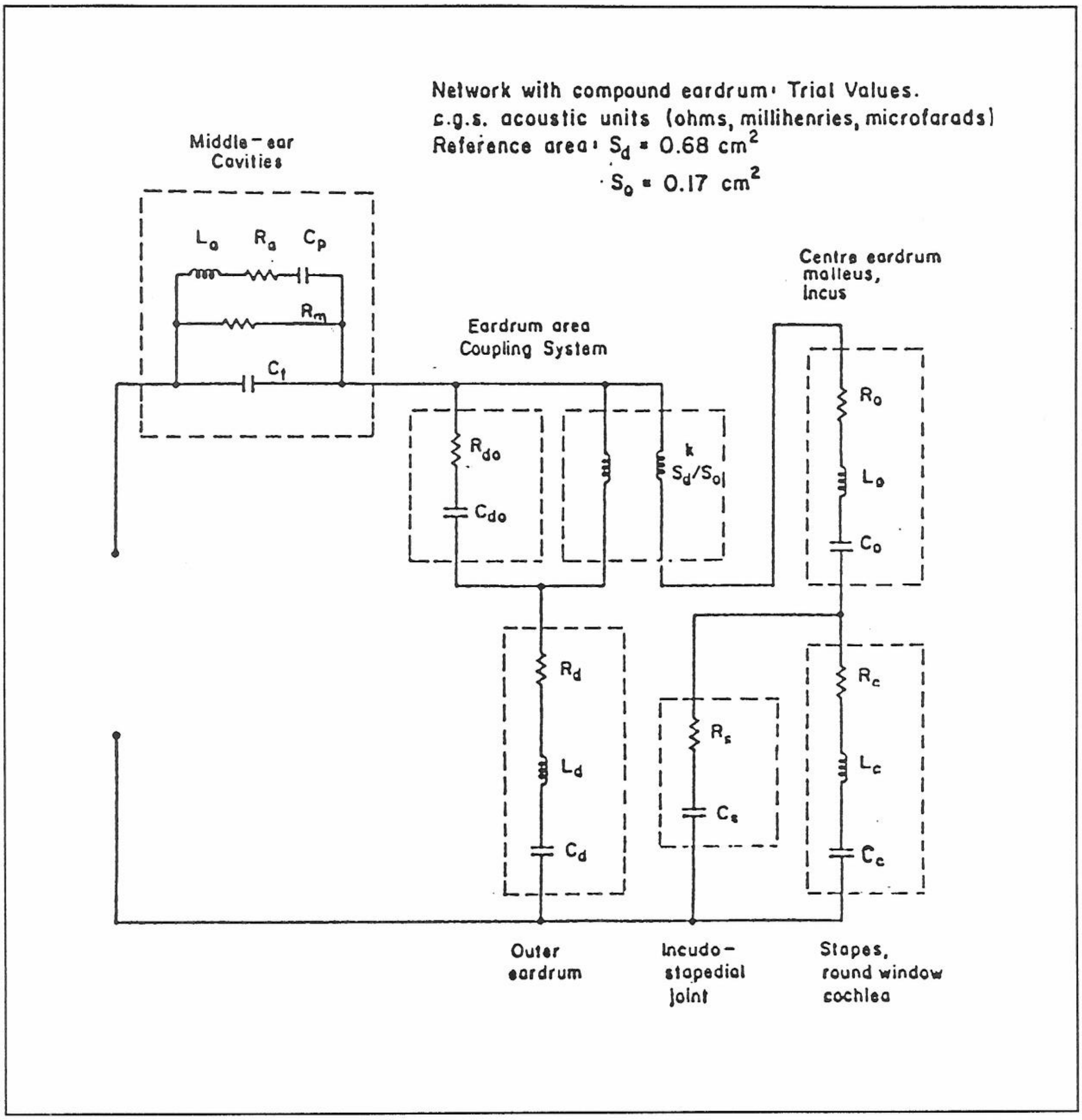


Figure H.1. Middle ear network. Values in acoustic units m.kg.s. Reference area: S_d=68mm^2 and S_o=17mm^2. From: [Shaw and Stinson, 1983].

Shaw's and Stinson's model is compared with Zwislocki's model in **Figure H.2**. The main difference between the two models is that the impedance in Zwislocki's model has two peaks at 1-3 kHz whereas Shaw and Stinson's model only has one peak around 2 kHz.

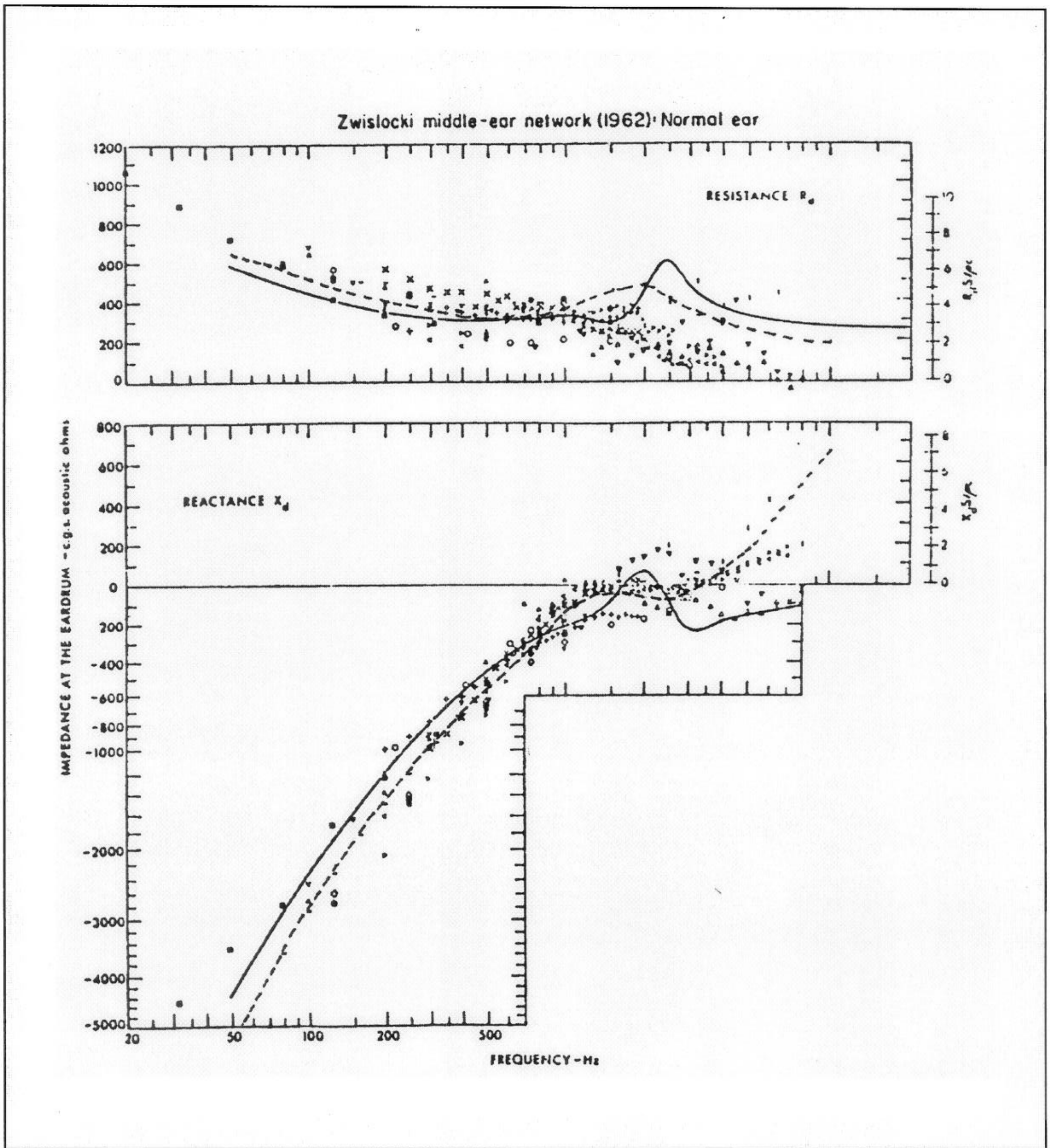


Figure H.2. Middle ear impedance. Points: Measured values, [Shaw E.A.G., 1977]. Solid line: Zwislocki model, [Shaw E.A.G., 1977]. Dashed line: Shaw and Stinson values, [Stinson and Shaw, 1996].

The effect of applying the middle ear impedance to a rigid ear canal is shown in **Figure H.3**. A tube terminated with the modified Zwislocki impedance network is compared to a rigidly ended tube. The tube is 23.5 mm long and 7.1 mm in diameter. The figure shows the difference of the sound pressure at the end of the tube (in front of the eardrum) to the incident sound pressure. The incident sound pressure was defined as 60 dB SPL and phase 0° at all frequencies.

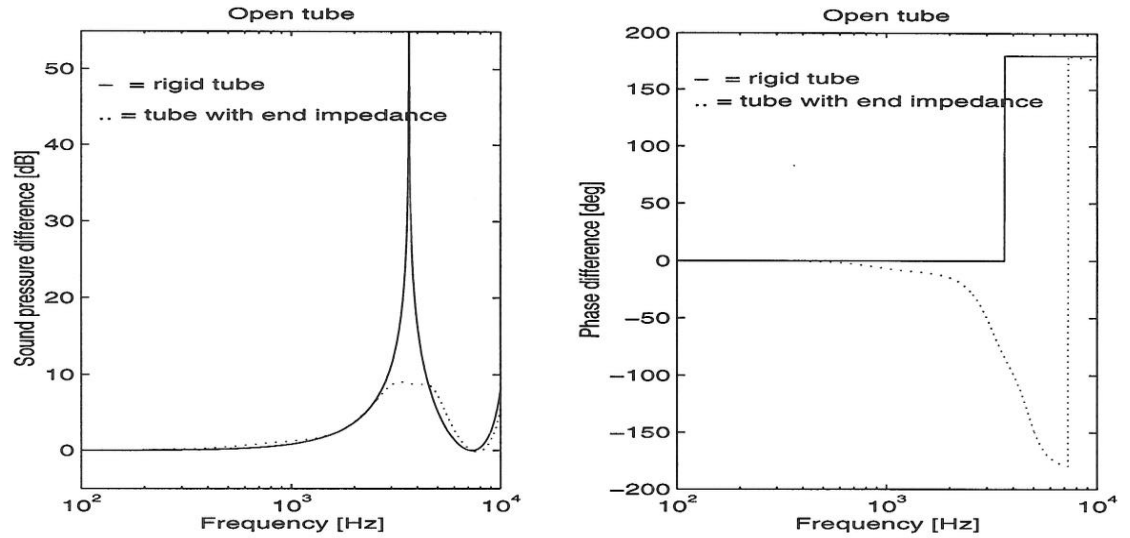


Figure H.3. Effect of the modified Zwislocki impedance network. Difference between the sound pressure at the end of the tube and at the entrance. Input: sound pressure at entrance to the open tube. Left: magnitude. Right: phase.

The main effect in the open tube is that the eardrum and middle ear impedance damps the resonance frequency. The phase has a negative shift instead of a positive shift. The middle ear impedance is insignificant below 2 kHz in the open tube. But in the closed tube, see **Figure H.4**, the impedance attenuates the sound pressure with 5 dB at low frequencies.

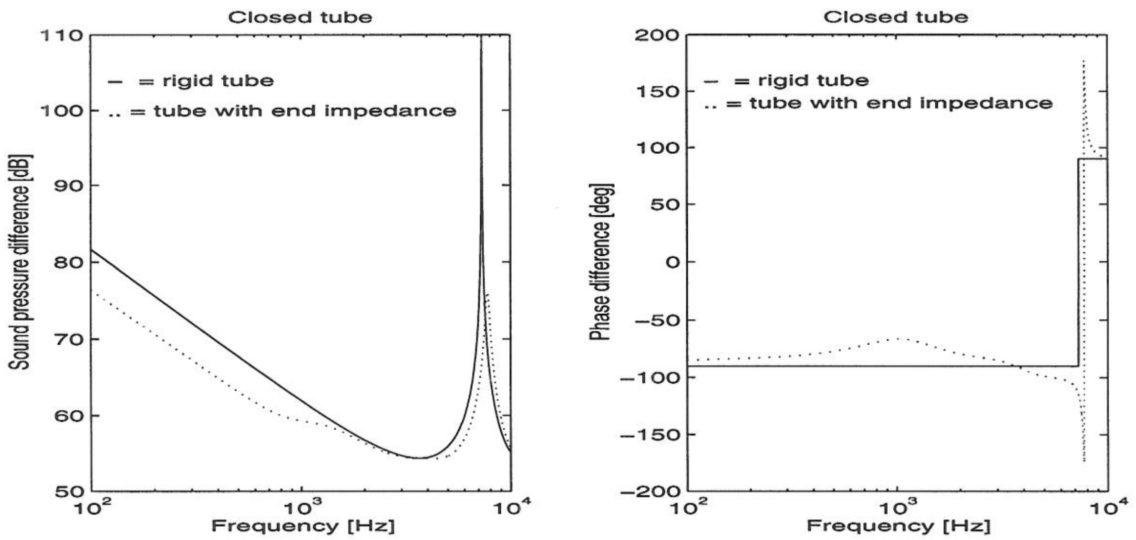


Figure H.4. Effect of the modified Zwislocki impedance network. Sound pressure and phase at the end of the tube and at the entrance. Input: volume velocity arbitrary chosen to 10^{-6} mm^3/s. Left: magnitude. Right: phase.

H.2 MODELS OF THE EAR CANAL

H.2.1 Cavity model

The ear canal can be treated as a cavity up to 1 kHz [Stinson and Lawton, 1989]. The sound pressure at 1 kHz changes only 2 dB along the length of the tube compared to a cavity. Theoretically, the sound pressure is nearly the same all over the tube below 1kHz and a lumped element model of the ear canal is reasonable. In a cavity the sound pressure is the same all in all points and the shape of the cavity is therefore is not important.

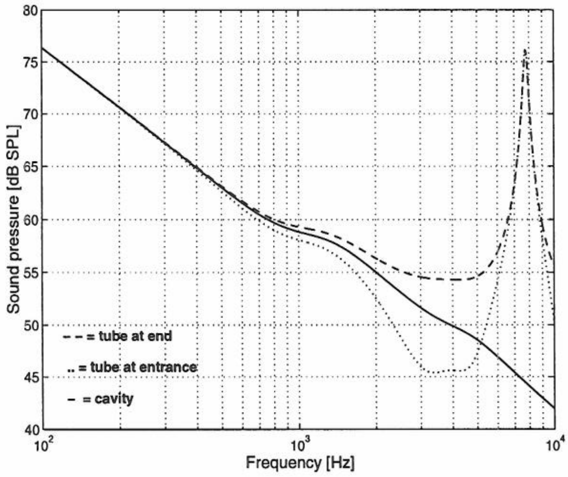
H.2.2 Uniform tube model

A bit more advanced model of the ear canal is a uniform tube with rigid walls and the eardrum positioned perpendicular to the canal walls. Wiener and Ross showed in [1946] that a cylindrical tube was adequate to model the ear canal response including the λ/4 resonance and the λ/2 anti-resonance

Stinson and Lawton [1989] state that the uniform tube model works well for f < 4 kHz. Furthermore, if the part right in front of the eardrum is neglected, the uniform tube model is also adequate up to 8 kHz. As explained earlier, higher modes are trapped at the eardrum and the plane wave tube model is therefore no longer valid in front of the eardrum.

The individual ear canal is of course not a cylindrical uniform tube. However, below 2 kHz a uniform tube model with average human dimensions is adequate, [Stinson, 1985]. At higher frequencies the individual ear canal dimensions becomes noticeable. In a general model it is not possible to account for individual ear canal shapes.

The closed tube and the cavity models are compared in **Figure H.5**. The terminating impedance of the tube was the modified Zwislocki network. This impedance was set in parallel to the cavity model. It is clear that the tube acts as a cavity up to 1 kHz. At higher frequencies the sound pressure in the tube depends on the position inside the tube. The open tube resonance is seen as a dip in the closed tube curve, the difference is about 5 dB between a cavity and uniform tube. The quarter-wave resonance and the half-wave anti-resonance is not present in the cavity model. Consequently the phase for the cavity model does not shift as the tube model phase.



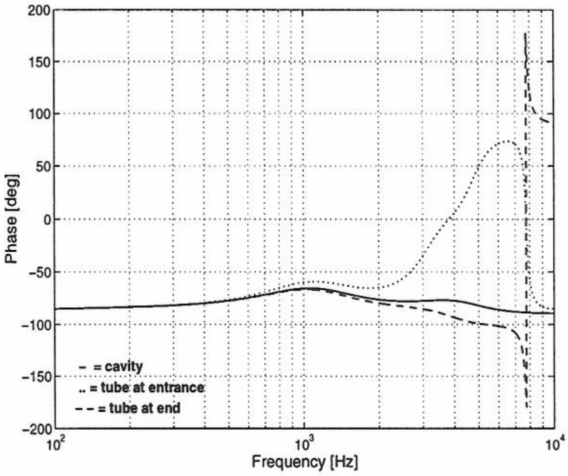


Figure H.5. Amplitude and phase of calculated pressure inside a cavity and a tube terminating with the impedance of the middle ear and eardrum. Length of tube = 23.5 mm, diameter = 7.1 mm. Volume velocity source arbitrary chosen to 10^{-6} mm^3/s.

The occlusion effect should be predicted well up to 2 kHz and in order to apply a hearing aid the model should be valid at least up to 6 kHz. Hence, the tube model is more adequate than the cavity model.

A tube is a transmission line and the measured sound pressure depends on the position of the probe microphone in the ear canal. A tube with length, l, and diameter, d, terminated with an impedance, Z_{end}, is sketched in **Figure H.6**.

The specific impedance, Z_0, is the impedance of the tube if there were no reflections, i.e. the terminating impedance should absorb the incident sound 100% so $Z_{end} = 0$. If the tube is rigid, the terminating impedance is infinite. Looking into the ear canal, Z_{end} represents the eardrum and the middle ear.

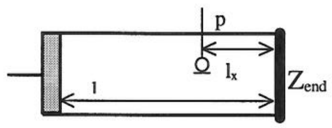


Figure H.6. Uniform tube closed in one end with an arbitrary impedance, Z_{end}. The piston in the other end creates a sound pressure p in point X.

In **Figure H.6** the sound pressure is measured in point X and the sound pressure is given by, [Jacobsen, 1993]:

$$p(\omega) = q(\omega)\frac{Z_{end}\cos kl_x + jZ_0 \sin kl_x}{j\dfrac{Z_{end}}{Z_0}\sin kl + \cos kl} \qquad (\text{H.1})$$

where;
l = total length of the tube
l_x = length from point X to Z
Z = terminating impedance
Z_0 = specific impedance = $(\rho c/S)$
S = cross-section area

Wiener and Ross [1946] measured the pressure transformation from the entrance to 1 mm in front of the eardrum. Shaw [1974b] took these data and data measured by Djupesland and Zwislocki and fitted a curve to the data. This curve is printed in **Figure H.7** along with calculations derived from equation (H.1) with l = 23.5 mm and d = 7.1 mm.

The calculated magnitude (dashed line) at the resonance frequency is about 1 dB lower than Shaw's curve. This is acceptable because the real ear data also deviates 1 dB from the Shaw's pooled data. If the impedance was higher and the eardrum had reflected more sound, the resonance peak would not be damped so much. The Zwislocki middle ear impedance is actually greater than the modified impedance between 3-4 kHz, see **Figure H.2**. But if the estimated data are compared to the measured data, also shown in **Figure H.2**, the modified model seems to fit the data better than the Zwislocki model, also at 3-4 kHz.

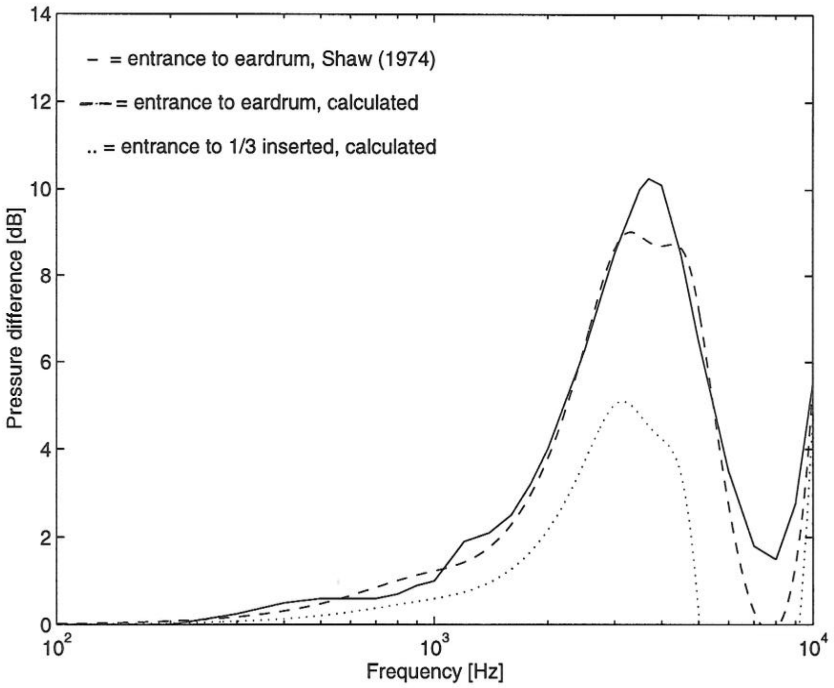


Figure H.7. Transfer function from inside the ear canal and at the entrance. Calculated curves and curve fitted to measured data by Shaw [1974b]. Ear canal length = 23.5 mm, diameter 7.1 mm.

The dashed curve in **Figure H.7** is the calculated difference between the sound pressure 1 mm in front of the eardrum and at the entrance. At very low frequencies, the plane wave nearly does not propagate, because the pressure at the eardrum is the same as at the entrance. Around 300 Hz it starts to propagate and the difference is still less than 2 dB up to 1500 Hz. The dotted curve shows the transfer function when the pressure is measured 1/3 in from the entrance.

H.2.3 A 'horn'-like model

The physical shape of the ear canal is more 'horn'-like than 'tube'-like. In the open ear canal, the tube model is quite precise up to 4 kHz and with some restrictions up to 8 kHz. In the occluded ear canal the non-uniform shape of the ear canal matters because the pressure depends on the enclosed volume. The pressure in a horn-like ear canal model is significantly higher than in a uniform tube, when the insertion depth of the earmould is the same. This was discussed in chapter 11.

The sound propagation in the ear canal can be calculated with the horn equation and has been used for cat ear canals up to 29 kHz, [Stinson and Khanna, 1994]. Details about the horn equation used to predict the sound pressure in the human ear canal can be found in Stinson and Khanna [1989] and Stinson and Lawton [1989].

H.2.4 Variations in individual ear canal size

The curves in **Figure H.7** was calculated with a ear canal length of 23.5 mm and diameter 7.1 mm. these values are average values but the individual dimensions might be very different. The data are taken from Salvenilli et al. [1991] who reports that the standard deviation on the ear canal length is ±2.5 mm. The ear canal cross-section is more oval than circular and it has a long and a short diameter. The long diameter is 9.4 ±1.5 mm and the short one is 4.8 ±0.5 mm. It gives an average diameter of 7.1 ±1.0 mm. The effect of these variations are calculated in **Figure H.8** and compared with the Shaw curve. A longer ear canal has a lower resonance frequency. A larger diameter amplifies the resonance peak but it does not shift the resonance frequency.

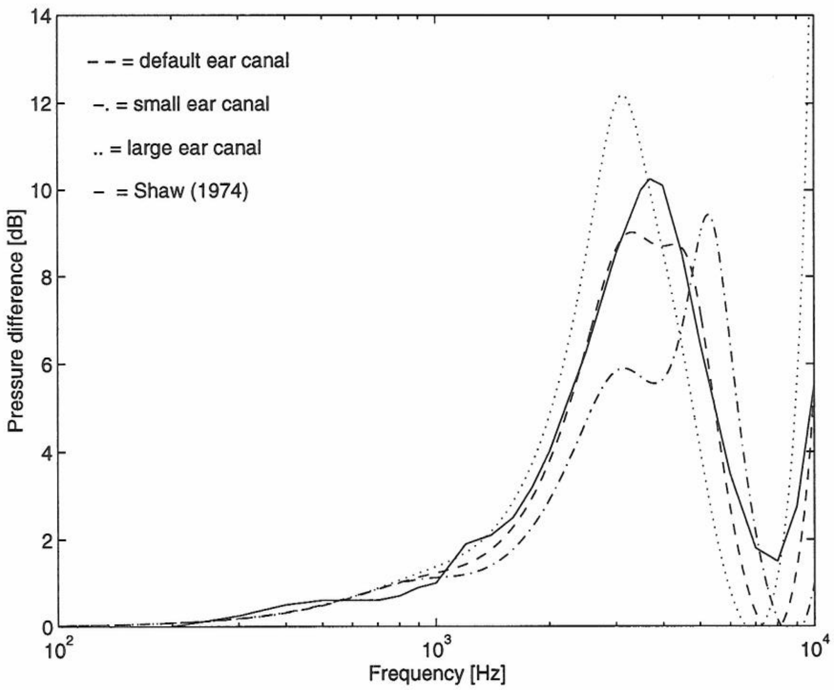


Figure H.8. Transfer function from the entrance to the eardrum. Calculated curves and a curve fitted to measured data by Shaw [1974b]. Default canal: length = 23.5 mm, diameter = 7.1 mm. Small canal: length = (23.5 - 2.5) mm, diameter = (7.1 - 1) mm. Large canal: length = (23.5 + 2.5) mm, diameter = (7.1 - 1) mm.

H.2.5 Ear canal resonance

The transmission from the ear canal entrance to the eardrum is plotted in **Figure H.7**. It shows a resonance at 3.7 kHz. If the sound source is placed outside the ear canal, then the resonance now appears around 2.6 kHz. This resonance is termed 'ear canal resonance'. Shaw [1974b] combined results from 12 studies and found that the real ear resonance lies around 2.5-2.7 kHz.

It has already been discussed that the effective acoustic length of the ear canal is longer than the physical length. The phenomenon is probably due to the concha and the foldings of tragus and crus helias. If the ear canal had rigid walls, the resonance should be at f_0 = 343 m/s / (4·0.0235 m) = 3.6 kHz. A resonance of 2.6 kHz corresponds to an ear canal of 33 mm. Now, the ear canal is not rigid and the eardrum and the middle ear attenuates and shifts the peak of the resonance a bit and an end correction of 7.5 mm to the physical length of 23.5 mm gives a canal resonance of 2.6 kHz. The calculated ear canal resonance is in **Figure H.9** compared to the composite made by Shaw [1974b] with incident waves at 0° and 90° to the nose.

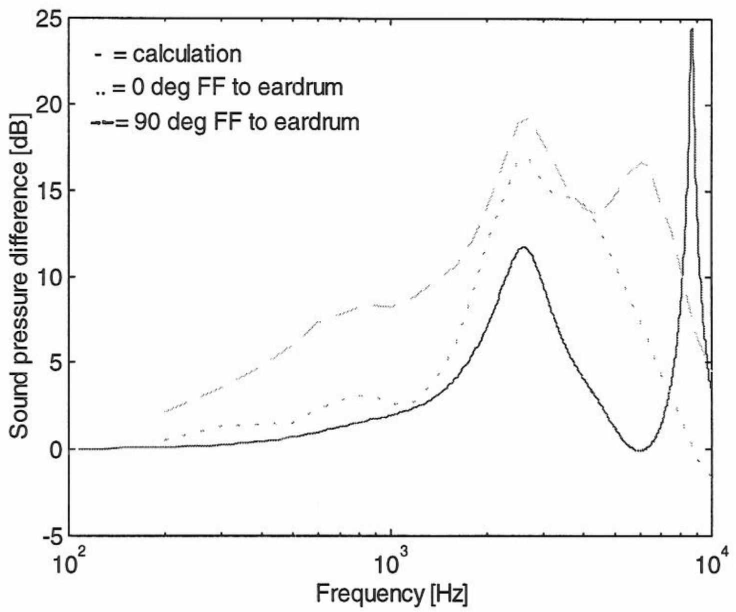


Figure H.9. Transformation of sound pressure from the free field to the eardrum. The 0° and the 90° curves shows the difference between the free field at the subjects position and to the eardrum. From: [Shaw and Vaillancourt, 1985]. The calculated curve uses an ear canal length of 23.5 + 7.5 mm.

The calculated ear canal resonance frequency corresponds well with Shaw's composite. Notice, that the ear canal resonance frequency is independent on the angle of incidence. The calculated resonance magnitude is 5-8 dB lower than the measurements because the acoustic effect of the body and pinna is not yet included in the calculations. The curves are also different shaped and the shape depends on the angle of incidence. The different shapes are mainly caused by diffractions from the concha, pinna flange and the head. These components are analyzed in the next section.

H.3 ACOUSTIC FEATURES OF THE PINNA AND HEAD

The pinna consist mainly of the pinna flange and the concha cavity. The concha and the pinna flange are most efficient at frequencies above 4 kHz. Below 2 kHz the pinna and concha has nearly no effect. The influence of the body features in a free field are shown in **Figure H.10** for 45° incident sound in the horizontal plane. The concha has a resonance at 5-6 kHz and it contributes with 10 dB when the incident sound comes from and angle 45° relative to the nose (45° azimuth). The pinna flange amplifies the sound with a few dB (3-6 kHz) when the sound wave comes from the front of the ear. If the sound wave comes from behind, then pinna reduces the sound pressure in the ear canal, especially at 4 kHz (not shown).

At 10 kHz the head contributes with 5-6 dB. The torso (the upper body) and neck amplifies 2 dB at 100-700Hz and reduces the sound in the frequency range 1-2 kHz. The influence of torso and neck is insignificant above 2 kHz.

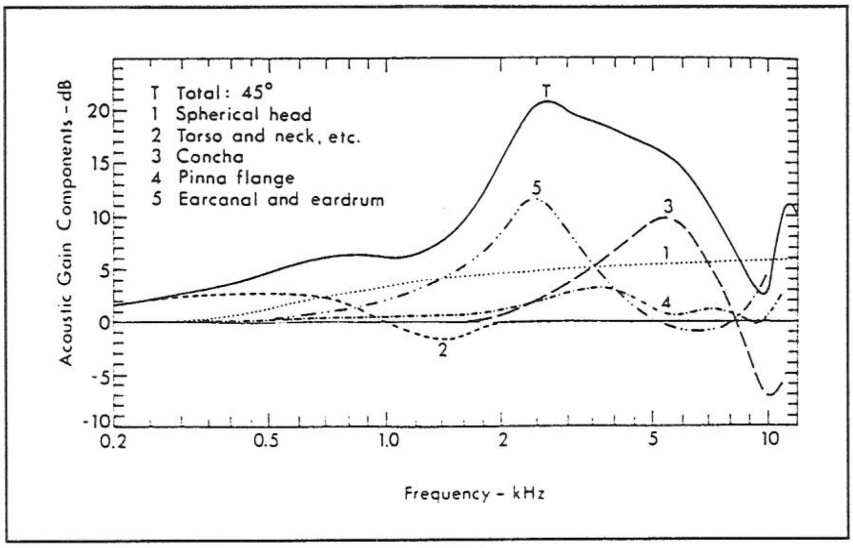


Figure H.10 Average pressure gain at the eardrum from an outer source placed in 45° azimuth. From: [Shaw, 1974a].

The curve in **Figure H.10** for the eardrum pressure is only valid for 45° of incidence. The curve shape above 4 kHz highly depends on the angle of incidence because of the acoustics of the concha cavum and the pinna. The reason is that higher modes occurs in the concha cavity and the diffractions from the pinna. The resonance of the concha shifts downwards in frequency when the source is moved downwards and the resonance shifts upwards in frequency when the source is moved to upwards. An example of the effect in the horizontal plane can be observed from Møller's [1992] measurements on 4 subjects. Møller concludes that plane waves exist even 6 mm outside the ear canal entrance. A bit further out, diffractions from pinna flange and concha disturbs the sound field. At 90° azimuth, concha creates a resonance at 6 kHz and an anti-resonance at 7-8 kHz. These resonance peaks disappears when the sound source is placed at 0° azimuth.

The head does not influence the resonance frequencies but has a shadowing effect. The shadow effect is for example shown in measurements from Møller et al. [1995]. Their aim was to find the head related transfer functions from the free field to the eardrum. The sound pressure in the ear canal entrance with the ear canal blocked was measured on 40 subjects. Comparisons between the response from a sound source placed in the centerline of the nose (0° azimuth) and a sound source placed in the centerline of the ear canal (90° azimuth) reveals that, if the source is placed at 90° azimuth, the sound pressure in the ear canal entrance becomes greater than if the source is placed in front of the nose (0° azimuth) . The head influences the sound field in the whole the frequency range. For example, at 90° azimuth the pressure is about 5 dB higher than at 0° azimuth below 1 kHz.

H.3.1 Models of concha

Schroeter and Poesselt [1986] modeled the concha as a simple radiation impedance, see chapter 3. But this model is a bit too simple because the $\lambda/4$ resonance of concha is not applied. One way to model the concha is to approximate the concha with a small tube of diameter 22 mm and depth 10 mm in a physical model (Shaw, 1974). The radiation impedance can be modeled as a piston in an infinite baffle. The simple model of concha is used in ear-like couplers. In the Zwislocki coupler the concha is a cylindrical cavity with diameter 25 mm and depth 9 mm connected to an ear canal of length 22.5 mm, [Ballachanda, 1995]. The tube model was used by Weinrich [1986]. He modeled the concha as a cylindrical tube of length 10 mm and with a diameter of 15 mm.

The ear canal entrance is connected to the concha in the side and not in the middle of concha. This asymmetrical placement has greatest effect above 6 kHz, [Shaw, 1974a] and it is therefore still acceptable to model the ear canal and the concha as two tubes connected in the same centerline. In the present model, only the part of concha, where plane wave exist, will be modeled as a tube, the sustaining part will be included in the radiation impedance. The effective volume of concha is 2.5 cm^3 , [Shaw, 1974b].

H.3.2 Radiation from the ear

Sound will be radiated out of the ear either from reflections or from direct sound produced in the ear canal. The pinna flange is neglected in the following and it is imagined that the concha ends in the side of the head.

Presuming that the head is placed in a plane wave field, then the radiation from the ear can be estimated with a rigid circular piston in a sphere. One more simplification of the situation could be to consider the head as an infinite baffle. This would be true at frequencies where the wave length is much smaller than the radius of the head. The average human head is 15.1 cm wide, 18.8 cm long and 13.0 cm high, [Burkhard and Sachs, 1975]. These measures approximates a diameter of 15 cm if the head was a sphere. Then, with a radius of 15/2 cm = 7.5 cm, the frequency must be smaller than $c/(2\pi*0.075)$ = 728 Hz (c = 343 m/s) if the infinite baffle shall provide fairly exact results.

A rigid piston in an infinite baffle has a radiation impedance, $Z_{rad,}$ given by, [Kinsler et al., 1982]:

$$Z_{rad} = \frac{\rho c}{\pi a^2}\left[R_r(2ka) + jX_r(2ka)\right] \qquad \text{(H.2)}$$

where;
a = radius of the piston

The radiation resistance, R_r, and the radiation reactance, X_r, is given by:

$$R_r(2ka) = 1 - \frac{2J(2ka)}{2ka}; \quad X_r(2ka) = \frac{2H(2ka)}{2ka} \qquad \text{(H.3)}$$

where;
J = Bessel function of 1'st order
H = Struve function of 1'st order

If $(2ka)$ is small, then the Bessel and Struve functions can be written as series and the radiation resistance and reactance becomes:

$$R_r(2ka) = \frac{(2ka)^2}{4\cdot 2} - \frac{(2ka)^4}{6\cdot 4^2\cdot 2} + \frac{(2ka)^6}{8\cdot 6^2\cdot 4^2\cdot 2} - \cdots \qquad \text{(H.4)}$$

$$X_r(2ka) = \frac{(4/\pi)(2ka)}{3} - \frac{(4/\pi)(2ka)^3}{5\cdot 3^2} + \frac{(4/\pi)(2ka)^5}{7\cdot 5^2\cdot 3^2} + \cdots \qquad \text{(H.5)}$$

The radiation impedance for the concha with 22 mm in diameter is practically the same whether it is calculated with three components from the Bessel and Struve series or with only the first component. The difference becomes greater with frequency and at 10 kHz, the difference is less than 0.5 dB (this calculation is not illustrated). Hence, only the first component from the series are needed and the expression for the radiation impedance becomes:

$$Z_{rad} = \frac{\rho c}{\pi a^2}\left[\frac{(ka)^2}{2} + j\frac{8(ka)}{3\pi}\right] \qquad \text{(H.6)}$$

The approximation in (H.6) can be derived to a lumped element model consisting of a pure resistance and a mass inertia. The resistance is found to be:

$$R_{rad} = \frac{\rho c}{\pi a^2}\left(\frac{128}{9\pi^2}\right) \qquad \text{(H.7)}$$

The mass element is expressed as:

$$L_{rad} = \frac{\rho}{(\pi a^2)}\frac{8a}{3\pi} \qquad \text{(H.8)}$$

The resistance is much smaller than the mass inertia when $(ka) \ll 1$. The mass can be seen as if the piston was loaded with a cylindrical air tube with length $0.85a$ and the same cross-sectional area as the piston. Thus, the radiation impedance can be modeled simply as an extension of the tube (the concha).

Schroeter and Poesselt [1986] used the end correction method. They neglected the concha and added 10.5 mm to the ear canal which was 18 mm. Their ear canal radius was 3.75 mm and according to equation (H.8) the additional length should be 3.1 mm. However, they added a longer length in order to match the ear canal frequency better. Their method agrees with Shaw's [1978] studies and conclusion.

Shaw concluded also that the fundamental resonance of the concha corresponds to a length nearly twice the physical length of concha. It agrees well with the end correction which gives a length of 10 mm + 0.85·(22/2) mm = 19.4 mm.

The influence of the radiation impedance is illustrated in **Figure H.11**. As expected, the end correction moves the resonance frequency downwards. The radiation impedance for a piston in a baffle moves of course the resonance frequency to the same as the end correction but it also damps the magnitude of the resonance.

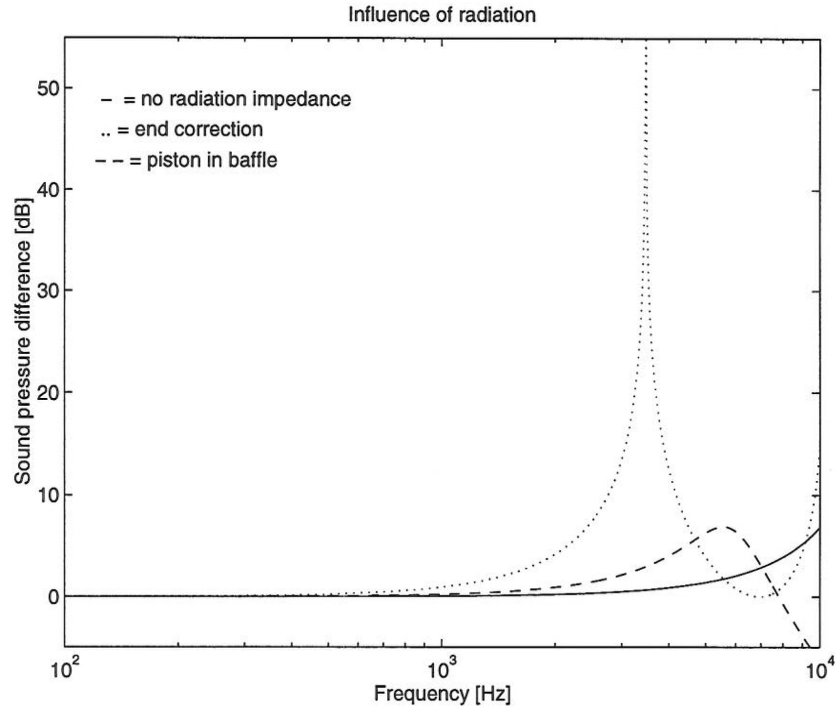


Figure H.11. Influence of radiation impedance on the sound pressure in bottom of concha to a sound source outside the ear. Blocked ear canal. Concha length = 11 mm, diameter = 22 mm. End correction = 0.85·a. Piston in baffle uses (H.6).

Until now it has been assumed that the head acts like an infinite baffle. This is in the theory not valid above 700 Hz. If the head has a shape as a sphere, then the radiation can be calculated as a piston in a sphere. Just like the piston in a baffle, the piston an a sphere can be expressed fairly simple, [Stevens, 1997].

$$Z_{rad} = \frac{\rho c}{\pi a^2}\left(\frac{\pi f^2}{c^2}\pi a^2\right)K_s(f) + j2\pi f\frac{\rho 0.8a}{\pi a^2} \qquad \text{(H.9)}$$

where;
$K_s(f)$ = factor that accounts for the baffling effect of the sphere
a = radius of the piston

The radiation impedance for a piston in a sphere is a bit more complicated than the baffle because the factor $K_s(f)$ depends on the frequency, on the size of the piston and on the angle from a line though the piston center. Equation (H.9) has particularly been used to model the human speech production and so values for $K_s(f)$ is determined for the mouth and not the ear. At low frequencies $K_s = 1$ and it then increases with frequency to a maximum of 1.7 at 2 kHz. Between 2-6 kHz $K_s = 1.5$ in average for the radiation outside the center from the mouth. It is not far from the truth to say that the mouth opening during speech in average is the same size as the concha, so a value of $K_s = 1.5$ is used. The diameter of the head is assumed to be 15 cm.

In order to evaluate how well the concha model and the radiation impedance predicts the real ear situation, the calculations shall be compared to measurements on real ears. Real ear data is obtained from head-related transfer functions [Møller et al., 1995] transposed with the effect of the head [Shaw, 1974a]. The radiation from the ear is highly depending on the angle to the center line of the ear at frequencies above 4 kHz. The model uses a reference point in the centerline of the ear canal and the radiation from the ear shall there be 'seen' from a point on the centerline. According to the principle of acoustic reciprocity, the source and the receiver in a sound field can be exchanged. Now, if the sound source is placed outside the ear and the sound pressure measured inside the ear, it is possible to compare the calculations with free field measurements.

Møller et al. [1995] measured the response at the entrance to the ear canal, when the ear canal was blocked. The sound source was placed in the centerline of the ear. That response cannot be used directly because it includes the effect of the head and torso and the interesting response is the transfer function from mid-concha to the ear canal entrance.

The effect of the head from a 45° angle azimuth is printed in **Figure H.10** as estimated by Shaw [1974a]. In **Figure H.12** The measured transfer function from free field to bottom of concha (dotted) is printed along with the head corrected data (dashed). The dashed curve is not zero at 2 kHz, because the head correction is too small. The influence of the head is greater at 90° incidence than at 45° [Møller et al., 1995]. It is therefore necessary to bring the dashed line down to zero at 2 kHz and a rough way to do that is to simply subtract 2 dB. The result is the solid line.

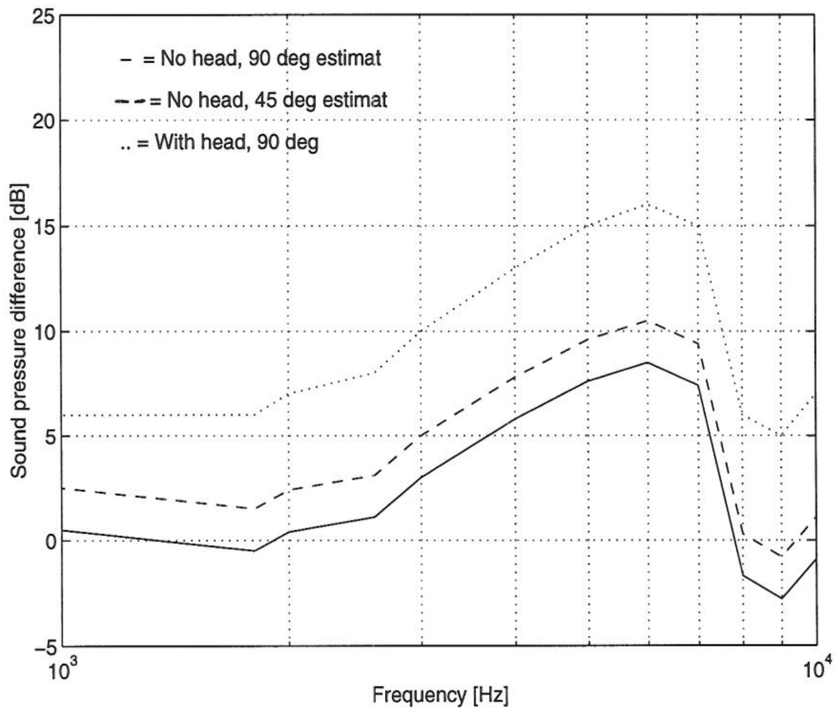


Figure H.12. Transfer functions from the ear canal entrance with blocked ear canal to free field. Dotted line: sound incidence 90° azimuth to the nose, [Møller et al., 1995]. Dashed line: the solid line minus the influence of a spherical head, 45° azimuth, from [Shaw, 1974a]. Solid line: Dashed line minus 2 dB to correct to 90° azimuth.

The solid line in **Figure H.12** now represents the transfer function from somewhere near concha to the ear canal entrance. This curve can then be compared to an estimate of the transfer function from the reference point (mid-concha) to the ear canal entrance, it is done in **Figure H.13**.
The dashed curve is the calculated transfer function with the radiation impedance modeled as a rigid piston in an infinite baffle. The dotted curve shows the calculation with a piston in a sphere and this model is better than the baffled piston model.

The calculated curve with a piston an a sphere estimates the measured curve fairly well, but not exactly. The deviations occurs because the model does not include the effect from the torso and neck and the pinna flange.

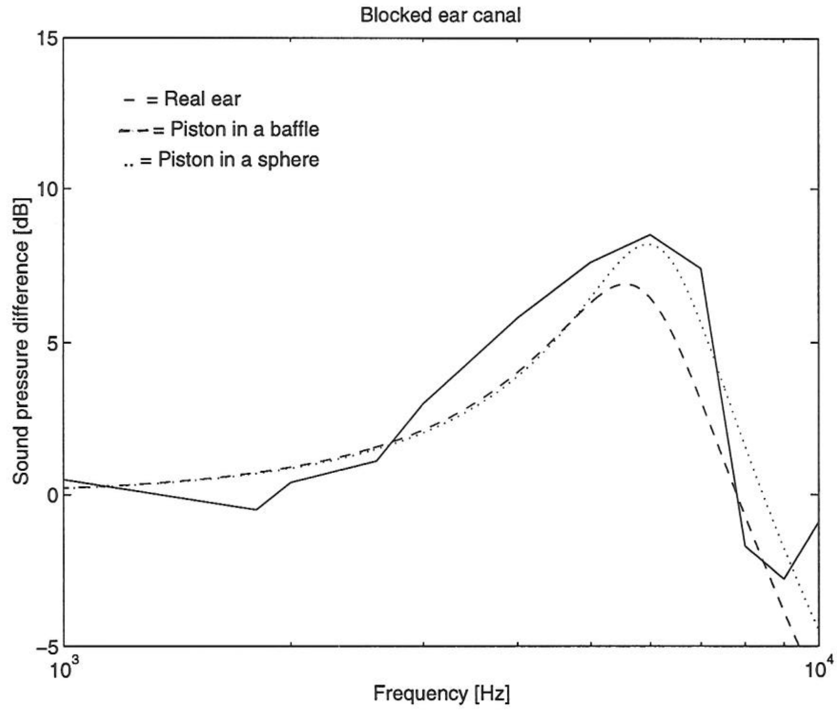


Figure H.13. Transfer function from the ear canal entrance to the reference point 6 mm out in concha. Solid line: corrected real ear measurement with blocked ear canal (the same as in **Figure H.12**). The other two curves are calculated curves.

I. APPENDIX: HUMAN FLESH IMPEDANCE

Most literature on vibrations in human tissue are concerned with very low frequencies (f < 200 Hz) or frequencies in the MHz range. There is very few data on vibrations of human tissue at the auditory frequencies.

One of the earliest studies of the tissue impedance over a broad frequency range was done by Oestreicher [1951] who modeled the behavior of an elastic viscous material in a oscillating sphere and applied that to human muscle tissue. Oestreicher's model agrees well with the mechanical impedance of the finger tissue, Gierke and Brammer [1995].

The skin impedance is in the literature often modeled with a lumped element network as shown in **Figure I.1**. The elasticity of the tissue is represented with the compliance Cs, the mass of the piece of tissue being tested is Ms, and the resistance is Rs. These values depends on the area of the tissue piece that is being tested. In order to be able to compare data from different surveys, it is necessary to normalize the values to the same test area.

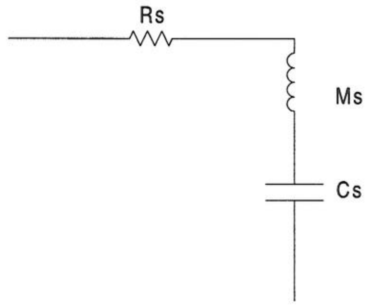


Figure I.1 Lumped element model of the impedance of human tissue.

Measured data from 6 studies are compared in **Figure I.2** All impedances are converted into mechanical units corresponding to a circular disk area of 1.75 cm^2, as prescribed in ISO/IEC 373. According to [Håkansson et al., 1986] it is acceptable to convert all data into the same disk area because the tissue impedance is proportional with the area. The resistance is the most sensitive parameter to this conversion, because it probably also is influenced by the static force applied during the measurements. The reactances are about the same above 1 kHz, but deviates at lower frequencies. Impedance data from very soft (the neck) to bony tissue are plotted. In agreement to expectations, the resistance becomes larger with harder tissue and the reactance becomes less negative with more elastic tissue.

The resistance data from Wargowske et al. [1995], Schroeter and Els [1982] and Gierke and Brammer [1995] varies with frequency, and the other's data do not. This is because these data are read directly from a plotted graph, whereas all other data are calculated from the models given in each reference.

Ishizaka et al. [1975] measured the impedance on the neck, the wrist and the cheek, the two latter are not shown in the figure. They derived a model based on data between 40-160 Hz. Hence, it might not be correct to extend their data up to higher frequencies, as it is done in **Figure I.2** and as applied in the model of Williams and Howell, [1990]. The resistance of the wrist corresponds well with the data of the finger, [Gierke and Brammer, 1995]. The reactance agrees less well and is higher than on the finger.

Håkansson's [1986] and Flottorp and Solberg's [1976] data on the mastoid agree very well in the reactance and it has the same shape as the data from Wargowske et al., [1995]. Out

of the two, Håkansson's resistance agree best with Wargowske's resistance between 400-3000 Hz.
The only data that was measured in the ear canal, come from Schroeter and Els [1982] who had a special transducer that covered all the outer 10 mm surface of the ear canal wall. Schroeter and Els measured also the impedance on the mastoid in 4 points around pinna with a disk of 1.75 cm^2. For comparison, their data in front of pinna are also plotted and it is clear that their data are lower than the other mastoid data from the literature. In fact, the resistance looks more like the finger resistance of Gierke and Brammer [1995] and the reactance lies just below 0 from 100-1000 Hz. The data from the 3 other points on the mastoid is in the same order of magnitude and does not change the indication that Schroeter and Els's data are smaller than data from other studies. Even then, their data from the ear canal will be analyzed closer, simply because it is the only available data for the ear canal.

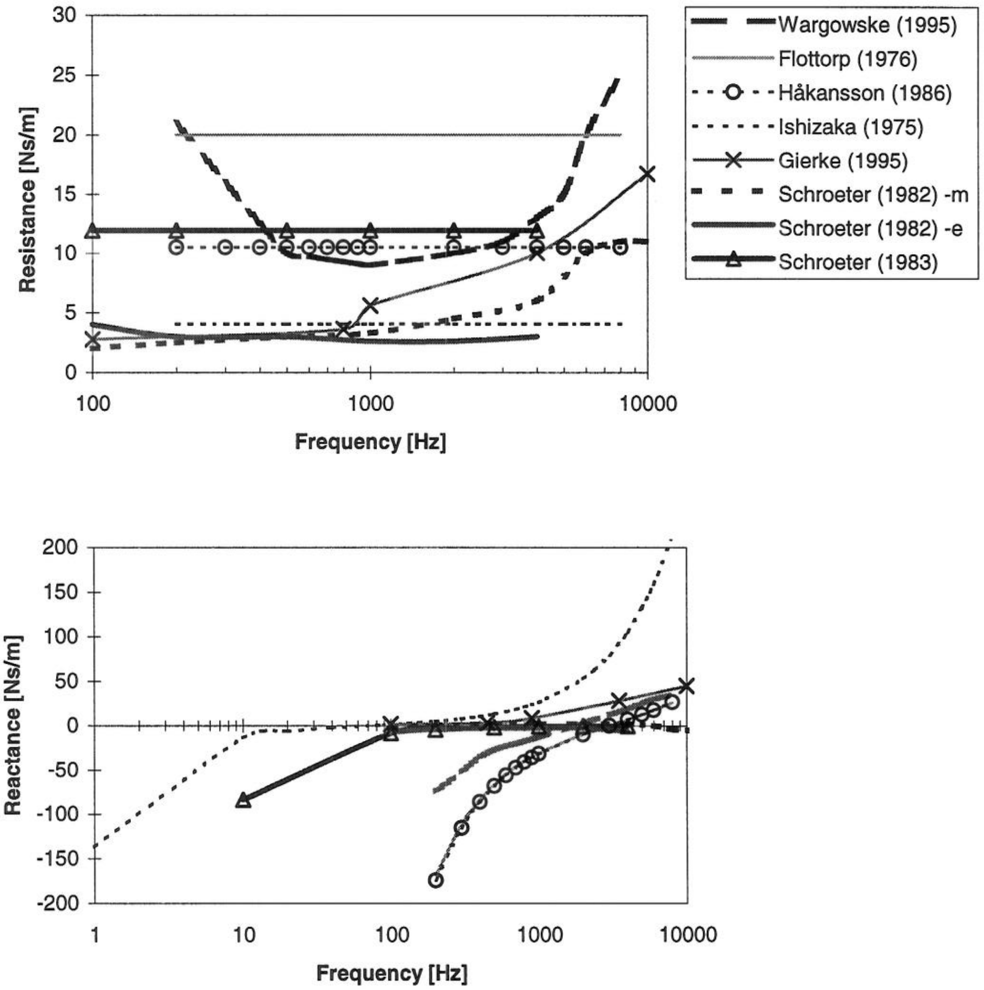


Figure I.2. Mechanical tissue impedance converted to the impedance of a circular disk (1.75 cm^2). Wargowske, Flottorp, Håkansson and Schroeter (1982) - m are measured on the mastoid or around pinna. Ishizaka: on the neck. Gierke: on the finger. Schroeter (1982) -e are measured in the ear canal. Schroeter (1983) is model values of the ear canal tissue. All values are calculated from the models given in the reference, except from Wargowske (1995), Gierke (1995) and Schroeter (1982) where the data are read from plotted graphs.

In **Figure I.2** the measured data in the ear canal from Shcroeter and Els [1982] and the model data used later by Shcroeter [1983] are plotted. The resistance for the model (Schroeter, 1983) is clearly to large compared to the measured data (Shcroeter, 1982 -m).

The measured reactance fits better with the model (enlarged in **Figure I.3**). Schroeter modeled the impedance as a pure compliance and a pure resistance. Thus the mass of the tissue was neglected. In **Figure I.3** Schroeter's model is compared to the measured data in the ear canal. The effect of adding a mass to Schroeter's model is also shown. Håkansson [1986] used for example a mass of $M_s=6\cdot10^{-4}$ Ns^2/m (acoustic units) in a model of the mastoid, but a mass with $M_s = 0.5\cdot10^{-5}$ Ns^2/m fits the measured data best.

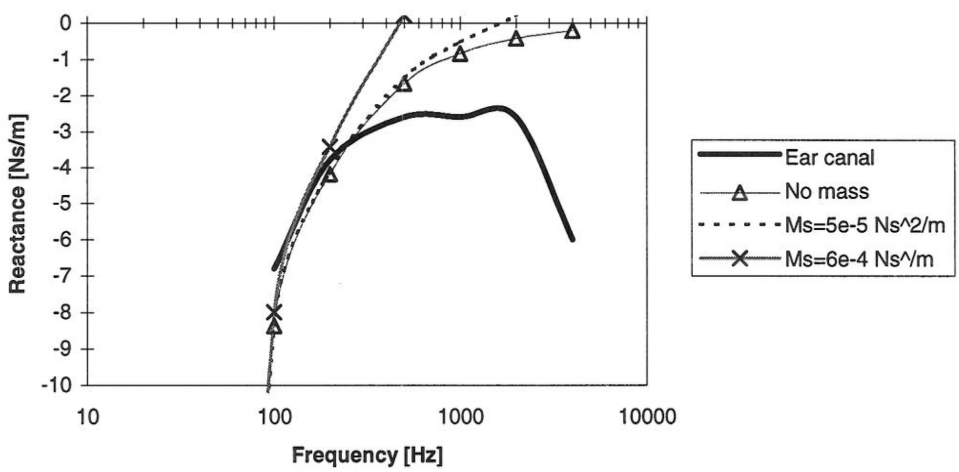


Figure I.3 Enlargement of Schroeter's model (-Δ-) and measured data (solid line) for the ear canal plotted in **Figure I.2**. Ear canal curve is measured on real ears. All other curves are Schroeter's model with varying mass, Ms.