Predicting hearing aid response in real ears

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A hearing aid fitted to different ears will produce very different sound pressure spectra in the ear canal. In addition, this variation in response is different among hearing aids. A description in terms of an electrical analog model of the ear and hearing aid system is given. The applicability of this model is tested through series of measurements. The measurement and prediction procedure was first verified on a coupler (ear simulator) with good results from 300 to 8000 Hz. Three types of hearing aids were then measured and used on five different human ears. Where the measured and predicted response was compared a fairly good agreement was obtained from 300 Hz to approximately 6000 Hz. A major source of error is probe misalignment. The theoretical description given in the present work is likely to be valid from low frequencies to at least 10 kHz. Although solutions for related problems have been given for low frequencies, a solution of the hearing aid fitting problem at high frequencies has not been published earlier. © 1998 Acoustical Society of America. [S0001-4966(98)04106-X]

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INTRODUCTION

The response from a hearing aid may show great variations when used on different ears. The variation may reach 20 dB at high frequencies (Olsson, 1985). This is a major problem when fitting hearing aids. The basis for selecting hearing aids is measurement of the hearing aid response on a coupler (ear simulator), which does not give enough information about the final response on an individual ear. When fitting hearing aids the hearing aid response is compared to the individual hearing loss. The fact that the hearing aid response is not unique makes the hearing aid fitting time consuming.

In order to estimate the variations a study was carried out at three hospitals in Stockholm (Berninger et al., 1992). The results from these tests in the frequency range 500 Hz to 4 kHz with one aid on 16 ears were large variations throughout the entire range. One explanation for these large variations is the fact that the output impedances of various hearing aids are different, and that the input impedances of the human ears also vary between individuals.

Egolf et al. presented a model for predicting response of in situ hearing aids (Egolf et al., 1977). The different parts of the hearing aid are modeled with four-pole parameters which are determined from measurements. The result on an ear simulator was good up to 5000 Hz and on real ears up to 1000 Hz.

In the present work a description in terms of an electrical analog model of the ear and hearing aid system is given. To predict the sound pressure developed in a real canal, two measurement steps are required. First, the electroacoustic behavior of the hearing aid must be characterized in terms of its Thevenin equivalent impedance $Z_H$ and sound pressure $P_H$. Second, the input impedance of the ear canal $Z_E$ must be measured. The sound pressure that will be predicted will then be $P_{EM} = P_H Z_E / (Z_E + Z_H)$ (see Fig. 1). The ear canal input impedance and the hearing aid equivalent impedance
are measured using an impedance probe consisting of an insert earphone and a probe microphone, as shown in Fig. 2. The electroacoustic properties of the probe are characterized via Thevenin equivalents \((P_0, Z_0)\) and determined ahead of time using the Allen procedure (Allen, 1985).

The proposed method will give the predicted sound pressure level at a point slightly outside the mold. All comparisons between measured and predicted sound pressure levels are carried out at this point.

The proposed measurement procedure is expected to be useful when fitting hearing aids. With \(P_H\) and \(Z_H\) from the large amount of hearing aids available stored in a data base, measurement of \(Z_E\) on the patient ear in combination with some calculations will give the predicted sound level. Fitting hearing aids can therefore be carried out by predicting the result without the need of hearing aids at hand.

I. THEORETICAL DESCRIPTION

The aim of the present work was to give a description of the acoustical system “hearing aid and ear.” It should be accurate enough to describe the acoustical parameters involved with sufficient accuracy and yet simple enough for easy applications at the clinics. The frequency range of interest is that of hearing aids today with some margin for further developments. The theoretical model used in this work is shown in Fig. 1. This electrical analogy is a low-frequency model expected to be valid for plane waves (below the first cut-on frequency). However, it may also include the effects of near-field modes. Acoustic impedance defined as sound pressure divided by volume velocity \((1 \text{ acoustic ohm} = 1 \text{ Pa} \cdot \text{s/m}^2)\) is used throughout this paper and the time factor is \(e^{-j\omega t}\).

The model describes the hearing aid (including tube and form-fitted hard plastic mold) as a sound pressure generator and an output impedance (Thevenin parameters). The ear is described with an input acoustic impedance. This model has no geometrical spread. The parameters involved are valid only at the point of connection between the hearing aid and the ear.

In order to determine the impedances for the system “hearing aid and ear” an “impedance probe” is used. This impedance probe is also modeled by an electrical analog circuit as described in Sec. II.

Measurements and calculations for verification of the model have been carried out in the frequency range 100 Hz to 10 kHz. Verification of the model consists of determining the pressure generator and impedances, calculating the sound pressure in the ear canal, and measuring the sound pressure for comparison (see Figs. 12–14).

II. IMPEDANCE MEASUREMENT METHOD

The impedance probe consists of an insert earphone (EAR-3A by Etymotic Research) and a probe microphone (ER-7C by Etymotic Research) (Fig. 2). The earphone is emitting sound into the ear canal through the hard plastic mold and the microphone is measuring the sound pressure at a position remote to the mold. The earphone and microphone together are calibrated and used as an impedance probe. With the Thevenin parameters \((P_0, Z_0)\) of the impedance probe known, the impedance of an object can be calculated from measurement data.

With the assumption that the earphone is giving (approximately) constant volume velocity this configuration of impedance probe has been in use since the 1950s [see “two tube method” in Sanborn (1990)]. In the present work a more nearly complete description of the probe is used. The impedance probe is modeled as an electrical analog with complex-valued Thvenin parameters consisting of a pressure generator \((P_0)\) and an internal impedance \((Z_0)\) as given in Fig. 2. For determining the two Thevenin parameters two
measurements on two known impedances are required. However, the accuracy of such a calibration is limited near antiresonance dips.

A calibration procedure useful for a wider frequency range has been presented by Allen (1985). Allen’s procedure makes use of calibration measurements on four known impedances. This will give an overdetermined system for determination of the Thevenin parameters [solved with a least mean square (LMS) method] leading to a more robust calibration. Keefe et al. (1992) has been using this method with some changes, and their version of it is used in the present work.

The known impedances are in this investigation those of circular uniform brass tubes with rigid ends. Kirchhoff’s solution for sound in a tube with rigid walls includes the effects of both viscosity and heat conduction on acoustic propagation through a rigid cylindrical tube. It assumes that the oscillatory flow is laminar, nonlinear terms in the equation of motion are negligible, and that the inner wall is isothermal. For use in the present work the input impedance and propagation wave number of these tubes are calculated with a high-frequency (small acoustic boundary layer) approximation of Kirchhoff’s exact solution given by Keefe (1984).

With characteristic acoustic impedance $Z_C i$, propagation wave number $\Gamma_i$, and tube length $L_i$, the input acoustic impedance of brass tube No. $i$ is given by (Keefe et al., 1992)

$$Z_i(k) = Z_{ci}(k) \coth(\Gamma_i(k)L_i).$$

When the impedance probe is connected to the impedance to be measured, $Z_x$, the relation between $P_x$ (the sound pressure measured by the microphone), $Z_x$, and the Thevenin parameters of the probe is

$$P_x = \frac{Z_x}{Z_0 + Z_x}.$$

Suppose the calibration is carried out with $M$ known impedances (in the present work $M = 10$). Equation (2) used for each of the $M$ measurements in the calibration gives the system

$$
\begin{bmatrix}
Z_1 & -P_1 \\
Z_2 & -P_2 \\
\vdots & \vdots \\
Z_M & -P_M
\end{bmatrix}
\begin{bmatrix}
P_0 \\
Z_0
\end{bmatrix}
= 
\begin{bmatrix}
P_1Z_1 \\
P_2Z_2 \\
\vdots \\
P_MZ_M
\end{bmatrix}.
$$

Explicit reference to frequency is suppressed in the system equations, and it is understood that they must be solved separately for each frequency. For optimization of this system the error function $\varepsilon(n)$ is first calculated for each frequency $n$ as

$$\varepsilon(n) = \sum_{i=1}^{M} |Z_iP_0 - P_iZ_0 - P_iZ_i|^2.$$  

Whatever values are chosen for the lengths, they must be applied for all frequencies when computing the Thevenin parameters. For visualizing the error, it is convenient to define a nondimensional error function $N(n)$,

$$N(n) = \frac{\varepsilon(n)}{\sum_{i=1}^{M} |P_iZ_i|^2}.$$  

The average normalized error $N_T$ quantifies the error across the optimization bandwidth and is defined to be

$$N_T = \frac{1}{n_2-n_1+1} \sum_{n=n_1}^{n_2} N(n).$$

The error function $N_T$ is a function of the $M$ closed tube lengths, and the lengths $L_i$ are chosen such that $N_T$ is minimized. This may be regarded as a weighted least-squares method where the weighting coefficient is the denominator of Eq. (5).

The minimization method used in the $M$-dimensional space is the modified Powell’s method. In each dimension the minimization technique of Brent is used (Press et al., 1986).

When the optimization is through, $P_0$ and $Z_0$ are given by the LMS solution (Allen, 1985):

$$
\begin{bmatrix}
P_0 \\
Z_0
\end{bmatrix} = \frac{1}{\Delta} \begin{bmatrix}
\sum_{i=1}^{M} |Z_i|^2 & -\sum_{i=1}^{M} Z_iP_i \\
\sum_{i=1}^{M} \bar{P}_iZ_i & -\sum_{i=1}^{M} |Z_i|^2
\end{bmatrix} \begin{bmatrix}
\sum_{i=1}^{M} |P_i|^2P_i \\
\sum_{i=1}^{M} |P_i|^2Z_i
\end{bmatrix},
$$

with

$$\Delta = \left( \sum_{i=1}^{M} |Z_i|^2 \right) \left( \sum_{i=1}^{M} |P_i|^2 \right) - \left( \sum_{i=1}^{M} P_iZ_i \right) \left( \sum_{i=1}^{M} \bar{P}_iP_i \right).$$

The tip of the probe microphone extends 5 mm past the flush surface of the plastic insert (see Fig. 2) in order to reduce the contribution of the evanescent mode coupling between the earphone source and the probe tip. This leaves a cavity between the probe tip and the surface of the earphone source. This cavity will have a volume depending on the tube diameter. The calibration will therefore give different results in the two tube diameters.

With the two sets of Thevenin parameters $P_0$ and $Z_0$ determined by the procedure above, the impedance probe is used without modification with the calibration carried out for diameters 7.5 mm, corresponding approximately to the outer part of the ear canal, and 3 mm, corresponding to the canal in the mold (see Fig. 1). This gives two sets of Thevenin parameters to be used with the two tube diameters.

With the two sets of Thevenin parameters $P_0$ and $Z_0$ determined by the procedure above, the impedance probe is used for measuring input impedance in tubes of diameters 7.5 and 3 mm. This will give $Z_E$ through measurements in the ear canal and $Z_H$ from measurement on the hearing aid through the mold and calculations according to the model in Fig. 2.

III. THE MEASUREMENT PROCEDURE IN BRIEF

In order to check the validity of the model in Fig. 1, $P_H$, $Z_H$, and $Z_E$ were calculated from measurements, the pre-
dicted pressure calculated, and the real pressure measured for comparison. All parameters in the comparison are valid in the point of comparison only.

With the impedance probe calibrated according to Sec. II, the ear input impedance \( Z_E \) is measured according to Fig. 2. \( P_{EM} \) is the measurement variable, \( P_0 \) and \( Z_0 \) are known from the calibration procedure, and \( Z_E \) is given by \( Z_E = Z_0(P_{EM}/(P_0 - P_{EM})) \) according to the model in Fig. 2.

Here \( Z_H \) is measured the same way as \( Z_E \) with the ear replaced by hearing aid plus mold. \( Z_H \) is measured into the mold canal entrance towards the hearing aid.

Also, \( P_H \) was calculated from the response obtained with the hearing aid connected to an IEC 711 coupler. With an electrical analog as in Fig. 1 but with the ear replaced by the coupler, \( P_{EM} \) was measured with the coupler built-in microphone and \( P_H \) calculated from this model \( P_H = P_{EM}(Z_H + Z_C)/Z_C \). This requires knowledge of the coupler input impedance \( Z_C \), which is determined the same way as \( Z_E \) and \( Z_H \).

Here \( P_H \) and \( Z_H \) are measured at the mold end, and \( Z_E \) is measured 5 mm in front of the mold according to Fig. 2. In order to make a correct prediction, \( P_H \) and \( Z_H \) are translated to the remote point of \( Z_E \) measurement. This is carried out with a T matrix formulation according to Sec. 7-7 of Pierce (1981). The ear canal does not reveal constant cross section but for a short distance it is considered sufficiently constant.

After this translation the predicted sound pressure \( P_{EM} \) is calculated according to Fig. 1 (with the exact point of comparison 5 mm from the mold end according to Fig. 2). For comparison the real sound pressure from the hearing aid is then measured at the same point with a probe microphone positioned according to Fig. 2.

This point of measurement is remote from the eardrum and is not directly related to the sensation of hearing. However, the difference in sound pressure along the ear canal is a problem not directly related to the prediction problem addressed by this work.

The measurement system used was Tektronix 2630 spectrum analyzer in swept sine mode. All measurements were carried out at 397 frequency points (100 Hz to 10 kHz in 25-Hz steps). Usually a bandwidth of 100 Hz was used with the tracking filter and an average of five measurements. For the calculation of complex-valued parameters the transfer function is needed. However, the reference in this case is the output signal of the analyzer. In order to get a sound pressure level with \( 20 \times 10^{-6} \) Pa as a reference the autospectrum from the calibrated probe microphone was used, 5 mm from the mold end as shown in Fig. 2. Although not able to detect all measurement errors, the coherence function is a useful parameter. In measurements like those used in the present work it will detect nonlinearities such as amplitude distortion and noise. The minimum coherence requirement was set to be 0.99, giving a maximum error in level of 0.1 dB for each measurement (Bendat and Piersol, 1986). This requirement was nearly always met.

IV. PROBLEMS AND LIMITATIONS OF THE METHOD

Different sources of error may distort the results. Those expected to dominate were checked and are discussed below.

A. Stapedius contraction

At high sound pressure levels the stapedius muscle is contracted. The muscle will pull the ossicle chain in such a way that the eardrum is pulled inwards and becomes stiff. This will in turn increase the input impedance of the middle ear. For sinusoidal stimulus the threshold of contraction is 90–95 dB SPL (Margolis, 1993). The stapedius reflex is bilateral, that is, a sufficiently high sound level into one ear will cause a contraction in both ears. It has been shown that the sound pressure level in the ear canal is increased approximately 2.5 dB when the stimulus tone was increased from 85 to 110 dB hearing level in the contralateral ear (Anderson, 1969).

The stapedius reflex is active throughout long-term exposure to an industrial noise which is variable with respect to frequency and amplitude (Borg et al., 1979). In the present work the stimulus signal was a sine wave stepped in frequency over the measurement range. It has not been found in the literature whether or not a change in frequency only will reactivate the stapedius muscle. If the stapedius was not reactivated from a frequency shift only, it may have caused problems turning on and off during our measurements.

In order to check the reactivation from frequency shifts, measurements were therefore carried out on two persons with normal audiograms. The stimulus was emitted into the left ear, from the generator via amplifier, attenuators, and earphone. The stimulus level was monitored through a probe microphone and a measurement amplifier. On the right side the same type of equipment for monitoring the change in impedance was connected. The impedance itself was not measured, but the change in sound pressure due to the change in impedance was.

In the first experiment the stimulus tone was 500 Hz, 110 dB SPL for 10 s. The probe tone was 800 Hz, 65 dB. Over the 10-s period the probe tone level decreased approximately 0.5 dB. This indicates that the stapedius is contracted at the onset of the stimulus tone and is released over the measurement time.

In the second experiment the stimulus tone was 300 Hz to 1 kHz in 25-Hz steps, 3 s for each frequency, 110 dB SPL. The frequency step and time are those of most measurements in the main investigations in this project. The probe tone was the same as in the first experiment. In this case there was no significant change in the level of the probe tone. This indicates that the stapedius is reactivated by the change in frequency only. Although not found in the literature, this result was expected.

B. Probe location in the ear canal

While making measurements in the ear canal it is important to know the position of the probe end. As will be shown below, it is important to measure the input impedance of the ear and the sound pressure (for comparison) at the same
This problem was to a large degree solved by making identical molds for the hearing aid and impedance probe on each ear.

A simulation model using MathCAD was made for checking the dependence of probe position in the hearing aid Thevenin impedance measurement on the predicted sound pressure level. The hearing aid receiver is modeled as a resistor $10^9$ acoustic ohms, and the middle ear is modeled with Shaw’s electrical analog (Shaw and Stinson, 1981). The tube (diameter: 2 mm, length: 60 mm) and ear canal (diameter: 7 mm, length: 15 mm) are modeled as loss-free straight tubes of constant cross section. An impedance translation procedure, Chap. 3-7 in Pierce (1981), was used for sound propagation in the tubes.

In Fig. 3 the real part [Fig. 3(a)] and the imaginary part [Fig. 3(b)] of the hearing aid Thevenin impedance are given. The left-most curves are valid for the correct position of the impedance probe. The right-most curves are valid for a probe position 1 mm closer to the hearing aid. In Fig. 3(c) the differences in predicted sound pressure level in the ear canal of these two cases are given.

The corresponding error caused by an impedance probe misalignment in the ear canal is given in Fig. 4. The same situation as in Fig. 3 is assumed. The difference in predicted sound pressure level between a correct measurement and a measurement with the impedance probe 1 mm closer to the ear is shown.

All measurements on the ear were carried out with hard plastic molds for probe mounting and tightening of the ear canal. The age of the mold may have an impact on the results of this project. The ear canal of the hearing aid user will change in size and shape over time. Therefore a fresh cast giving a mold that fits tightly is essential.

C. Near-field modes

Propagating higher modes in the human ear canal are not present below the first cut-on frequency. This is expected to be approximately 18 kHz for adults (Rabbitt, 1988). For children this frequency limit is even higher. Higher modes in the sense of near-field modes (primarily near the ear canal entrance and the eardrum) may be found above 2.5 kHz (Rabinowitz, 1981). According to Hudde (1989), the near-field modes may exist as low as $1/10$ of the first cut-on frequency but there is no distinct limit.

The step in cross-sectional area between the sound canal of the mold and the ear canal will also produce near-field modes. These modes are commonly described as a mass loading of the mold canal or as an extension of the mold canal. From an impedance point of view this can easily be taken into account (Karal, 1953). Measurements in these near fields will, however, give unpredictable results and, therefore, measurements are taken some distance from these area steps.

V. COMPARISON WITH CALCULATED TUBE IMPEDANCE

In order to evaluate the errors in the impedance measurements, a comparison between measured and calculated input impedance of two tubes was carried out. Tube dimensions not used in the calibration measurements were used.
Measured and calculated tube impedance is given in Fig. 5 for a small diameter tube ($\varnothing = 3$ mm, length = 403 mm). At low frequencies the difference is small and increasing toward high frequencies, and it is large in the range 7.5–9.5 kHz. In this range the EAR-3A earphone gives a low output level. The coherence is more than 0.999 throughout the frequency range for the tube measurement and all calibration measurements involved. However, the earphone response makes the autospectrum fall from approximately 3 kHz and is 30 dB down at 8 kHz. The damping in the tube is also increasing with frequency.

In Fig. 6 measured and calculated impedance is given for a large diameter tube ($\varnothing = 7.5$ mm, length = 380 mm). The damping is less in this tube and the result is better. The phase plot does reveal sharp peaks at 4.5 and 9 kHz in addition to other minor errors. These peaks may be due to the fact that some of the calibration tubes have impedance dips very close to the tube length determined by the calibration program.

VI. PREDICTION ON A COUPLER

The measurement and prediction procedure was first carried out with the IEC 711 coupler (ear simulator) used instead of the human ear. In order to check the reproducibility four measurements of each parameter were carried out. The Philips M49 hearing aid was used.

Figures 7 and 8 show measured and predicted sound pressure levels in coupler for best and worst result, respectively. From 230 Hz to 8 kHz the error is within ±5 dB. Above this frequency range large errors occurs.

VII. PREDICTION ON REAL EARS

For verification of the model on real ears three behind-the-ear hearing aids and five ears where used. The following hearing aids where used:

(i) Widex ES1,
(ii) Philips M49,
(iii) Phonak Pico SC.

A. Hearing aid Thevenin impedance

The Thevenin impedance of the Widex ES1 hearing aids is given in Fig. 9 for three measurements. The coherence is very good in these measurements except for the very lowest

![Image](https://example.com/image.png)
frequencies. The Thevenin impedance is dependent not only on the receiver, tube, and mold, but also on the output impedance of the hearing aid amplifier.

B. Hearing aid Thevenin pressure

These measurements were carried out with the same hearing aid input signal as in the prediction check in Sec. VII D.

The result is depending not only on the same parameters as the Thevenin impedance, but also on the signal processing of the hearing aid and the microphone characteristics if used.

The coherence function was in excess of 0.999 in these measurements, except in the frequency range where the hearing aid output level is low. The transfer function for hearing aid+coupler, Xfer\(P_H\), representing \(P_H\) as translated to the end of the mold (see Sec. III), is given for the Widex ES1 hearing aid in Fig. 10.

C. Ear impedance

All real ears used in the main investigations of this project are pathological. Results from the impedance audiometry gives an idea of whether or not the ear input impedance is normal. Of the five real ears used, only one subject (A) has a documented normal result from the impedance audiometry.

Due to the stapedius contraction the human ear is expected to have an input impedance varying with sound level. Impedance measurements on the ear were therefore carried out at two different sound pressure levels. SPL(max) denotes the maximum sound pressure level in one measurement. In Fig. 11 mean, max, and min from three measurements at 100–105 dB SPL(max) are given for subject E. The measurements at 80–85 dB SPL(max) show very little difference from these results.

In some of the results the reactance differs substantially from the expected cavity shape. This occurs for the high- and low-level measurements in case E. The reason for this is likely to be leakage, although Vaseline was used in an attempt to get an airtight seal. In the frequency range 4000–7000 Hz the result is very different between measurements. The six impedance measurements on each ear were carried out in pairs, one measurement at each level without remounting the probe.
The coherence function from these measurements exceeds 0.99 except for the frequency range 7–9 kHz, where it sometimes was as low as 0.7.

**D. Prediction check**

In Figs. 12–14 examples of measured and predicted (from measured parameters in Fig. 1) response are given for the three hearing aids on subject A. The measurements were carried out over a wider frequency range than the hearing aids are expected to reproduce. Noise problems were experienced at the low levels in the high-frequency dips.

As all parameters above were calculated from three different measurements, the prediction check was also carried out three times. The error “predicted sound pressure level minus measured sound pressure level” is given for all hearing aids on subject E in Figs. 15–17. Maximum, minimum, and average (in dB values) out of three measurements and calculations of each parameter are given. Subjects D and E show better reproducibility than the others; this may be due to molds of a later date than in the other cases. For these two subjects a good mold fit was noted.

In general, there is a low-frequency region and a high-frequency region with large spread. A mid-frequency region, 1–6 kHz, usually reveals better accuracy (approximately ±5 dB in most cases). The frequency range 3–4 kHz is very important when fitting hearing aids. The error in this range is in general within ±3 dB.

Average (in dB values), maximum, and minimum were also calculated for each hearing aid over all subjects. Results are given in Fig. 18 for Widex ES1, in Fig. 19 for Philips M49, and in Fig. 20 for Phonak Pico SC.

The hearing aids are expected to have Thevenin parameters independent of sound pressure level as mentioned earlier. The human ear is expected to have an input impedance varying with sound pressure level. The prediction check was therefore carried out in two cases, with ear impedance measured at 80–85 dB(max) and 100–105 dB(max). The difference between averages over all hearing aids and subjects measured at 100–105 dB(max) and 80–85 dB(max) are given in Fig. 21. There is no significant difference between

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**FIG. 12.** Measured (solid) and predicted (dashed) sound pressure level from Widex ES1 on subject A.

**FIG. 13.** Measured (solid) and predicted (dashed) sound pressure level from Philips M49 on subject A.

**FIG. 14.** Measured (solid) and predicted (dashed) sound pressure level from Phonak Pico SC on subject A.
the results at the two levels. The peak at 8.4 kHz is due to measurement errors.

VIII. DISCUSSION AND CONCLUSIONS

An electrical analog model (Fig. 1) has been given as a theoretical description of the acoustical system “hearing aid and ear.” This model has been tested in the frequency range 100 Hz to 10 kHz.

The comparison between predicted and measured sound pressure level on coupler is clearly showing the influence of probe misalignment in Fig. 8. The error is on the order of 4 dB up to 8 kHz. Apart from this type of error the result is good from 300 Hz up to 8 kHz. Large errors occur at 8–9 kHz. These are due to calibration errors in this frequency range of the impedance probe at small tube diameters (see Fig. 5) and low hearing aid response at these frequencies.

The prediction check on real ears shows larger errors as expected. Below 250 Hz large errors occur in all measurements. A common problem at low frequencies (<1 kHz) is leakage. This seems to be present for some subjects but not all. A good mold fit, and thus a tight mold, was noted on subjects D and E.

In the frequency range 1–6 kHz the average predicted sound pressure level is usually within approximately ±5 dB of the measured level. The coherence functions from the $Z_H$ measurements are exceeding 0.999. In the $\text{Xfer}_{PH}$ measurements it exceeds 0.999, except for the frequency range where the hearing aid output is low (7–9 kHz). The coherence functions from $Z_E$ measurements exceeded 0.999, except for frequencies above 7 kHz where it was sometimes as low as 0.7. The reproducibility of $Z_H$ (Fig. 9), $\text{Xfer}_{PH}$ (Fig. 10), and $Z_E$ (Fig. 11) is not excellent. However, the influence in detail of each parameter on the prediction error is not always easily detected, since all parameters involved are varying between different prediction results. An alternative way of comparison would be to study the variation of one parameter at a time with the other parameters fixed. Since the fixed parameters would be estimates of the real ones, the results of such a comparison would be uncertain. The comparison used in the present work simulates the results obtained in a measurement situation at a clinic.

The large variations in the frequency range 6–9 kHz are caused partly by the low sound pressure levels in the $\text{Xfer}_{PH}$, $Z_E$ and $P_{EM}$ measurements. This causes problems since the noise level of the probe microphone is rather high (55 dB SPL equivalent, 20 Hz to 20 kHz bandwidth). The variations are caused partly by the error in impedance probe calibration mentioned in Sec. V and shown in Fig. 5. It is also important to keep in mind that all measurements, including prediction check, also include errors from probe misalignment. In Figs. 18–20 the prediction errors for all subjects on each hearing aid are given. In general the reproducibility is bad at high frequencies but is somewhat better on subjects D and E. As these two subjects revealed a
better mold fit than the others, this indicates the importance of a good mold fit and thus a more nearly accurate probe alignment.

The overall shape of the averaged prediction error curves, negative at low frequencies and positive at high frequencies, may be caused by probe misalignments. However, this would mean that this type of error is systematic. The example in Fig. 14 shows dips at different frequencies for measured and predicted result. This error is probably caused by probe misalignment. These errors are simulated for the case of $Z_H$ and $Z_E$ measurements as described in Sec. IV B and shown in Figs. 3 and 4.

Measurements of $Z_E$ and $P_{EM}$ reveal errors in the frequency range 7–9 kHz. This is due to the frequency response of the insert earphone and the hearing aids. The tube earphone model ER-3A has the response of the TDH-39 supra-aural earphone. This means low levels at high frequencies. The EAR-3A was chosen for its high-level capabilities. As the sound pressure level used is not critical in the prediction of hearing aid response, insert earphones for use at lower levels may be used. In that case earphone models with a more nearly flat frequency response are available.

A main reason for the bad result at high frequencies is the low hearing aid response. The hearing aids used in this project were chosen from their “deviation in response on different ears” (Svārd, 1995) as this is of primary importance. The development in hearing aid technology is directed towards better high-frequency response. The measurement and calculation procedure given in this report is thus expected to give a better result as development goes on.

Summing up sources of errors, we have the following.

(i) Calibration errors, shown as deviations from expected results in Figs. 5 and 6. In the present work the calibration tube lengths were measured with a rule. The acoustical measurement of calibration tube lengths in Keefe et al. (1992) should be used.

(ii) Response of EAR-3A earphone. Measurements can be carried out at lower sound pressure levels. This means that an earphone with a more flat response can be used.

(iii) Noise level of ER-7C probe microphone. The noise from the probe microphone causes errors in the frequency ranges were the sound pressure level is low.

(iv) Response of hearing aids. At high frequencies the output level is low. In combination with a rather high microphone noise this causes problem.

(v) Probe misalignments. Misalignments cause errors increasing with frequency. Molds made from not too old individual casts of the ear canal are necessary.

To conclude, the described measurement and calculation procedure appears to be useful between approximately 300 Hz and 6 kHz. In this frequency range probe positioning errors dominate. At higher frequencies large errors occur, mainly due to the response of the hearing aids and insert earphone used.

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