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Effects of ear-canal pressurization on middle-ear bone- and air-conduction responses

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Abstract

In extremely loud noise environments, it is important to not only protect one's hearing against noise transmitted through the air-conduction (AC) pathway, but also through the bone-conduction (BC) pathways. Much of the energy transmitted through the BC pathways is concentrated in the mid-frequency range around 1.5–2 kHz, which is likely due to the structural resonance of the middle ear. One potential approach for mitigating this mid-frequency BC noise transmission is to introduce a positive or negative static pressure in the ear canal, which is known to reduce BC as well as AC hearing sensitivity. In the present study, middle-ear ossicular velocities at the umbo and stapes were measured using human cadaver temporal bones in response to both BC and AC excitations, while static air pressures of ± 400 mm H₂O were applied in the ear canal. For the maximum negative pressure of -400 mm H₂O, mean BC stapes-velocity reductions of about 5–8 dB were observed in the frequency range from 0.8 to 2.5 kHz, with a peak reduction of $8.6(\pm 4.7)$ dB at 1.6 kHz. Finite-element analysis indicates that the peak BC-response reduction tends to be in the mid-frequency range because the middle-ear BC resonance, which is typically around 1.5–2 kHz, is suppressed by the pressure-induced stiffening of the middle-ear structure. The measured data also show that the BC responses are reduced more for negative static pressures than for positive static pressures. This may be attributable to a difference in the distribution of the stiffening among the middle-ear components depending on the polarity of the static pressure. The characteristics of the BC-response reductions are found to be largely consistent with the available psychoacoustic data, and are therefore indicative of the relative importance of the middle-ear mechanism in BC hearing.

Keywords

Bone conduction; Hearing protection; Middle-ear response; Middle-ear dynamics; Ear-canal air pressure; Auditory biomechanics

1. Introduction

Conventional hearing-protection devices (HPDs), such as earplugs and earmuffs, reduce the risk of hearing damage by reducing the noise transmitted through air conduction (AC). AC is the normal mode of hearing in which the inner ear (cochlea) is excited through the eardrum vibrations that result from acoustic pressure in the ear canal. However, the maximum level of hearing protection provided by a conventional HPD is normally limited by bone-conduction (BC) sound transmission, where the sound energy is transmitted to the cochlea through acoustically-induced skull-bone vibrations that bypass the ear canals occluded by an HPD. The limitation on hearing-protection performance imposed by this BC sound transmission is commonly referred to as the “BC limit”. The effects of the BC sound transmission are normally of little concern in most noise environments, but they become critically important in an extremely loud environment such as on the flight deck of an aircraft carrier, where the noise can reach levels as high as 140–150 dB SPL (sound pressure level). In such extreme environments, even wearing an HPD that can attenuate noise down to the BC limit may not be sufficient, since the BC-transmitted sound may still be loud enough to cause permanent hearing loss.

The most noticeable feature of the HPD BC limit is a peak feature in the mid-frequency range centered around 1.5–2 kHz (Zwislocki, 1957; Berger et al., 2003; Reinfeldt et al., 2007), which indicates prominent BC sound transmission in this frequency range. Assuming that the noise has a broadband spectrum, such as the exhaust noise of an aircraft jet engine, and also assuming that one is wearing an HPD that provides noise attenuation down to the BC-limit level, the overall BC sound spectrum will then be dominated by the sound energy associated with this mid-frequency peak around 1.5–2 kHz. Therefore, one needs to improve the HPD attenuation especially in this mid-frequency range in order to substantially improve the HPD performance.

BC sound transmissions are said to be primarily comprised of the following three components: (a) “compressional”, (b) “inertial-ossicular”, and (c) “external-canal” (Silman and Silverman, 1991; Stenfelt et al., 2002). The compressional BC component results when skull vibrations are transmitted directly to the cochlea via distortional vibrations of the bone surrounding the cochlea. The inertial-ossicular BC component results when the BC excitation is transmitted to the cochlea through middle-ear ossicular vibrations. This is called the “*inertial-ossicular*” component since it is the mass inertia of the middle-ear structures that introduce relative motions of the ossicles with respect to the vibrating base bone structure, which in turn excite the cochlea through stapes vibrations. The external-canal BC component results when the BC-induced vibrations of the ear canal, primarily the cartilaginous part, produce acoustic pressures that excite the tympanic membrane (TM).

There have been indications that the prominent mid-frequency BC limit originates from the inertial-ossicular BC component (Carhart, 1971; Tonndorf, 1972; Linstrom et al., 2001). Although this component is typically called “inertial-ossicular”, a more precise term may be the “ossicular-resonance” mechanism, since it likely involves a middle-ear structural-resonance phenomenon which is not just the result of the ossicular-mass inertia. Recent

studies have shown that the middle-ear system resonates on average at around 1.5–2 kHz in response to BC excitations (Stenfelt et al., 2002; Homma et al., 2009).

In light of the evidence for the middle-ear origin of the mid-frequency BC-limit feature, one potential approach for BC-sound mitigation is to apply static air pressure in the ear canal, which is known to reduce the hearing sensitivity through a reduction in the middle-ear mobility. Previous psychoacoustic studies have indicated that such ear-canal pressurization not only reduces hearing sensitivity for AC, but also for BC (Humes, 1979; Aazh et al., 2005). Humes (1979) further observed that there is a prominent peak reduction in BC hearing at 2 kHz, which he attributed to the loss of the middle-ear BC contribution.

In this study, human temporal-bone measurements were performed to observe the effects of ear-canal pressurization on the middle-ear BC as well as AC responses. Previous temporal-bone measurements have obtained such data for AC responses (Murakami et al., 1997; Gan et al., 2006), but not for BC responses. A finite-element (FE) model of a human middle ear was also used to gain further insight into the dynamics of the BC- and AC-response alterations due to ear-canal pressurization. An improved understanding of the mechanisms of BC-hearing suppression due to ear-canal pressurization is a critical step toward the potential future utilization of this phenomenon for hearing protection in extremely noisy environments.

2. Methods and materials

2.1. Temporal-bone measurements

The effects of ear-canal pressurization on middle-ear dynamic responses to both BC and AC excitations were measured with human cadaver temporal bones. The experimental setup is largely based on the setup described in Homma et al. (2009), except for additional provisions to control the ear-canal static air pressure and to measure stapes velocity. The experimental setup is illustrated in Fig. 1.

2.1.1. Temporal-bone preparation—Temporal bones were extracted from human cadavers using a Schuknecht bone saw at the time of autopsy. The TM and the middle ear were inspected in each bone using an operating microscope; bones with abnormal TMs or middle ears were excluded from the investigation. A total of eight temporal bones were used in this study. Five of them were used in the experiment within 48 h of death; these were labeled as TB1 (78-year-old male, right ear), TB2 (68-year-old male, right ear), TB3 (68-year-old male, left ear), TB4 (73-year-old male, left ear), and TB5 (86-year-old male, right ear). The remaining three temporal bones, TB6, TB7, and TB8, were previously frozen for one month before being thawed and used in the experiment. No subject information on these three bones was available. For all preparations, the attached connective tissue was removed and the bony wall of the external ear canal was drilled down to 2 mm from the TM annulus. A 25-mm-long plastic tube with an internal diameter of 8.5 mm was placed against the bony ear-canal remnant; it was placed so that the axis of the tube was approximately perpendicular to the plane of the TM annulus. The plastic tube was held in place with epoxy in order to minimize air leaks around the plastic tube. In addition, a transparent plastic lid was attached to the tube opening using beeswax and glue in order to maintain static

pressures in the tube, and also to acoustically isolate the tube canal from background noise. Preliminary measurements of AC and BC middle-ear responses with and without this transparent lid did not show significant differences, which indicated that the potential loading of the TM due to the closed air volume in the tube was minimal. For stapes-velocity measurements, a simple mastoidectomy and posterior hypotympanotomy were performed using a surgical drill. Removal of the mastoid portion of the facial nerve and surrounding bone was also performed in order to obtain a good view of the stapes footplate. The average diameter of the individual stapes access holes was about 3 mm. The access hole was covered with a transparent glass plate.

2.1.2. AC- and BC-excitation methods—Each temporal-bone specimen was encased in plaster, and then attached to a shaker (B&K type 4810) in order to simulate BC excitations. The shaker was rigidly connected to the bone by a screw held in place with cement. The shaker was attached in such a way that the vibration axis was approximately perpendicular to the plane of the TM annulus. This particular BC-excitation axis was chosen to be consistent with two previous investigations (Stenfelt et al., 2002; Homma et al., 2009). The previous temporal-bone data by Stenfelt et al. (2002) also indicated that the ossicular BC vibration may be most sensitive to the vibration given in this direction, although the differences in BC-vibration responses appear to be rather small for BC excitations given in different directions. For the AC excitation, a hearing aid receiver (Knowles, #2955) was used to introduce acoustic pressure to vibrate the TM. The peak sound pressure level for the AC stimulation was 90–93 dB, which is well below the 120–130 dB SPL range where nonlinear distortions are expected to occur (Voss et al., 2000). The middle-ear vibration levels produced by the BC excitation were comparable to those produced by the AC excitation, which ensured that the middle-ear structures were not overdriven during the measurements.

2.1.3. AC- and BC-response measurements—A laser-vibrometer sensor head (Polytec, HLV-1000) was used to measure velocity responses of the middle ear. In order to achieve good reflection of the laser light, retro-reflective beads were positioned at measurement locations. These micro beads were negligibly small in size (about 5–10 μm in diameter and 1 μg in mass), and thus introduced negligible mass loading. All measurement data were acquired by using a PC-based data acquisition system (SYSid, Berkeley, CA). A probe-tube microphone (Etymotic Research, ER-7C, Elk Grove Village, IL) was also installed to measure the acoustic pressure in the ear-canal tube. The tip of the probe tube was positioned at a distance of about 2 mm from the TM.

AC responses were expressed as the velocity obtained at each measurement point normalized by the ear-canal acoustic pressure, v/p_{ec} . The main velocities of interest in this study were the velocities measured at the umbo, v_{umbo} , and at the stapes footplate, v_{stps} . The velocity at the lateral process (LP) of the malleus handle, v_{lp} , was also measured in order to examine the relative motions between the umbo and the LP, which are indicative of the 2D rigid-body motions of the malleus handle (Homma et al., 2009). Fig. 1(b) shows the locations for the umbo and the LP measurements. The velocities at these two locations were

measured by aiming the laser through the ear-canal tube such that the measurement angle of the laser was approximately perpendicular to the TM plane and in line with the shaker axis.

Stapes velocity was measured at the approximate center location on the stapes footplate. The angle of the laser with respect to a line perpendicular to the stapes footplate was estimated to be around 40–60°. No corrections were made to the stapes-velocity data to compensate for the angle of the laser, since the present study was primarily concerned with the differences in the response magnitudes between different cases, and not with the absolute magnitudes.

BC responses were measured at the same locations as the AC responses, but with the excitation introduced by the shaker. Since the whole bone structure encasing the middle ear was also vibrating in this case, the middle-ear velocities had to be expressed in terms of velocities relative to the base-bone velocity introduced by the shaker, v_b . The velocities for BC are therefore expressed as the differential velocity Δv , calculated by subtracting v_b from the measured velocity, v , normalized by v_b (Stenfelt et al., 2002; Homma et al., 2009):

$$\Delta v/v_b = (v - v_b)/v_b \quad (1)$$

It should be noted that the differential velocity, Δv , for the AC case is equivalent to the absolute velocity, v , since the base velocity is equal to zero in that case. As indicated in Fig. 1, the base velocity measurements for the umbo and the LP differential velocities were made at a point on a bony part of the ear canal, in the same direction as the umbo-velocity measurement. For the stapes, the base velocity was obtained at the cochlear promontory, in the same direction as the stapes-velocity measurement.

The frequency ranges of the measured data discussed in this study are 0.4–4 kHz for the AC responses and 0.7–4 kHz for the BC responses. The BC responses below 0.7 kHz have been omitted from the discussion. This is because the BC-response sensitivity tends to become progressively lower with decreasing frequency, which makes the low-frequency BC response data susceptible to noise. As will be shown shortly, the ossicular structure starts to have appreciable differential velocity in response to BC excitations only around and above its resonance frequency at 1.5–2 kHz. Below the resonance frequency, the BC response shows a rather steep roll-off in magnitude as the frequency decreases. BC responses at low frequencies are also likely to be more susceptible to measurement errors than AC responses, due to the need to compute the differences of two measured velocities (ossicular and base velocities). Since the two velocities are similar in magnitude and phase at low frequencies, the differences between them become smaller with frequency, and therefore more susceptible to noise and measurement errors.

2.1.4. Ear-canal pressurization—A miniature air pump (Schwarzer, SP135 FZ) was used to control the ear-canal static pressure. Due to a practical difficulty of ensuring a perfectly airtight seal around the ear-canal tubing, the air pump had to be continuously active during a measurement in order to maintain the desired level of air pressure in the ear-canal tube. This was found to introduce a periodic pump noise that could be picked up by the probe-tube microphone. This extraneous noise was attenuated by a silencer, which was installed in line with the pump as shown in Fig. 1. The silencer was made of a PVC pipe

section (1.6 cm in diameter, 5 cm in length), and had two internal chambers which were inter-connected through a thin flexible tube (about 0.5 mm in diameter, 7 cm in length). The use of this silencer sufficiently reduced the pump noise such that there was no significant evidence of this noise in the measured data.

The pressure levels used in this study were +400, +200, 0, -200, and -400 mm H₂O. A U-tube manometer was used for monitoring the ear-canal static pressure. A positive pressure means that the ear-canal pressure is higher than the ambient pressure, and a negative pressure means that the ear-canal pressure is lower than the ambient pressure. The maximum static pressure that was employed for the present study was limited to ± 400 mm H₂O. Achieving and maintaining a pressure level higher than this was not possible for the current study due to the gradual development of air leaks in some temporal bones, which demanded higher performance from the pump. It is suspected that the small air leaks might have been introduced through potential micro channels through the bone near the TM due to bone drilling to fit the plastic tube.

It was also observed that, for some temporal bones, the baseline responses (i.e. zero-pressure cases) showed some variation depending on the polarity of the ear-canal pressure that had been applied just prior to the baseline measurement. Therefore, prior to each response measurement with a static pressure, a pre-conditioning procedure was performed in which the TM was momentarily pressurized (about 100 mm H₂O in magnitude) in the polarity of interest. For the discussions that follow, baseline zero-pressure cases are designated as “+0 mm H₂O” and “-0 mm H₂O” to, respectively, indicate that the positive or negative pressure pre-conditioning step has been taken.

2.2. Finite-element analysis

A finite-element (FE) model of a human middle ear was also used to analyze the response characteristics observed in the temporal-bone data. Fig. 2 shows the human middle-ear FE model.

2.2.1. Changes from the previous model—The present model is for the most part identical to the middle-ear model developed previously by Homma et al. (2009). Component geometries of the three middle-ear ossicles and the TM have been derived from micro-CT-scanned geometry data from a human temporal bone. The FE-model mesh was created by using HYPER-MESH (Altair Engineering, Troy, MI), a FE pre-/post-processing program, and the simulation was performed by using ACTRAN (Free Field Technologies, Belgium), a vibro-acoustic FE solver.

A few adjustments have been made to the FE model for the present study. The superior ligament of the malleus has been added to the model, which was motivated by anatomical observations (Sim and Puria, 2008) and the realization during the present study that this ligament may play an important role in the stiffening of the middle ear. The material of the malleus-incus joint was made more flexible, which is more realistic (Willi et al., 2002; Nakajima et al., 2005; Sim and Puria, 2008; O'Connor and Puria, 2008) than in the previous model where the joint was assumed to be rigid. Table 1 lists the material properties of the middle-ear components in the present model, along with those from the previous model. The

middle-ear-ligament materials in the present model were modified in such a way that the elastic modulus values were generally more uniform among the different ligaments. The loss factors, η , for some components were also increased from the previous model in order to better match the model results to the present temporal-bone data. Note that the loss factors for most components, except for the ossicles and the cochlear load, are assumed to be linearly increasing with frequency. As previously discussed in Homma et al. (2009), this approach is consistent with the stiffness-proportional Rayleigh damping model. For these frequency-dependent loss factors, reference values at 1 kHz are shown in Table 1. The material-property values used in the previous model are also shown in curly brackets. Poisson's ratios were assumed to be 0.3 for all components. Further information on the material-property values of the middle-ear components can be found in Homma et al. (2009).

As discussed in Homma et al. (2009), the first three middle-ear vibration modes mainly characterize the dynamics of the middle ear in the frequency range of discussion here, which is up to around 4 kHz. This basic characteristic did not change for the present model, but the frequencies of the three vibration modes are slightly shifted: 1.10, 1.54, and 2.38 kHz for the present model compared with 1.15, 1.69, and 2.76 kHz for the previous model. The frequency responses were similar between the two models, but the resonance peaks were more damped for the present model due to the increase in the material loss factors in some components as shown in Table 1.

2.2.2. Simulated AC and BC excitations—AC excitations were simulated by applying uniformly-distributed dynamic pressures over the TM surface on the ear-canal side. BC excitations were simulated by assigning uniform (both magnitude and phase) displacement vectors at the boundaries of the structure, such as at the ends of the ligaments and tendons, and at the edge of the tympanic annulus. This produced simulated rigid-body vibrations of the base temporal-bone structure. The direction of the BC excitation was perpendicular to the plane of the tympanic annulus, which was approximately consistent with the temporal-bone measurements.

2.2.3. Simulated ear-canal pressure effect—The effects of ear-canal pressurization on AC and BC responses were emulated in this model by adjusting the elastic modulus values of the middle-ear components that may be contributing to the stiffening of the middle-ear structure. These included the TM, the TM annulus, the anterior, lateral, and superior ligaments of the malleus, the tensor-tympani tendon, and the annular ligament of the stapes. This was essentially an iterative trial-and-error model-tuning process. First, modulus gain factors (i.e. multiplication factors) were determined for the model components by matching simulated AC responses at the umbo and the stapes against the corresponding measured data at different static pressure levels. The effects on the umbo/LP response ratio, which indicates the underlying malleus-handle motion, were also considered during the tuning process. Once the modulus gain factors of all components were determined for each static pressure case, then the same factors were applied for the subsequent BC-response simulations. Furthermore, in order to gain insight into the approximate locations of the stiffening within the middle-ear structure, the modulus gain factors were differentiated into

the following three groups: Group I (for the TM and TM annulus), Group II (for the anterior, lateral, and superior ligaments of the malleus, as well as the tensor-tympani tendon), and Group III (for the annular ligament of the stapes). Fig. 3 shows the gain factors determined as a result of the model-tuning process.

The elastic-modulus values of other elastic components in the middle-ear model, namely the incus ligament, stapes tendon, and incus–stapes and incus–malleus joints, were assumed to be constant. This is based on a judgment that the level of pressure-induced stiffening occurring at these components may be relatively small. A previous FE-simulation study by Wang et al. (2007), which simulated the pressure-induced stiffening of middle-ear components and indicated relatively little stiffening due to pressure in these components, thereby influenced the present decision to exclude these components for the simulations of the effects of static pressure.

The grouping and the exclusion of the components were also necessary in order to make it practical to perform the manual trial-and-error iteration process by keeping the model's degrees of freedom sufficiently low. If the degrees of freedom were further increased by allowing the tuning of the elastic modulus of each and every elastic component in the model, then the potential to obtain a better model fit would have been improved. However, this would have made the manual iteration process impractical, especially considering that the model had to simulate both AC and BC responses. It will be shown shortly that, despite this restraint on the degrees of freedom, a reasonable degree of agreement was obtained between the FE-simulated and measured pressure-effect data.

3. Results

3.1. Effects of ear-canal static pressure on AC responses

Fig. 4 shows the measured AC responses for each of the ear-canal pressurization cases, along with the corresponding FE-simulation results. The measured data shown are the means of the eight temporal bones. Both the umbo and the stapes responses are shown.

Discussing the measured data first, the baseline umbo and stapes AC responses (i.e. with zero static pressure) show typical response characteristics, with the response maximum occurring in the mid-frequency range around 1 kHz for both the umbo and the stapes. The baseline responses for “+0” and “−0” mm H₂O (see Section 2) are largely consistent. With the ear-canal pressurization, AC responses are reduced by about 10–12 dB and 15–18 dB for ear-canal pressures of ± 200 mm H₂O and ± 400 mm H₂O, respectively. Magnitudes of the AC-response reductions are observed to be similar between positive and negative pressures. The AC-response reductions are observed mainly at frequencies below the response peak of the baseline responses around 1 kHz. The response peaks are shifted to higher frequencies as the ear-canal pressurization level increases, with these shifted response peaks resulting in slight amplifications over the baseline for a small range of high frequencies.

These characteristics are consistent with the effects of structural stiffening, and they are also exhibited by the FE-simulation results in Fig. 4, in which the effects of ear-canal pressurization were emulated through the stiffening of middle-ear components. The

characteristics of the structural stiffening are more clearly observable in the FE results than in the mean temporal-bone data, since detailed modal characteristics in the mean temporal-bone data tend to be smoothed over by averaging. FE results clearly show the presence of a resonance peak at around 1.1 kHz for the zero-pressure case. Then, with the static pressure applied, the resonance peak shifts to around 2 kHz for ± 200 mm H₂O and 2.5–3 kHz for ± 400 mm H₂O, which causes the response amplifications at the shifted-resonance-peak frequencies.

Fig. 5 shows the measured and the FE-simulated relative responses between the umbo and the LP (see Fig. 1), which reveal the 2D rigid-body motions of the malleus handle underlying the TM. Examining the baseline zero-pressure cases first, in Fig. 5, the umbo/LP magnitude ratios tend to be relatively constant at around 10 dB in the low-frequency range. This constant-magnitude-ratio region ends at around 1 kHz, and the ratio drops in magnitude to around 5 dB at higher frequencies. This rather distinct shift in the malleus-handle motion results from a shift in the dominant vibrational mode of the middle ear, which typically occurs in the 1–2 kHz range (Homma et al., 2009). The cases with ear-canal pressurization show rather pronounced differences between the positive and negative pressure cases. As shown in the measured data in Fig. 5(a), the transition region, where the magnitude shifts from 10 dB to around 5 dB, is seen to be shifted to higher frequencies for negative pressures, but not for positive pressures. Similar characteristics are also exhibited by the FE-model results shown in Fig. 5(b), as a result of the asymmetric distribution of the components' modulus gain factors with respect to pressure polarity as indicated in Fig. 3, although the degree of the high-frequency shifting seems to be more strongly exhibited by the FE model than by the measured data for the positive-pressure cases.

3.2. Effects of ear-canal static pressure on BC responses

Fig. 6 shows the effects of the ear-canal pressurization on the middle-ear BC responses obtained at the umbo, v_{umbo}/v_b , and at the stapes, v_{stps}/v_b , as obtained from the temporal-bone measurements and the FE simulations.

Observing the temporal-bone data in Fig. 6(a and c) first, the BC responses of both the umbo and the stapes show magnitude reductions due to the ear-canal pressurization at frequencies in the 1–2 kHz range. In a similar manner to the AC cases in Fig. 4, some high-frequency amplifications are also recognizable at frequencies above 3 kHz. There also appear to be noticeable differences, with respect to pressure polarity, in the degree of the BC-response reductions, such that negative pressures show a higher degree of response reduction than positive pressures. It can also be seen that the baseline zero-pressure stapes responses at +0 and –0 mm H₂O show some differences, although the respective baseline umbo responses are relatively more consistent.

The corresponding FE results shown in Fig. 6(b and d) show a level of similarity in terms of the overall response characteristics to those of the measured data. Similar to the measured data, the FE-simulated results show the most pronounced BC-response reductions to be in the mid-frequency range around 1–2 kHz. The FE results clearly indicate that this reduction is due to a frequency shift of the resonance peak, which is around 1.6 kHz for the baseline case, into higher frequencies. The FE results also exhibit the high-frequency response

amplifications due apparently to the repositioned resonance peak. However, the FE results for the umbo responses appear to largely overestimate the absolute level of the BC-response reductions, although the simulated BC reductions for the stapes appear to be more reasonable.

3.3. Comparison of measured AC- and BC-response attenuations

Fig. 7 compares the measured attenuations in the AC and BC responses obtained at both the umbo and the stapes due to the ear-canal pressurization. The results are shown in one-third-octave-frequency-band format, which is obtained by first integrating the measured narrow-band temporal-bone response data of Figs. 4 and 6 falling within each one-third-octave frequency band, and then calculating the dB differences between the pressurized and the respective baseline cases.

Fig. 7 shows that the attenuation effects for BC are most prominent in the mid-frequency range centered around 1–2 kHz, while the attenuations for AC responses are most prominent at frequencies below 1 kHz. The magnitudes of the BC-response reductions are observed to be higher for negative pressures than for positive pressures, with up to $8.6(\pm 4.7)$ dB of peak reduction at 1.6 kHz for negative pressures, and up to about $5.3(\pm 7.9)$ dB of peak reduction at 1.25 kHz for positive pressures. The cross-over frequency at which the response attenuation crosses the zero-dB line from positive (i.e. attenuation) to negative (i.e. amplification) tends to be higher for BC than for AC. The attenuation characteristics between the umbo and the stapes responses are observed to be similar.

4. Discussion

4.1. Comparisons with psychoacoustic data

Fig. 8 compares the BC-response attenuations, based on the present temporal-bone measurements, with the psychoacoustic data from Humes (1979) and Aazh et al. (2005). In these psychoacoustic studies, BC hearing-threshold changes due to ear-canal pressurization were measured with a BC vibrator placed on the mastoid region of the head. As shown in Fig. 8, the attenuation trends in the temporal-bone data for ± 200 and ± 400 mm H₂O resemble those of the psychoacoustic data which were obtained for ± 300 mm H₂O. Similar attenuation differences with respect to ear-canal pressure polarity are also observable in both the temporal-bone and psychoacoustic data. The characteristics of the present temporal-bone data in Fig. 8 are, for the mid-frequency range centered around 1.5–2 kHz, largely consistent with past studies that describe the involvement of the middle-ear BC mechanism in BC hearing (Carhart, 1971; Tonndorf, 1972; Linstrom et al., 2001; Stenfelt et al., 2002; Homma et al., 2009).

The 95% confidence intervals at each frequency band have been calculated for the temporal-bone data for ± 400 mm H₂O, and are shown in Fig. 8. For -400 mm H₂O, the confidence intervals at 1, 1.6, 2, and 4 kHz do not include the zero-dB line, which indicates that statistically-significant changes are obtained at these frequencies. However, for $+400$ mm H₂O, the confidence intervals indicate that the measured changes are not statistically significant.

The temporal-bone BC-attenuation data appear to show a decreasing trend toward low frequencies below 1 kHz, judging from the overall decrease in the mean values as the frequency decreases from 1.6 to 0.8 kHz, while those in the psychoacoustic data show an overall increasing trend below 1 kHz. Assuming this extrapolation of the BC data is valid, the discrepancy with the psychoacoustic data may be attributable to the differing contributions of the external-canal BC component between the temporal-bone measurements and the live-subject measurements. The external-canal BC component results from the eardrum being driven by the acoustic pressure created from the BC-induced vibrations of the cartilaginous ear canal, which is often occluded by an earplug during psychoacoustic BC testing. It should be noted that an earplug, especially if it is not deeply inserted into the bony portion of the ear canal, can vibrate in response to BC excitations and introduce significant acoustic pressures within the occluded air volume. In this case, the middle ear is driven essentially by AC excitations, with the only difference being that the ear-canal acoustic pressure was introduced by the BC excitation in this case. For live-subject ears, this external BC component dominates the BC sound at low frequencies (Khanna et al., 1976). On the other hand, the external-canal BC component was negligible in the temporal-bone measurements, by virtue of there being no significant differences between the temporal-bone BC responses with and without the lid covering the ear-canal tube (Fig. 1). This indicates that in the temporal-bone cases, the middle-ear structure was driven primarily by the base bone vibration and not by the acoustic pressure in the ear-canal tube. With the low-frequency BC responses in the live test subjects being dominated by the external ear-canal BC component, it is reasonable to observe significant hearing-suppression effects at low frequencies in the psychoacoustic data, which are consistent with the AC-response attenuation levels shown for the temporal-bone data at low frequencies in Fig. 7.

Fig. 8 also shows the psychoacoustic data for ear-canal pressures of ± 500 mm H₂O (Humes, 1979), which is even higher than the ± 400 mm H₂O used in the present temporal-bone measurements. At ± 500 mm H₂O, the data shows BC-hearing-sensitivity reductions that are significantly greater than the reductions with lower pressure levels. It is plausible that at these high pressures the annular ligament of the stapes, which appears to be stiffened by only a relatively small amount compared to other components for the range of pressures used in this study, may be stiffened even further, thereby resulting in the significant increase in the BC hearing reductions. This reasoning is based on the knowledge from Carhart (1971) and also from the Gelle test (Kyle, 1907) that the immobilization of the stapes footplate by otosclerosis results in a significant reduction in BC hearing sensitivity. According to the Carhart's study, the mean BC sensitivity reduction due to otosclerosis is 15 dB at 2 kHz. This value is consistent with the 13 dB reduction observed for the -500 mm H₂O case in Fig. 8. This seems to suggest that, with a very high pressure, stiffening at the stapes annular ligament may become more substantial, which could result in greater BC-response-reduction levels than those observed for the present study. Therefore, it would be of interest in the future to improve the pressurization technique in the temporal-bone measurements to allow ± 500 mm H₂O, to determine if the same degree of BC-response reductions occurs at the stapes for these higher pressures.

4.2. Shifts in the middle-ear dynamic response

From the characteristic shifts of the response peaks observed in the temporal-bone data and from the FE results, it is clear that the primary effect of ear-canal pressurization on middle-ear dynamics is one of structural stiffening. The response-shift characteristics observed in the present temporal-bone AC responses in Fig. 4 are largely consistent with those obtained by previous studies (Murakami et al., 1997; Gan et al., 2006). Suppression of the AC response due to ear-canal pressurization mainly occurs at frequencies below around 2 kHz, and is accompanied by some response amplifications at higher frequencies. Since the primary middle-ear resonance frequency for AC is typically around 1 kHz, the structural stiffening mainly reduces the AC responses at and below this frequency. The high-frequency amplifications are consequences of the middle-ear resonance being shifted to higher frequencies due to stiffening. These characteristics are more clearly observed in the corresponding FE results in Fig. 4(b and d) rather than in the mean temporal-bone data, for which the fine individual-response details tend to be smoothed over by the averaging.

It can be observed from the temporal-bone data and the FE results that the middle-ear stiffening introduced by the ear-canal pressurization attenuates not only the AC responses, but also the BC responses. For a -400 mm H₂O static pressure, about 5–8 dB of BC-response reductions were observed in the frequency range from 0.8 to 2.5 kHz with the most significant reduction obtained around 1–2 kHz with a peak attenuation of $8.6(\pm 4.7)$ dB at 1.6 kHz. It is also observed that the frequency range of the BC-response suppressions tends to extend into higher frequencies than the frequency range of the AC-response suppressions. This observation is consistent with the previous finding that the primary resonance frequency for BC is typically in the 1.5–2 kHz range, which is higher than the 0.8–1.2 kHz range for AC (Stenfelt et al., 2002; Homma et al., 2009). A recent study by Homma et al. (2009) revealed that the middle-ear ossicular structure exhibits two resonance modes in the 1–2 kHz range, where the first mode, with its mean resonance frequency of 1.2 kHz, is the dominant resonance mode for AC, while the second mode, with its mean resonance frequency of 1.7 kHz, is the dominant resonance mode for BC. Since the stiffening causes a prominent attenuation at the nominal resonance frequency, it is reasonable that the most BC attenuation should be seen in Fig. 7 in the 1.5–2 kHz range, which is near the typical BC resonance frequency. These characteristics are more clearly exhibited by the FE results in Fig. 6, which show the shift in the BC resonance peak at around 1.6 kHz due to structural stiffening, and the resulting BC-response attenuation peak at that frequency.

At this juncture, a few words of clarification about the fundamental nature of the middle-ear structural-resonance phenomena may be of merit to support the discussion. It is important to recognize that the “middle-ear resonance mode” is a property of the middle-ear structure as a whole, and not a property that is associated with any particular middle-ear components. Therefore, the resonance peaks seen in the FE-simulated BC responses at about 1.6 kHz in Fig. 6(b) for the umbo and (d) for the stapes are originated from the middle-ear resonance mode, differing only by the location of observation.

4.3. Mechanisms of middle-ear structural stiffening

The details of the mechanisms underlying the middle-ear stiffening due to static-pressure loading of the eardrum have not been clearly understood. Opinions appear to differ as to the fundamental nature of the middle-ear stiffening, which could involve several distinct mechanisms such as: (a) material stiffening, (b) the effects of large structural deformations, and (c) pre-stress stiffening. The first mechanism refers to the effects of nonlinear material properties where the elastic material modulus is dependent on the static strain applied. The second mechanism refers to the effects of the change in the geometry due to large structural deformations. The third mechanism refers to the stiffening of the structure as a result of the presence of tension force. Examples of this last mechanism are strings or thin membranes, whose effective structural stiffness depends on the applied tension.

Ladak et al. (2006) studied nonlinear static deformations of a pressurized cat eardrum, without accounting for material nonlinearity in their simulations. Since their simulation results appear to agree with experimental data to a reasonable degree, in spite of ignoring material nonlinearity, this implies the relative importance of the other two factors over the effects of material stiffening. In contrast, FE-simulation results offered by Wang et al. (2007) did account for the material stiffening effect, and in that case it appeared to contribute significantly to the overall middle-ear stiffening. They also suggested that the pre-stress stiffening effect may not be significant, although their analysis on this point was not conclusive.

The FE modeling in the present study did not involve explicitly simulating the underlying nonlinear mechanical phenomena involved. Instead, the middle-ear stiffening was simulated by altering the elastic modulus values of components such that the observed changes in dynamic characteristics agreed with those from the measured data. This effectively lumped all structural stiffening effects into the material stiffening mechanism, with the other two mechanisms ignored. Therefore, the present FE analysis did not reveal the nature of the middle-ear stiffening mechanism.

4.4. Location of stiffening and the effects of pressurization polarity

It is known that the middle-ear stiffening does not involve just the TM, but also involves various other middle-ear components, such as ligaments and tendons. However, it is still not clear as to the exact degree of the stiffening that is occurring at each component and its relative contribution to the stiffening of the middle-ear structure as a whole. The present FE results indicate that the higher level of stiffening may be occurring at the TM and at the components in the close vicinity of the TM, rather than at the stapes, for the range of pressures employed in this study. The component modulus gain factors shown in Fig. 3 show significantly higher gains for Group I (the TM and TM annular ligament) and Group II (the superior, anterior, and lateral ligaments of the malleus, as well as the tensor-tympani tendon) components than for the Group III component (the stapes annular ligament). This was motivated from an observation during the model-tuning process that an excessive stiffening of the stapes annular ligament tended to result in higher response attenuations at the stapes relative to those at the umbo, instead of the relatively comparable attenuations between the two locations observed in the temporal-bone data. This is intuitively reasonable

considering that the static pressure in the ear canal acts on the TM, therefore the resulting loading would be relatively high at the TM and components near the TM.

Fig. 3 also suggests a potential difference in the distribution of stiffening depending on the polarity of the ear-canal pressurization. The figure shows that the stiffening is equally distributed between the Group I and Group II components for negative pressures, while for positive pressures the distribution is skewed more toward the Group I components than the Group II components. This asymmetric distribution with respect to pressure polarity was produced during the model-tuning process by an attempt to simulate the distinct differences observed in the malleus-handle responses between positive and negative pressures, as seen in Fig. 5. As shown Fig. 5, the FE results with this particular stiffening distribution appear to capture these differences. This may be an indication that the ligaments around the malleus and the tensor-tympani tendon are strained more for negative pressures than for positive pressures, whereas the TM may be strained more for positive pressures than for negative pressures. Other studies have shown results that appear to be consistent with this observation. A FE-modeling study by Wang et al. (2007) also indicated that the stress level in the TM is higher when the TM is displaced inward by negative middle-ear cavity pressure (equivalent to the positive ear-canal pressure in the present case). Their FE result also shows a similar trend in which the superior ligament of the malleus and the tensor-tympani tendon are stiffened more when the TM is pushed outward due to positive middle-ear cavity pressure (equivalent to the negative ear-canal pressure in the present case). Hüttenbrink (1988) observed in his temporal-bone measurements that the increase in the static displacements of the malleus handle, as a result of cutting the superior ligament of the malleus and the tensor-tympani tendon, are more significant for negative ear-canal pressure than for positive pressure. This result suggests that the two components may be strained more for negative ear-canal pressures than for positive pressures, which is consistent with the present analysis.

It is believed that this difference in the distribution of stiffening is likely responsible for the differences observed in the BC-re-sponse attenuation data in Fig. 7, where negative ear-canal pressures are more effective than positive pressures at producing response attenuations. The FE results in Fig. 6(b and d) exhibit some characteristics that may relate to these observed differences. For example, the FE results in Fig. 6 show that the degree to which the BC resonance is shifted in frequency is more pronounced for negative pressures than for positive pressures. This suggests that the negative pressures are more effective at increasing the effective stiffness associated with the BC resonance mode than the positive pressures. This appears to be intuitively reasonable if one considers the motional characteristics of the BC resonance mode. As discussed in Homma et al. (2009), the vibratory mode of the middle-ear structure at the BC resonance frequency, which has been found to be different from the motion associated with the AC resonance mode, exhibits a relatively high level of participation by the malleus ligaments and the tensor-tympani tendon compared to other middle-ear components. With these components possibly stiffened more for negative pressures than for positive pressures, as suggested in the preceding discussion, it is reasonable to observe the BC resonance being shifted higher in frequency for negative pressures than for positive pressures.

4.5. Potential implications for hearing-protection applications

The motivation of the present study is to ultimately provide improved hearing protection for workers who are subjected to extremely loud acoustic environments (e.g. 140–150 dB). As discussed earlier, if one is to improve hearing protection beyond the HPD performance limit set by the BC noise transmissions (i.e. the BC limit), then the attenuation of the HPD at around 1.5–2 kHz needs to be significantly improved beyond the BC-limit level, which is typically about 40 dB (Zwislocki, 1957; Berger et al., 2003; Reinfeldt et al., 2007). The present temporal-bone data indicate that, by introducing an ear-canal static pressure of -400 mm H₂O, the BC response at the stapes could be reduced by $8.6(\pm 4.7)$ dB in the frequency range centered at 1.5–2 kHz. If the BC-limit peak at 1.5–2 kHz is in fact controlled by the middle-ear BC component, then this could translate into a comparable level of improvement in effective noise-attenuation beyond the normal 40 dB limit at this critical frequency range. This is a significant improvement considering that even reducing the noise level by 3 dB would mean a 50% reduction in noise energy, which translates into a doubling of the allowable noise-exposure time. However, it should be noted that the BC noise transmission in the hearing protection application occurs in the form of more distributed acoustical excitations of the skull rather than a direct, single-point mechanical excitation from a BC vibrator. Therefore it is still not clear if the BC-response reduction observed in the present study directly translates into the noise-attenuation improvement. This would need to be investigated in the future using a psychoacoustic experiment conducted with acoustically-induced BC excitations. It is also seen from Figs. 7 and 8 that there is about 4 dB of high-frequency BC-response amplification at around 4 kHz. Although its magnitude is smaller than that of the BC-response reduction, this is clearly undesirable since it means an amplification of high-frequency noise. Again, it remains to be seen how much of this high-frequency amplification would actually manifest itself in the future psychoacoustic experiment with acoustically-induced BC excitations.

5. Conclusion

The effects of ear-canal pressurization on the middle-ear responses under BC- and AC-excitation conditions were studied using temporal-bone measurements coupled with FE simulations. The temporal-bone measurement data show attenuations of the BC responses in the mid-frequency range around 1–2 kHz as a result of ear-canal pressurization. These BC-response attenuation characteristics are largely consistent with the available psychoacoustic data, thus further supporting the importance of the middle-ear BC component contribution to BC sound transmission in the mid-frequency range. FE analysis indicates that the structural stiffening due to pressure loading of the eardrum causes the primary BC resonance, which is typically at around 1.5–2 kHz, to shift to a higher frequency, thus resulting in a pronounced BC-response reduction centered around that frequency. FE analysis also suggests that the middle-ear stiffening may be associated more with the TM and the ligaments around the malleus, but less with the stapes annular ligament for the range of static pressures studied. Measurements also indicate that the BC responses are attenuated more effectively by negative ear-canal pressures than by positive pressures. This may be because the malleus ligaments and the tensor-tympani tendon are strained and stiffened more due to negative

pressures than due to positive pressures, thus resulting in the further suppression of the ossicular vibration mode dominant in BC.

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Abbreviations

AC	air conduction
BC	bone conduction
FE	finite element
HPD	hearing protection device
SPL	sound pressure level
TM	tympanic membrane
LP	lateral process

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Fig. 1. Temporal-bone measurement setup. (a) Overall view, (b) TM measurement locations viewed from above through the plastic tube, from the perspective of the microscope mounted with the laser-vibrometer.

Fig. 2.

A human middle-ear FE-model (left ear). The present model is mostly identical to a previous model described by Homma et al. (2009), except for the addition of the superior ligament of the malleus and adjustments to the material properties of some components (see Table 1).

Fig. 3.

Elastic-modulus gain factors used to emulate the middle-ear stiffening effects due to ear-canal pressurization. Gain factors were differentiated among three component groups: Group I (for the TM and TM annulus), Group II (for the lateral, anterior, and superior ligaments of the malleus, as well as for the tensor-tympani tendon) and Group III (for the annular ligament of the stapes). Note that larger modulus gain factors are assigned for components belonging to Groups I and II, while for Group III the values are relatively minimal. Also note the asymmetry with respect to the pressure polarity for Groups I and II.

Fig. 4. AC-response shifts due to ear-canal pressurization. (a) Measured mean and standard deviation (S.D.) of umbo responses, v_{umbo}/p_{ec} . (b) FE-simulated umbo responses, v_{umbo}/p_{ec} . (c) Measured mean and S.D. of stapes responses, v_{stps}/p_{ec} . (d) FE-simulated stapes responses, v_{stps}/p_{ec} .

Fig. 5. Effects of ear-canal pressurization on the umbo/LP AC-response velocity ratios. (a) Measured mean and S.D., (b) FE-simulated responses.

Fig. 6.

BC-response changes due to ear-canal pressurization. (a) Measured mean and S.D. of umbo responses, v_{umbo}/v_b , (b) FE-simulated umbo responses, v_{umbo}/v_b , (c) Measured mean and S.D. of stapes responses, v_{stps}/v_b , (d) FE-simulated stapes responses, v_{stps}/v_b .

Fig. 7.

Comparisons of the mean and S.D. ($N = 8$) of measured AC- and BC-response attenuations due to ear-canal pressurization. (a) Umbo, negative pressures, (b) Umbo, positive pressures, (c) Stapes, negative pressures, (d) Stapes, positive pressures. Plots are given in one-third-octave-band format.

Fig. 8.

Comparison of BC-response attenuations due to ear-canal pressurization from the temporal-bone measurement data (stapes responses, $N = 8$) with the psychoacoustic data from the literature. (a) Negative pressures, (b) positive pressures. Temporal-bone data are from the BC-response attenuation results obtained at the stapes. The psychoacoustic data are reports of the changes in the BC-threshold shift due to static pressure. Confidence intervals (95%) are also shown for the ± 400 mm H₂O temporal-bone data. Psychoacoustic data for ± 300 and ± 500 mm H₂O ($N = 8$) are from Humes (1979). The data for -350 mm H₂O ($N = 22$) are from Aazh et al. (2005).

Table 1

Component material properties of the middle-ear FE-model. Values used in the previous model (Homma et al., 2009), if different from the present model, are shown in curly brackets. Those components whose elastic modulus values were adjusted for emulating the pressure stiffening effects (see Fig. 3) are indicated by “*”. Poisson’s ratios were assumed to be 0.3 for all components. References on available material-property values from the literature are also found in Homma et al. (2009).

Component	Elastic modulus E [N/m ²]	Density ρ [kg/m ³]	Loss factor η
Incus ossicle	1.41×10^{10}	2.15×10^3	0.01(constant)
Malleus ossicle	1.41×10^{10}	2.39×10^3	0.01(constant)
Stapes ossicle	1.41×10^{10}	2.20×10^3	0.01(constant)
Malleus/incus joint	$7 \times 10^6\{1.41 \times 10^{10}\}$	1.2×10^3	0.15 at 1 kHz
Incus/stapes joint	4.4×10^5	1.2×10^3	0.15 at 1 kHz
Tensor tympani*	$5 \times 10^6\{1.9 \times 10^7\}$	1.2×10^3	0.3{0.15} at 1 kHz
Anterior ligament*	$5 \times 10^6\{1.5 \times 10^7\}$	1.2×10^3	0.3{0.15} at 1 kHz
Lateral ligament*	$10 \times 10^6\{5.0 \times 10^5\}$	1.2×10^3	0.3{0.15} at 1 kHz
Superior ligament*	2×10^6	1.2×10^3	0.3 at 1 kHz
Stapes tendon	3.8×10^5	1.2×10^3	0.15 at 1 kHz
Incus ligament	4.8×10^6	1.2×10^3	0.15 at 1 kHz
Tympanic membrane, pars tensa*	$2 \times 10^7\{3 \times 10^7\}$	1.2×10^3	0.7{0.15} at 1 kHz
Tympanic membrane, pars flaccida	0.7×10^7	1.2×10^3	0.15 at 1 kHz
Tympanic annulus*	$4 \times 10^5\{6 \times 10^5\}$	1.2×10^3	0.3{0.15} at 1 kHz
Stapes annular ligament*	4.1×10^5	1.2×10^3	0.25 at 1 kHz
Cochlear load	Effective resistance: 30 G Ω (mass and stiffness are assumed to be insignificant)		