Observations on the Relations among Occlusion Effect, Compliance, and Vent Size

Roberto Carle*  
Soren Laugesen†  
Claus Nielsen‡

Abstract

In a clinical experiment, it was found that there is a high correlation between the compliance measured by tympanometry and the minimum size of the earmold vent, which just solves the client's occlusion problem related to his/her own voice when using a hearing aid. For ears with sensorineural hearing losses, compliance explained 59 percent of the variation in vent size, whereas the average low-frequency hearing loss explained as little as 0.3 percent. In a laboratory experiment, the objective occlusion effect measured with the participants' own voices showed a similar relationship with compliance. Whereas the former relationship between compliance and vent size may be explained by a simple model, the latter relationship turns out to be the opposite of what a first-order model predicts. Hence, compliance must be indicative of another aspect of the occlusion mechanism, which has a more profound influence on the observed occlusion effect than compliance itself.

Key Words: Body conduction, compliance, occlusion, own voice, soft tissue, tympanometry, vent

Abbreviations: BTE = behind the ear, CIC = completely in the canal, ITE = in the ear, LF = low frequency, OV = own voice, SPL = sound pressure level

Sumario

Se encontró en un experimento clínico que existe una alta correlación entre la compliancia medida por timpanometría y el tamaño mínimo del orificio de ventilación (vent) en el molde auditivo, que resuelve en el cliente el problema de oclusión en cuanto a la percepción de su propia voz, cuando utiliza un auxiliar auditivo. Para oídos con hipoacusias sensorineurales, la compliancia explicó el 59% de las variaciones en el tamaño del orificio (vent), mientras que la hipoacusia promedio para frecuencias graves sólo explicó un 0.3% de éstas. En un experimento de laboratorio el efecto objetivo de oclusión medido con las propias voces de los sujetos mostró una relación similar con la compliancia. Mientras que la primera relación entre compliancia y tamaño del orificio (vent) puede ser explicada por un modelo simple, la segunda relación mostró resultados opuestos a lo que predice un modelo de primer orden. Por lo tanto, se concluye que la compliancia debe ser indicativa de otro aspecto del mecanismo de oclusión, el cual tiene una influencia más profunda que la propia compliancia en el efecto de oclusión observado.

Palabras Clave: Compliancia, conducción corporal, oclusión, propia voz, tejido blando, timpanometría, vent

Abreviaturas: BTE = retroauricular, CIC = intracanal, HTL = nivel auditivo umbral, ITE = intraauricular, LF = baja frecuencia, OV = voz propia, SPL = nivel de presión sonora

One of the most common complaints of hearing aid users is that their own voice (OV) sounds unnatural: boomy, hollow, or echoing. Poor sound quality of OV is also one of the top 10 reasons why some hearing aids end up in the drawer (Kochkin, 2000). This problem with OV is very often caused by the so-called occlusion effect, which occurs because the body-conducted contribution to a person's perception of OV is trapped in the cavity between the occluding earmold of the hearing instrument and the tympanic membrane. The result is a build-up of sound pressure at low frequencies (LFs) that may be as much as 30 dB relative to the open-ear condition.

*Audiovox, Livorno, Italy; †Oticon Research Research Centre, Eriksholm, Snekkersten, Denmark.  
Reprint requests: Soren Laugesen, Oticon Research Centre, Eriksholm, Kongevejen 243, 3070 Snekkersten, Denmark
(Wimmer, 1986; Mueller et al, 1996; Hansen, 1997). Typically, the occlusion effect has a flat maximum between 80 and 500 Hz and vanishes above 1 kHz. In the open-ear condition and at the LFs considered here, the body-conducted contribution is insignificant compared with the air-conducted contribution (Hansen and Stinson, 1998).

In today’s hearing aid dispensing there are basically three ways to address the client’s eventual occlusion problem with OV. First, the earmold (or in-the-ear [ITE] hearing aid) may be equipped with a vent, through which the body-conducted part of OV can dissipate (Kampe and Wynne, 1996; Mueller et al, 1996). Second, inspired by the original work of Zwislocki (1953), it has been shown that completely-in-the-canal (CIC) instruments that are fitted with a seal in the bony part of the ear canal can solve or at least reduce the occlusion problem in many cases (Killion et al, 1988; Garcia, 1994; Mueller, 1994; Garcia and Staab, 1995). Unfortunately, bony sealed CIC hearing aids have earned a bad reputation for introducing physical discomfort and are hence rarely dispensed (Garcia and Staab, 1995; Powers and Mueller, 1995; Sweetow and Valla, 1997). Third, occlusion problems may be dealt with by counseling, along the lines of “You’ll get used to it!” (Pogash and Williams, 2001). In this article, we consider only the first option, venting.

As pointed out by Pogash and Williams (2001), it can be advantageous to address occlusion problems up front; hence, it is of interest to predict when and to what extent occlusion will be a problem in a hearing aid fitting. A straightforward reasoning is that occlusion problems with OV occur when the body-conducted contribution dominates the hearing aid-conducted contribution of OV. Since the level of the hearing aid-conducted contribution depends on the insertion gain of the hearing instrument, which is usually derived from the hearing loss, a dependency on the hearing loss at low frequencies is indicated. This leads to the expectation that a small LF hearing loss implies a big occlusion problem and vice versa. Also, this line of reasoning suggests that the smaller the LF hearing loss, the larger the vent that will be required to solve the occlusion problem. This view is reflected in the vent recommendation that often is part of hearing aid manufacturers’ fitting guides (Cox and Risberg, 1986; Courtois et al, 1988; Skinner, 1988). However, the vent recommendation is not based on hearing loss alone. This is because the allowable vent size may be limited by either feedback considerations (if much high-frequency gain is required) or by space limitations (mostly in small ITE instruments). It is not surprising that occlusion problems occur if a diminished vent size has been chosen (e.g., to avoid feedback). However, clinical practice shows that even when the vent size has not been diminished, a considerable number of fittings are still troubled by occlusion problems. This indicates that vent prescriptions derived only from LF hearing loss sometimes are inappropriate.

The aim of this article is to suggest and substantiate that another measure in the standard audiologic test battery, namely compliance as determined by tympanometry (Margolis and Shanks, 1985), has considerable potential for improving on the existing method of vent prescription with respect to occlusion problems. This proposition stems from a clinical observation made by the first author from a client who exhibited both an extreme compliance value and an extremely severe occlusion problem. The concept has been studied in two independent experiments, the results of which are presented and discussed below.

**EXPERIMENT 1**

The first study was carried out between January and July 2000 at the first author’s hearing aid dispensing facility in Italy. The key part is the determination of the minimum acceptable vent size with respect to the perception of OV.

**Material**

The test subjects who participated in this experiment were ordinary hearing aid customers who came to the first author’s dispensing facility for treatment. The relevant parts of that fitting protocol and the equipment used are reviewed.

During the indicated period of time, 82 clients (32 males, 50 females) with ages ranging from 33 to 92 years participated in the study. No specific selection criterion was used, except that the subjects confirmed that they were willing to participate. Of these 82 clients, 33 were first-time users and 49 were experienced users. With 37 subjects being bilateral hearing aid users, a total of 119 ears were included. The hearing losses ranged from mild to profound. All ears...
were examined by otoscopy using a handheld Heine Mini 2000 Otoscope, and the status of the tympanic membrane was noted. Tympanometry was performed on all ears using an Interacoustics AT22 Impedance Audiometer (226-Hz probe tone), the tympanogram was plotted, and the static compliance value noted. Standard audiometric air-conduction and masked unoccluded bone-conduction hearing thresholds were determined with an Interacoustics AC33 Audiometer. Finally, ear impressions were taken after making sure that no earwax was present in the ear canals. A cotton block was placed just beyond the second bend of the ear canal before the impressions were taken.

The hearing aids fitted to the 119 ears were 96 behind the ear (BTE), 14 ITE, 7 CIC, and 2 spectacles with hearing aids, mostly nonlinear two-channel compression aids. The ITE and CIC aids were made to terminate at the second bend of the ear canal. For each of these, the maximum possible vent diameter was requested—up to 2.3 mm. This allowed the use of the Select-A-Tube vent selection system, which is a suite of flexible 8-mm-long tubes that can be inserted into the vent hole of the hearing aid and also into each other to create different vent diameters. For the BTE aids and the spectacles, canal lock-type earmolds were ordered. This type of earmold is a canal mold with a small wing in the concha. All earmolds terminated around the second bend of the ear canal to ensure a proper sound bore direction (Andre, 1993; Jester, 1993). The earmolds were made without venting, and the sound bore was placed in a way that allowed subsequent drilling of vents by the experimenter. With the exception of a few soft earmolds, all earmolds were made in hard acrylic material.

**Method**

Each client paid three visits to the clinic. At the first visit, otoscopy, tympanometry, audiometry, and ear impression taking were carried out. At the second visit, the hearing aids were fitted and the minimum acceptable vent size was determined (the procedure for this is described in detail below). Finally, at the third follow-up visit, fine-tuning of the hearing aid and verification of vent size were performed. Consecutive visits were separated by 2 or 3 weeks.

The determination of the minimum acceptable vent size with respect to occlusion and the sound quality of OV was done as follows. During the initial fitting of the hearing aid, the vent was closed, and the client was asked to describe his/her OV. This evaluation was done with the hearing aid switched on. The client was encouraged to use the following four categories: (1) louder than normal, (2) with more bass, (3) with more treble, or (4) with an echo, like talking in a barrel or in a large empty room. Any of the first three answers was interpreted as an indication of an amplification problem, which was subsequently addressed by fine-tuning of the hearing aid. If the client responded with category 4, it was interpreted as an occlusion problem, and a small vent was introduced. In such cases, the vent size was increased until the complaint disappeared. To verify the response from the client, the vent size was then reduced until the complaint of occlusion emerged again. The vent size was changed systematically with the objective of finding the smallest acceptable vent size with regard to OV occlusion problems. The final vent configuration was then recorded. The whole session of hearing aid fitting and determination of vent size typically took 1 hour. No prior counseling on OV problems was done unless the client brought up the subject himself. In those few cases, the prior discussion of OV problems was kept to a minimum. For the bilateral hearing aid users, the vent size determination was performed on one ear at the time (the non-test ear unoccluded), followed by a final evaluation with both hearing aids in place. Thus, the eventual OV occlusion problem was solved for all clients, except for two clients who reported a small and acceptable residual problem. In some cases, the introduction of a large vent required that the hearing aid’s gain at high frequencies be reduced to avoid feedback. At the end of the fitting, three clients reported minor remaining problems with feedback to the extent that howling occurred when a telephone handset was placed close to the ear. These problems were not further addressed.

For the BTE earmolds, increasing the vent was done by drilling successively larger vent holes. Reducing the vent was done by means of the Select-A-Tube system. For the ITE and CIC hearing aids, all vent variations were accomplished with the Select-A-Tube system. In both cases, the Select-A-Tube vent plugs were applied only from the outside.

If necessary, the hearing aids were fine-tuned after the vent size determination to compensate for the loss of LF amplification, which accompanies the introduction of a vent.
Data Preparation

To facilitate the forthcoming analyses, each vent used in the study has been transformed into an equivalent uniform parallel vent with a common length, equal to the median vent length of $l_q = 19$ mm. Thus, each vent may be characterized by its equivalent diameter $d_{eq}$ alone. The transformation has been done according to the assumption that the acoustic properties of the vent at LF frequencies are determined by the resonance frequency of the Helmholtz resonator formed by the vent and the cavity between the earmold and the tympanic membrane (Studebaker and Zachman, 1970). Ignoring end corrections (which are very small with the present range of vent dimensions), this resonance frequency is according to Kinsler and Frey (1950),

$$f_0 = \frac{c}{2\pi} \sqrt{\frac{S}{V}} = \frac{cd}{4\sqrt{\pi V}}, \quad (1)$$

where $c$ is the speed of sound; $S = \frac{1}{2} \pi d^2$, $l$ and $d$ denote the physical area, length, and diameter of the vent, respectively; and $V$ is the cavity volume. Now, substituting $l$ and $d$ with $l_q$ and $d_{eq}$ and assuming that $V$ remains the same, keeping $f_0$ constant yields

$$\frac{cd}{4\sqrt{\pi V}} = \frac{cd_{eq}}{4\sqrt{\pi V}} \Rightarrow d_{eq} = d \sqrt{\frac{l_q}{l}}. \quad (2)$$

This simple equation applies to a uniform parallel vent. For the (very few) cases for which the final vent was plugged from the outside to form a vent consisting of two steps, a similar analysis (Studebaker and Zachman, 1970) yields

$$d_{eq} = d_{SAT} \sqrt{\frac{l_q}{l_{SAT} + (d_{SAT}/d_2)^2}}, \quad (3)$$

where $d_{SAT}$ and $l_{SAT}$ relate to the Select-A-Tube and $d_2$ and $l_2$ to the remaining wider portion of the vent.

Model

Furthermore, equation 1 may be used to formulate models for the data. The relations between minimum acceptable vent size and compliance, as well as average LF hearing loss (mean of the hearing threshold level values at 250 and 500 Hz), are considered. A key observation is that the volume ($V$) in equation 1 comprises both the physical volume ($V_{ph}$) of the cavity between the earmold and the tympanic membrane and the equivalent volume corresponding to the compliance ($V_c$), that is, $V = V_{ph} + V_c$. Thus, rearranging equation 1 yields

$$S_{eq} = \frac{4\pi f_0^2}{c^2} (V_{ph} + V_c), \quad (4)$$

where $S_{eq} = \frac{1}{2} \pi d^2$. Let us assume that all test subjects are aiming for the same relief from occlusion, in terms of the same resonance frequency of the Helmholtz resonator ($f_0$). Then equation 4 clearly suggests a linear relation between the equivalent vent area ($S_{eq}$) and compliance ($V_c$). Second, it is assumed that for constant volume ($V$), the required relief from occlusion ($f_0$) is related to the LF hearing loss along the lines discussed in the introduction. Owing to the lack of a model for the latter relation, a linear relation is proposed between $d_{eq}$ and average LF hearing loss (cf., the proportionality between $d$ and $f_0$ in equation 1).

Results

The data have been stratified into two groups. The first comprises 81 ears with a pure sensorineural hearing loss and is denoted the sensorineural group. The criterion for a pure sensorineural hearing loss was an air-bone gap of less than 10 dB at all frequencies found by inspection of the audiogram. The second group is denoted the mixed loss group and comprises 38 ears with either abnormal tympanic membranes or poor middle ear function as judged from otoscopy and tympanometry. Also, ears showing an air-bone gap larger than 10 dB at any frequency are included in the mixed loss group. Below, the data are presented and analyzed separately for each group.

Sensorineural Group

The main result is shown in Figure 1, in which $S_{eq}$ is plotted as a function of compliance. In one instance, the maximum reading of the impedance audiometer of 2.5 mL was exceeded, as indicated in Figure 1. The results shown in Figure 1 indicate that there is a strong relationship between compliance and the area of the minimum acceptable vent, with large compliance being associated with the need for a large vent. From Figure 1, it seems that the data at $V_c = 2.5$ mL are relatively spread out. This is actually not unexpected. In the following section, it is argued that $d_{eq}$ is determined with a constant standard deviation. Hence, the standard deviation of $S_{eq} = \frac{1}{2} \pi d^2$ increases with larger values of $S_{eq}$, which occur at large values of $V_c$. 


Figure 1  Plot of the minimum acceptable equivalent vent size (area, $S_{eq}$) as a function of compliance for 81 ears with sensorineural hearing losses. Data from experienced users are indicated by $\Delta$, data from first-time users are indicated by $\nabla$. One compliance observation was clearly truncated by the maximum reading of 2.5 ml of the impedance audiometer; this data point is indicated by a filled triangle. The result of a linear regression is indicated by a solid line (correlation coefficient $r = .77$).

With this in mind, the assumption of a linear model seems justified, and, accordingly, the regression trend line has been added to the figure.

Second, Figure 2 shows $d_{eq}$ as a function of the average LF hearing loss. Contrary to the prediction of the second model proposed above, Figure 2 indicates little if any relation between the LF hearing loss and the minimum acceptable vent size. These observations are confirmed by the results of the statistical analyses, which are summarized in Table 1. The appropriate F tests show that although the proposed linear relation between compliance and $S_{eq}$ is indeed very highly significant, there is no correlation between LF hearing loss and $d_{eq}$ in these data.

It is often argued that first-time hearing aid users perceive occlusion problems with OV differently from experienced users, but it has not been possible to find any substantial evidence for this supposition in the literature. The data presented here do not indicate any difference between experienced and first-time users with respect to the size of vent required to solve their occlusion problem with OV. This can be seen from visual inspection of Figures 1 and 2 and has been confirmed by a statistical analysis of the data from the first-time hearing aid users alone. These latter results were essentially the same as those above and hence are not shown.

Mixed Loss Group

In this case, the relation between compliance and $S_{eq}$ is shown in Figure 3, together with the trend line from the sensorineural group (see Fig. 1). Again, one compliance value exceeded the maximum reading. From Figure 3, it is clear that there are two subgroups in this mixed loss group. One subgroup closely follows the trend from the sensorineural group, whereas ears in the other subgroup were in no need of a vent at all. A plot of $d_{eq}$ as a function of average LF hearing

| Table 1  Summary Results of the Regression Analyses on the Sensorineural Group |
|-----------------|-----------------|-----------------|-----------------|
| Independent Variable | Dependent Variable | $R^2$, Variance Explained (%) | $p$, Reject Hypothesis |
| Compliance, $\psi$ | $S_{eq}$ | 59 | < .001 |
| Average low-frequency hearing threshold level | $d_{eq}$ | 0.3 | .6 |

Data from both experienced and first-time users are included.
A repeated analysis with only the first-time hearing aid users in this mixed loss group again suggests that their results are very similar to those of the experienced users. This is also confirmed by visual inspection of Figure 3.

Discussion

It may well be argued that these data are subject to bias. Clearly, the present experiment was by no means blind. The experimenter (the first author) obviously knew what the experiment was about, and since tympanometry was done prior to the hearing aid fitting, a bias to the result of the vent size determination was possible. However, there are two aspects that promote confidence in the data: (1) the systematic way in which the vent size determination was carried out (described above) and (2) the fact that the experiment took place in a commercial setting helps to remove bias. This is because installing a vent that was unnecessarily large often would lead to lack of LF gain and feedback problems (or unacceptable gain reductions at high frequencies), whereas a vent that was too small would lead to occlusion problems. Either way, the client would be dissatisfied with the hearing aid. In a more usual scientific experiment, that may be acceptable for the duration of the test, but not so in the present case, in which the test subjects paid for the hearing aids.

Concentrating first on the sensorineural group, it was found that compliance explained 59 percent of the variance in vent area, which leaves a residual variance of 41 percent. Several factors contribute to this residual variance, as the following breakdown will show. First, the compliance measurement is subject to uncertainty, partly because the compliance value stated by the impedance audiometer also depends slightly on the resistance and inductance components, but mainly because of stochastic measurement inaccuracy. Regarding the latter, Gaihede (1996) found that the standard deviation of the difference between two compliance measurements was 0.079 mL. Thus, the

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<th>R², Variance Explained (%)</th>
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<td>Compliance, $V_e$</td>
<td>$S_{eq}$</td>
<td>32</td>
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<td>Average low-frequency hearing threshold level</td>
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standard deviation on a single compliance measurement is \( \sigma_x = \frac{1}{\sqrt{2}} \cdot 0.079 \) mL = 0.056 mL. Second, the vent size determination is subject to an intrinsic variance because a discrete set of diameters was available and because deliberate length adjustments were not used. Basically, vents were introduced in steps of 0.5 mm, which means that the standard deviation of the vent diameter is \( \sigma_{d_{w}} = \sqrt{12} \cdot 0.5 = 0.14 \) mm (Leijon, 1992). To determine how much of the total variance in the data is accounted for by the individual variances on \( V \) and \( d_{w} \), a Monte Carlo simulation was performed. A synthetic data set in perfect agreement with equation 4 and the regression line in Figure 1 was constructed. All \( V \) and \( d_{w} \) values in this synthetic data set were then perturbed by adding noise terms according to the standard deviations \( \sigma_{V} \) and \( \sigma_{d_{w}} \). This perturbed data set was then entered into the regression analysis, which showed that the perturbed compliance values are able to explain 92 percent of the variance in the perturbed \( S_{w} \) values. Thus, it is estimated that the individual variances on \( V \) and \( d_{w} \) explain 100% - 92% = 8% of the total variance in the real data. This leaves a residual variance of 41% - 8% = 33%, which is attributable to model imperfections. Three specific model imperfections are discussed below.

First, the compliance values that have been used above are determined at equilibrium static pressure—also denoted peak pressure, according to Margolis and Shanks (1985). However, a middle ear pressure of ±25 daPa is perfectly normal (Arlinger, 1991), which means that the effective compliance value at ambient pressure may be different—and always smaller—than that measured by the tympanometry. This indicates both a bias and a further stochastic variation to the present compliance data. Second, it has been assumed that the physical volume (\( V_{ph} \)) in equation 4 is constant. Clearly, this quantity varies between the test subjects, and this variation accounts for some of the observed residual variance. Third, it has been assumed (for Fig. 1 at least) that in the vent size determination, all test subjects are aiming at the same vent response, in terms of the Helmholtz resonance frequency \( (f_{h}) \) in equation 4. However, it is expected that the target \( f_{h} \) varies between subjects. As discussed above, it was predicted that \( f_{h} \) would be related to the LF hearing loss, but no evidence for such a relation was found in the data (see Fig. 2). In this respect, it should be noted that the expected relation actually is between minimum vent size and LF insertion gain rather than LF hearing loss, as stated in the introduction. Unfortunately, it has not been possible to collect information on (real-ear) insertion gains for this analysis, but it is expected that some of the residual variance could have been explained by LF insertion gain, as an alternative to LF hearing loss. Finally, some of the variation in target \( f_{h} \) can be referred to the variation in the subjects' personal thresholds of acceptance of occlusion. Such variability has previously been reported by Kampe and Wynne (1996) and Sweetow and Valla (1997).

**EXPERIMENT 2**

The second experiment was carried out at the research facility in Denmark, where the second and third authors are employed. The study comprised measurements of compliance and objective measurements of the occlusion effect with the test subjects’ OVs. The latter measurements comprised recording of ear canal sound pressure level (SPL) spectra in three conditions: open ear (baseline for the determination of the occlusion effect) and two occluded conditions. The occlusion effect measurements were originally made for another purpose (Nielsen and Laugesen, 2000), which explains why CIC hearing aid shells were used for the occluded conditions (see below).

**Material and Method**

Seven individuals participated in the study. The condition of the participants’ ear canals and tympanic membranes was examined by video-otoscopy. Tympanometry was performed with an Interacoustics AZ26 Impedance Audiometer in a standard clinical procedure (the instrument was used in its automatic mode), and the compliance value was noted. No indications of abnormal middle ear function were observed. All participants except one had ear impressions taken bilaterally, which were deep enough that at least 2 mm of the bony part of the ear canal were included around the entire circumference of the impression. Based on these impressions, two CIC hearing aid shells were made in standard hard acrylic material for each ear. The two shells were identical in length and insertion depth and differed only in tip configuration. One was configured to create a complete seal in the bony part of the ear canal; on the other, the bony seal was removed by tapering the tip of the CIC shell. The tapering was done only to the part of the shell that extended into the
bony portion of the ear canal and was kept to an absolute minimum. All shells were fitted with a 0.8-mm pressure relief vent, which is so small that it has very little influence on the measured occlusion effect. Finally, each CIC shell was equipped with a round miniature hearing aid microphone facing into the cavity between the hearing aid shell and the tympanic membrane. The microphones were mechanically connected to the CIC shells by means of short pieces of silicone tube. For the open ear measurement, a single common microphone was prepared only with silicone tube, litze wires, and a connector for signal and power supply. During the measurement, the microphone was simply inserted into the ear canal and held in place by securing the litze wires to the head of the subject by adhesive tape. (Note that insertion depth is relatively unimportant in the frequency range considered here.) Finally, a second common microphone was embedded in a standard yellow EAR 3A foam plug with the tube removed and used in the nontest ear of the test subject as a measurement reference. Each microphone used in the study was calibrated relative to a Bruel & Kjaer 4134 condenser microphone so that results may be presented as dB SPL.

For the actual measurements, the subjects were seated in the anechoic chamber and given a text from which they were instructed to read aloud for the duration of each measurement (a few minutes). All measurements were performed with a Bruel & Kjaer 2032 Dual Channel Frequency Analyzer, from which data were transferred to a computer and stored for further analysis. In each measurement condition, the reference microphone with the foam plug was connected to channel A of the Bruel & Kjaer 2032 and the microphone in the test ear was connected to channel B. Thus, transfer functions between the reference microphone and the test microphone were determined for each of the three conditions: open ear, tapered CIC occluded, and bony seal CIC occluded. Each transfer function was determined as an average over at least 200 recorded spectra. The use of transfer functions relative to a common reference alleviates the need for a constant voice level between the three measurement conditions and ensures that the three results for each ear can be presented very accurately relative to each other. However, because of the potential variation in voice level, the exact positioning on the dB SPL scale was done according to the average SPL at the nontest ear across the three measurement conditions.

Data Preparation

First, a single illustrative example result of the occlusion effect measurements is shown in Figure 4 with SPL spectra obtained from one ear summed into 1/3-octave band values for the three measurement conditions. From Figure 4, it is seen that the build-up of sound pressure in the cavity is substantial in both occluded cases. In agreement with Zwislocki (1953), Killion et al (1988), Mueller (1994), Garcia (1994), and Garcia and Staab (1995), a considerably smaller occlusion effect is observed with the bony sealed CIC device than with the tapered CIC device. Because of the way in which the two CIC shells were produced, the cavity volume is almost exactly equal in the two conditions, and it must be concluded that the observed difference between the two occlusion effects is caused by the bony seal alone. This confirms the generally accepted notion that the main contributor to the body-conducted sound that can be observed in front of the tympanic membrane is vibration of the soft tissue in the outer part of the ear canal. While the soft tissue is free to generate sound into the cavity with the tapered CIC device, these vibrations are greatly reduced with a bony sealed CIC device. In the latter case, the remaining body-conducted sound, which is apparent in Figure 4 as excess SPL above the open ear case, is thought to be composed of several other components generated by different physical mechanisms (Tonndorf, 1972).
For each of the 13 test ears and for each occluded condition, a broad-band occlusion effect has been calculated as follows. First, the occluded 1/3-octave band SPL values were normalized by the open ear SPL to yield the occlusion effect. Second, an average of the linear magnitude of the occlusion effect was performed across the 140- to 700-Hz frequency band, in which data were valid for all ears and all conditions according to the measured coherences between the test and reference signals. Outside this frequency range, low coherence indicated unreliable data, either because of a lack of excitation signal or because of incoherent signals not stemming from OV (e.g., sounds produced by the mandibular joints). In any case, by synthesizing each occlusion effect into a single value, it is possible to plot the relation between the measured compliance and the broad-band occlusion effect on the same ear.

Results

The occlusion effects measured with the tapered CICs are plotted in Figure 5. Here it is seen that there is a pronounced relationship between the two, with large compliance values being associated with large occlusion effects. A regression analysis shows that the correlation coefficient is $r = .59$, and the appropriate F test shows that the hypothesis of a linear relation between compliance and occlusion effect will be accepted on a 5 percent significance level ($p = .04$). Second, the occlusion effects measured with the bony sealed CICs can be viewed in a similar way. The graphic representation is shown in Figure 6, which—apart from the expected lower occlusion effect values—indicates a somewhat weaker relationship between compliance and broad-band occlusion effect in this case. This observation is confirmed by the regression analysis, which shows that the correlation coefficient is $r = .41$. This correlation is not significant according to the F test ($p = .16$).

Discussion

Although both compliance values and occlusion effects measured with the tapered CICs are considered to be trustworthy, it can be argued that the data obtained with the—nominally—bony sealed CICs are somewhat questionable. This is because it is difficult to guarantee that a bony seal was, in fact, obtained, owing to the well-known difficulties in impression taking and shell production for bony sealed CICs (Bentler, 1994; Hosford-Dunn, 1996; Klein, 1996). However, since the tapered and the bony sealed CIC shells were otherwise identical, any difference between the two measured occlusion effects must be attributable to the presence or absence of the bony seal, as already mentioned above. On 11 of all 13 ears under test, inspection reveals differences of more than 5 dB between the two occlusion effects across a frequency range of at
least an octave. This substantially exceeds the test-retest variance, which in a pilot study was found to be in the order of 1 to 2 dB. A repeat of the regression analysis on these 11 ears yielded a correlation coefficient of 0.50, which is somewhat higher than before but, according to the F test, still not significant (p = .12). Hence, the observed difference in correlation between compliance and occlusion effect found with either tapered or bony sealed test CICs seems to be reliable, although still somewhat speculative because of the low number of observations. This result is further discussed in the following section. The two potentially nonbony seal data points are indicated separately in Figure 6.

**GENERAL DISCUSSION**

In experiment 1, the simple Helmholtz resonator model for the effectiveness of a given vent as a function of compliance (V) was strongly confirmed by the results from the sensorineural group. The compliance can easily be the dominant term in the total volume V = Vp + Vv from equation 4, particularly when the physical volume of the cavity between the earmold and the tympanic membrane (Vp) is small because of fairly deep earmolds, as used in this study. Thus, a large compliance value should require a large vent. Based on a personal communication by Scheller (1989, cited in Staab and Finlay, 1991), a similar reasoning has previously been sketched by Staab and Finlay (1991), although these authors only considered the effect of variations in Vp. In studies of the effect of compliance on the sound produced by a hearing aid receiver—rather than OV—a similar trend was also found in simulations by Kates (1988) and in coupler and real-ear measurements by Gilman et al (1981).

Although the considerations of vent effectiveness alone seem to explain the relation between compliance and vent size shown in experiment 1, the results from experiment 2 actually indicate a further relation between compliance and vent size. This is attributable to the observed relation between compliance and the magnitude of the objective occlusion effect, together with the assumption that a greater objective occlusion effect requires a larger vent to solve the OV occlusion problem. As for experiment 1, a model that can explain the observation from experiment 2 is sought. The first and simplest model that can be used is one commonly used in the literature on CIC hearing aids (Staab, 1996). In that approach, the ear canal volume is described as a hard-walled cavity, in which the volume comprises both the physical volume and the volume equivalent to the compliance, as in equation 4. In the present experiment, the occluding CIC shells were made with 0.8-mm pressure relief vents; hence, the cavity model is a reasonable assumption above about 200 Hz and up to about 6 kHz, where the wavelength becomes comparable to the dimensions of the cavity. The body-conducted sound from OV is modeled as a piston with constant volume displacement Xbody into the ear canal cavity. In that case, the sound pressure in the cavity is according to Kinsler and Frey (1950)

\[ p = \frac{\rho c^2}{V} X_{\text{body}} = \frac{\rho c^2}{V_p + V_v} X_{\text{body}}, \]  

where \( \rho \) is the density of air. From equation 5, it is seen that this simple model predicts that the sound pressure in the occluded condition becomes smaller if compliance (V) is large. Because of the extremely low source impedance in the open ear condition, the open ear sound pressure caused by OV is virtually independent of compliance (Gilman et al, 1981). Hence, taking the ratio between the open ear and the occluded sound pressures, the occlusion effect will be small when compliance is large. This obviously contradicts the data from experiment 2.

Instead of simply discarding the first-order model in equation 5, the validity of this model has been verified in several ways. First, the first-order model was confirmed by comparing it with a more advanced numeric model (Hansen, 1998) by changing only the parameter representing the compliance of the tympanic membrane in the advanced model. Second, the second and third authors carried out informal pilot experiments in which the occlusion effect was measured with both normal middle ear pressure and offset middle ear pressure introduced by the Valsalva and Toynbee maneuvers. These experiments showed increasing occlusion effect when the tympanic membrane was stiffened by the pressure offset, also in agreement with the first-order model. Finally, the tympanogram itself reflects that for a constant volume displacement source, the sound pressure has a minimum when the tympanic membrane is in its most flexible equilibrium position (Northern, 1980). This is also in accordance with the first-order model. All in all, the first-order model is confirmed to the extent that if all other things are equal, a small compliance will produce a large occlusion effect and vice versa. Since the data from experiment 2 show the opposite rela-
It must be concluded that all other things are not equal and that compliance must be indicative of something else, which has a greater influence on the occlusion effect than compliance itself. In the following, an explanation for this is tentatively proposed.

Although compliance usually is associated with the tympanic membrane, the tympanic membrane does not, in fact, make a major contribution to the total compliance in a normal middle ear. This is easily seen from the fact that disruption of the ossicular chain results in an extremely high compliance value (Jerger and Hayes, 1980). To a first approximation, the ossicular chain is a mere transformer, which does not contribute to the total compliance either. Thus, compliance is mainly determined by the stiffnesses of the round and oval windows of the cochlea. This can also be realized from the fact that in otosclerotic ears, in which the structure of the footplate on stapes in the oval window and the structure of the membranes in cochlea are affected by pathologic bony growth, very small compliance values are observed, although the tympanic membrane and the ossicular chain are normal (Breitlau, 1973; Shanks, 1984). A final observation is that in early embryonic life, the ear canal, tympanic membrane, middle ear, and cochlea develop from the same kind of tissue, which, at around the fourth to fifth week of gestation, separates and then changes into different structures, such as bone, cartilage, and skin (Wright, 1997). The relevance of this observation will become clear below.

With respect to the occlusion effect, it is generally accepted (Zwislocki, 1953; Garcia, 1994; Mueller, 1994; Garcia and Staab, 1995) that the soft tissue is responsible for the majority of the body-conducted contribution to sound of OV in the occluded situation. This also explains the effect of a bony seal, as seen in Figure 4. Considering the aforementioned familiarity between the different anatomic elements of the ear, it is hypothesized that compliance is indicative of the flexibility of the soft tissue around the ear canal. Assuming further that more flexible soft tissue will transmit more body-conducted sound from OV, this could explain why a large compliance is associated with a large occlusion effect, even though increasing compliance itself will tend to reduce the occlusion effect as predicted by the simple model above.

It is interesting that the present hypothesis also lends an explanation to the poorer correlation between compliance and occlusion effect observed for the bony sealed test CICs in experiment 2. For the bony sealed occluded situation, most of the soft tissue contribution to the sound of OV has been eliminated. The remaining sound pressure is then composed of several other components, which are not related to the soft tissue and compliance. Thus, the observed lower correlation in the bony sealed CIC condition is actually in agreement with the proposed explanation.

**Conductive and Mixed Hearing Losses**

In experiment 1, the main result is that when there is a considerable conductive component to the hearing loss, occlusion is not a problem, and no vent is required. This seems to be generally accepted knowledge in clinical practice, probably derived from the fact that bone-conduction thresholds are unaffected by an occluding earphone when there is a conductive component to the hearing loss (Dirks, 1985). A more thorough explanation for this conception can be assembled based on the material presented above.

First, with a large conductive hearing loss, the main paths for perception of OV will be directly into the cochlea via bone, tissue, and fluid (Tonndorf, 1972; Porschmann, 2000; Sohmer et al, 2000) for which normal hearing sensitivity applies. This will certainly be true in the open ear case but possibly also in the occluded case, even if the body-conducted outer ear component has been amplified substantially because of occlusion. Thus, if the occlusion effect (the amplification of the outer ear component of OV) is inaudible, a vent is obviously not required.

Second, even if the occlusion effect is audible to the subject, the interaction with the amplified sound from the hearing aid is likely to be different when there is a conductive component to the hearing loss. With a sensorineural hearing loss and a modern compression hearing aid, the insertion gain applied to OV is likely to be relatively small since OV is a rather strong signal because of the proximity of the source to the hearing aid's microphone. As stated in the introduction, a small insertion gain at LF is expected to lead to occlusion problems because in that case, the body-conducted OV component is more likely to overpower the hearing aid-conducted component. Consequently, a relatively large vent is needed. With a conductive hearing loss, substantial gain is required, even at high input levels, which means that the sound from the hearing aid is more likely to mask the occluded body-conducted outer ear component.
This also suggests that less or no vent is required for a hearing loss with a significant conductive component.

Finally, a conductive hearing loss is often accompanied by a very small compliance, which has been shown to increase the effectiveness of a given vent.

A Note on Clinical Use

It is worth noting that although an impedance audiometer is a fairly standard piece of audiologic equipment, such a device is usually used for diagnostic purposes based simply on the shape of the compliance curve and the static pressure at which maximum compliance occurs (Jerger and Hayes, 1980; Margolis and Shanks, 1985). Thus, the actual numeric values of compliance and residual volume are rarely used; hence, there has so far been very little incentive to perform accurate calibration of these impedance audiometers. Furthermore, Gaihede (1996) has shown that the compliance value stated by an impedance audiometer depends quite dramatically on the sweep speed employed in the measurement.

Owing to these potential sources of error, it would be inappropriate to propose a prescriptive vent size rule for clinical use based on the material presented above. This has accordingly not been the purpose of this article, which merely has intended to present and discuss the observed relations.

CONCLUSIONS

In today’s hearing aid fittings, the prescription of vent with regard to occlusion and sound quality of OV considers only the perceptual dimension, which suggests that a small hearing loss at LFs should result in a large vent and vice versa. In the present study, it has been shown that there are two other factors that also have a strong influence on the minimal vent size that will solve the occlusion problem in a given hearing aid fitting.

Considering first the purely physical side of things, it is well known that the magnitude of the occlusion effect varies considerably among individuals. The results presented here indicate that some of this variation can be predicted from the compliance, as it is determined in a standard tympanometry. This relation is to the effect that a large compliance value foretells a large occlusion effect, which further calls for a relatively large vent. Since a simple first-order model predicts the opposite relation, an alternative explanation has been proposed. It is hypothesized that compliance is indicative of the sound transmission characteristics of the soft tissue, which is responsible for the majority of the body-conducted sound from one’s OV.

Second, concerning rehabilitation, it has been shown that the effectiveness of a given vent depends on compliance, such that with a large compliance the vent diameter needs to be greater to obtain relief from occlusion. A simple model predicts a linear relation between compliance and vent area, and the experimental data obtained on ears with a pure sensorineural hearing loss confirmed this relation very strongly. Even when there was a limited conductive component to the hearing loss, the relation remained intact between compliance and the minimum vent size that just solved the subject’s occlusion problem. For ears with a substantial conductive component to the hearing loss, occlusion was not a problem, and no vent was required.

REFERENCES


