The outer ear pathway during hearing by bone conduction

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A B S T R A C T

There have been conflicting reports in the literature about the importance of the induced ear canal sound pressure for the perception of bone-conducted (BC) sound. Here we investigated this by comparing the ear canal sound pressure at threshold for air-conducted (AC) and BC stimulation. Twenty-one adults with subjectively normal hearing function participated. They were tested for their hearing thresholds in the frequency range 250 Hz to 12.5 kHz with AC and BC stimulation and the ear canal sound pressure within 5 mm of the eardrum was obtained with probe tube microphones. Contralateral masking used with BC stimulation shifted the hearing threshold by 5 to 10 dB due to central masking effects. When the ear canal sound pressures at threshold were investigated, the results indicate that the ear canal component for hearing BC sound is around 10 dB below other contributors at frequencies below 2 kHz and similar to other important contributors at frequencies between 2 and 4 kHz. At frequencies above 4 kHz, the contribution from the ear canal sound pressure on BC hearing declines and was around 40 dB below other contributors at 12.5 kHz. The contribution of the ear canal sound pressure in the mid-frequency region is facilitated by the ear canal resonance occurring in this frequency area. The results were similar irrespective of stimulation position. The study also revealed problems estimating the force out of BC transducers caused by a shift in resonance frequency when the artificial mastoid impedance deviates from the impedance of human mastoids. The current study indicates that model predictions have underestimated the contribution from the ear canal sound pressure on BC hearing by around 10 dB.

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1. Introduction

Air-conduction (AC) and bone-conduction (BC) are the two mechanisms by which sound is heard. Most sounds are heard through AC in which sounds are perceived when airborne sound induces vibration of the eardrum in the ear canal. The eardrum vibrations are then transmitted through the middle ear ossicles to the inner ear fluid. These vibrations excite the hair cells in the cochlea that transmit the information through the auditory nerve cells to the brain that finally perceives the sound by processing the electrical impulses. In the BC mechanism of hearing sounds, vibrations conducted through the skull bones can excite the inner ear and cochlea directly bypassing the outer and the middle ears (Dauman, 2013; Stenfelt, 2011, 2016, 2020; Stenfelt and Goode, 2005). In such case, the transmission medium is the skull bone rather than the air as in AC. However, the final excitation of the basilar membrane in the inner ear is the same for both AC and BC sounds where an airborne sound can cancel a BC sound (Khanna et al., 1976; Stenfelt, 2007; V Békésy, 1932).

The theory for BC hearing suggests that there are different pathways that the BC sound takes to ultimately excite the basilar membrane in the inner ear (Stenfelt and Goode, 2005; Tomndorf, 1966). These pathways should be understood as different mechanisms that can transform and transmit a vibration of the skull bone to a vibration of the basilar membrane in the inner ear. Throughout history, several such pathways have been suggested, and nowadays five pathways are usually considered (Stenfelt, 2011; Stenfelt and Goode, 2005):

1 Sound pressure induced in the ear canal (Stenfelt et al., 2003).
2 Inertial forces on the middle ear ossicles (Dobrev et al., 2020; Homma et al., 2009; Röösli et al., 2012; Stenfelt, 2006; Stenfelt et al., 2002).
3 Inertial forces of the inner ear ossicles (Kim et al., 2011; Stenfelt, 2015).
4 Compression and expansion of the inner ear space (Stenfelt, 2015), and

Abbreviations: AC, Air-conduction; BC, Bone-conduction; 3AFC, Three-alternative forced-choice; MAP, Minimal audible pressure; RETFL, Reference equivalent threshold force levels; TM, Tympanic membrane; dB FL, Decibels in force level; dB SPL, Decibels in sound pressure level; dB HL, Decibels in hearing level.

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5 Sound pressure transmission from the intracranial space (Dobrev et al., 2019; Roosli et al., 2016; Sohmer et al., 2000).

Of these, the inner ear component, i.e. inertial forces of the inner ear fluid and compression and expansion of the inner ear space are considered the most important (Stenfelt, 2016, 2020). The inertial forces on the middle ear ossicles have been suggested to be important at frequencies around the ossicles resonance frequency, which occurs at 1.5 to 2 kHz for BC stimulation (Homma et al., 2009; Stenfelt et al., 2002). The sound pressure induced in the ear canal and the sound pressure transmission from the skull interior are considered to have only minor importance for BC hearing in the normal ear. It should be noted that the relative importance of the different pathways depends on the stimulation position and the above reasoning is for a BC stimulation at the mastoid behind the ear where the transducer is not in contact with the pinna.

According to recent modelling studies, the ear canal sound pressure component is 20 to 40 dB below other contributors for BC perception (Stenfelt, 2016, 2020). These data use ear canal sound pressures obtained in cadaver heads (Stenfelt et al., 2003) which may deviate from the ear canal sound pressure in a live human. There are indications that the ear canal sound pressure component in BC hearing has a relatively similar importance to other components, at least in limited frequency ranges. Huizing (1960) measured the ear canal sound pressure for both AC and BC stimulation in an open ear canal and reported the sound pressure to be greater for BC stimulation compared to AC stimulation at threshold for frequencies below 500 Hz, while the opposite was reported for frequencies above 500 Hz. If the ear canal sound pressure with AC stimulation at threshold is the lowest level that can be heard, any sound level above this limit is detectable. And if that is the case with BC stimulation, it indicates that the ear canal sound pressure pathway is important for hearing BC sound. In a similar way, Khanna et al. (1976) found the open outer ear pathway to dominate BC hearing at frequencies below 800 Hz but not above that frequency.

One argument against the importance of the outer ear canal sound pressure for BC hearing is that the ear canal sound pressure increase after occluding the ear canal has been reported to be 5 to 15 dB greater than the change in hearing thresholds (Goldstein and Hayes, 1971; Huizing, 1960). Reinfeldt et al. (2013) showed that the change in hearing thresholds and ear canal sound pressure with occluding the ear canal differed at low frequencies (below 500 Hz) and high frequencies (above 3000 Hz), but not in the mid-frequency range. The differences were also dependent on the stimulation position. Another study on the occlusion effect where the occlusion was achieved at three positions: deep in the ear canal, close to the ear canal opening, or circumaural, showed nearly the same changes of ear canal sound pressure and hearing thresholds at frequencies above 300 Hz, while a difference of approximately 10 dB was found at the lowest frequencies investigated (Stenfelt and Reinfeldt, 2007). One explanation for the differences at the lowest frequencies can be the masking caused by the noise produced by the body itself and is estimated to falsely alter the hearing thresholds by 5 dB at frequencies at and below 250 Hz (Berger and Kerivan, 1983).

The importance of the BC ear canal sound pressure pathway is not entirely clear and the aim of the current study is to investigate the importance of this pathway in the frequency range 0.25 to 12.5 kHz. This is accomplished by measurements of the ear canal sound pressure at hearing thresholds when stimulation is by AC and by BC on the skin-covered mastoid. To also investigate the influence of stimulation position, the analysis of the BC stimulation is done for both ipsilateral and contralateral mastoid stimulation.

2. Method

2.1. Ethics

The current study was approved by the Swedish Ethical Review Authority (2020-01235).

2.2. Participants

Participants for the study were recruited by public advertisement. Inclusion criteria for the study were an age between 18 and 60 years, to have subjectively reported normal hearing, and no tinnitus problem. Twenty-one of the respondents opted to participate after they received information about the study and signed an informed consent form. The participants’ mean age was 33 years (range 18 to 51 years) and seven were females. They received 500 Swedish krona for their participation.

2.3. Experimental setup

The AC stimulation was provided by ER3 insert earphones (Etymotic Research Inc, IL) where the foam insert of the ER3 was reduced to three small foam wings facilitating an open ear and avoiding an occlusion effect from the inserts themselves. The BC stimulation was provided by a specially designed bilateral BC transducer headset (Fig. 1) consisting of two motor units from the Radioear B81 BC transducer (Radioear, DK) connected to a CochlearTM Baha® SoundArc (Cochlear BAS, SE). This gave a placement of the two BC transducers on the mastoid in line with the upper part of the pinna with a static force between 2 and 3 Newtons (Fig. 1). The transducer-skin interface was a circular plastic adapter with a diameter of 15 mm. It was ensured that the transducers did not touch the pinna during BC stimulation. After the placement, the BC transducers were in position for the entire testing procedure.

The stimulation was provided by a computer connected to a 24-bit 96 kHz external sound card (TASCAM US-16×08, TEAC Corp., JP), and the output from two of the sound card outputs was routed through a power amplifier (Rotel RA-04 SE, Rotel Co, Ltd, JP) to the BC transducers (Fig. 2). The electric signals to the BC transducers were monitored through two inputs of an I/O card (NI USB-4431, National Instruments Corp, TX). Two probe tube microphones (ER7C, Etymotic Research Inc, IL) were placed bilaterally in the ear canal such that the openings of the probe tubes were within 5 mm from the tympanic membrane (TM). The ER7C microphones were connected to two inputs on the NI USB-4431 I/O card (Fig. 2). The ER3 earphones were connected to two outputs of the sound card directly, and both the sound card and the NI USB-4431 I/O card were connected to the same computer (Fig. 2). Specially developed software in MATLAB® provided the stimulation through the sound card and measured the signals through the NI USB-4431 I/O card simultaneously.

Due to standing waves in the ear canal at high frequencies, the position of the probe-tube microphone at up to 5 mm from the TM means that it can be just a couple of millimeters from a sound pressure minima in the ear canal at the highest frequency tested, 12.5 kHz. At 5 mm from a rigid terminating end of a tube, the sound pressure is 7 dB below the sound pressure at the terminating end (TM) for a frequency of 12.5 kHz. This deviation is -4.1 dB at 10 kHz and -2.9 dB at 8 kHz. Consequently, at the highest frequency the sound pressure measurement can be down to 7 dB below the sound pressure at the TM. The positions of the ER7C probe tube microphone is constant during all measurements for a participant, and this uncertainty affects the inter-individual data but not the intra-subject results.
2.4. Procedure

The research participant was seated in a chair in a sound isolated test booth, had a computer mouse on a small table on the right chair rest, and a computer screen in front. After placing the BC transducers, the ER3 modified inserts, and the ER7C probe tube microphones, the ear canal sound pressures were obtained with AC stimulation. The stimulation was provided bilaterally, and the ear canal sound pressures were obtained bilaterally. After that, the AC hearing thresholds were obtained. All hearing thresholds were measured with an adaptive three-alternative forced-choice (3AFC) procedure with one one-up two-down algorithm. During the stimulation presentation, three horizontally aligned boxes numbered 1 to 3 were presented on the screen in front of the participant. In sequence, each box became green for one second with 500 ms between each green box. During one of the three periods indicated by a green box, the stimulation was presented, and the participant had to decide during which period the tone was heard by clicking with the mouse on the corresponding box. In the absence of a “Not sure” or “Did not hear” button, the participants were required to respond by selecting some box even if no tone was heard in any of the three periods to proceed to the next step. The first presentation level was around 40 dB above the normal hearing threshold to make sure that the presented tone is initially heard. Making the tone initially audible facilitates a faster judgment on whether the tone is heard or not in the subsequent steps. A correct identification of the stimulation period meant a decrease of the stimulation level by 10 dB and an incorrect identification led to an increase by 10 dB. This was done for the two first runs, then the step-size was reduced to 5 dB for the next two runs (one run means changing from increasing level to decreasing level, or vice versa). After that, four runs were conducted with a step size of 1 dB, and two correctly identified stimulation periods were required to decrease the level while only one error resulted in an increase of the stimulation level. The hearing threshold was computed as the mean of the two last peaks and valleys (mean of four extreme points). To avoid false positives during unheard stimulation levels where participants had a tendency to maintain clicking on the same box (stimulation interval), a subsequent stimulation never appeared directly at the same interval. Such an arrangement ensured that when the same box was clicked consistently, no peak or valley was reached. This information was not disclosed to the participant.

The hearing thresholds using this routine were obtained for AC stimulation in the left ear (test ear) at frequencies of 0.125, 0.25, 0.5 1, 2, 4, 8, and 12.5 kHz. Based on the hearing thresholds and the ear canal sound pressures obtained in the previous measurement, the ear canal sound pressures at thresholds were estimated with the assumption of linearity. After determining the AC thresholds, the modified ER3 insert was removed from the left ear while the modified ER3 insert in the right ear remained and was used to provide masking during the BC testing. The masking level was set as 30 dB above the BC stimulation level throughout the testing.
A foam earplug (3M™ E-A-R™ Classic™, 3M, MN) was inserted approximately 15 mm into the ear canal and the occluded ipsilateral and contralateral ear canal sound pressures with BC stimulation were obtained in sequence. After that, the foam earplug was removed, and the open ear canal sound pressure with BC stimulation was measured for ipsilateral and contralateral stimulation. Keeping the left ear canal open while masking the right ear canal with noise using the modified ER3 insert, the BC thresholds using the 3AFC procedure previously described was obtained with both ipsilateral (left side) and contralateral (right side) stimulation at the frequencies 0.25, 0.315, 0.4, 0.5, 0.63, 0.8, 1, 1.25, 1.6, 2, 2.5, 3.15, 4, 5, 6.3, 8, 10, and 12.5 kHz. The entire testing time was 2 to 3 h including short breaks when requested.

2.5. Calibrations

The BC transducer calibration was conducted on a Brüel & Kjær Artificial Mastoid Type 4930. The dynamic output forces of the BC transducers as estimated on the artificial mastoid with 1-volt stimulation to the transducers are shown in Fig. 3A. Even if the transducer is based on the Radioear B81 motor unit, the output sensitivity curve is slightly different from the normal B81 curves since the influence from the housing is removed in the current design, and there is no resonance peak in the 4 kHz region. The impedance of an artificial mastoid is only specified for frequencies between 125 Hz and 8 kHz (IEC 60318-6, 2007), but the frequency range used in the current study extends to 12.5 kHz. The force sensitivity of the Brüel & Kjær Artificial Mastoid Type 4930 was obtained by providing vibrations to a Brüel & Kjær Impedance Head type 8000 that was applied to the artificial mastoid. The force sensitivity for a vibration transducer coupled to the artificial mastoid is obtained by measurement of the impedance head force gauge and acceleration signals and the output from the artificial mastoid simultaneously (Scott et al., 2015). The output force of the BC transducers used in this study was estimated based on the force sensitivity of the artificial mastoid. This means that the estimated stimulation force of the BC transducers at the highest frequencies can be biased due to impedance differences between the artificial mastoid and a human mastoid. However, the impedance of the artificial mastoid given in IEC 60318-6 (2007) is up to 5 dB higher compared to the human mastoid (Flottorp and Solberg, 1976), and the deviation in estimated stimulation force at the highest frequencies is most likely in the same order as those at frequencies specified in IEC 60318-6 (2007) (see appendix).

The harmonic distortion of the BC transducers was also investigated. The distortion of the Radioear B81 transducer is most problematic at low frequencies and according to Eichenauer et al. (2014) the distortion of the 2nd harmonic of a 250 Hz tone is approximately 1% at 45 dB HL. The 2nd and 3rd harmonic distortions at a stimulation level of 40 dB HL are shown in Fig. 3B for the current BC transducers. The 2nd harmonic distortion was close to -40 dB (1%) at frequencies up to 400 Hz, above which it falls down to -80 dB at 2 kHz. The third harmonic distortion was close to -30 dB (3%) at frequencies up to 200 Hz above which it falls to -80 dB at 1 kHz. At frequencies above 2 kHz, the distortion measurements were below the noise floor of the measurement system, and the curves indicate the noise level in relation to a 40 dB HL stimulation. Consequently, the distortion levels were below the curves in Fig. 3B at frequencies above 2 kHz.

The sensitivities of the ER7C probe tube microphones were calibrated by placing the opening of the probe tube and a Brüel & Kjær type 4189 microphone in a Brüel & Kjær anechoic test box type 4322. The position of the probe tube opening was within 2 mm of the Brüel & Kjær microphone and the distance between their tips and the inbuilt speaker of the box was approximately 50 mm. By playing sound through the speaker in the frequency range

![Fig. 3](https://via.placeholder.com/150)

**Fig. 3.** (A) The sensitivity of the three transducers used in the current study in terms of force levels, as dB FL (dB re 1μN), per input voltage. The output forces of the transducers are estimated by measurements on the artificial mastoid Brüel & Kjær type 4930 with approximately a static force of 3 N. (B) The second and third harmonic distortion component in relation to the fundamental when the stimulation is 40 dB above threshold. (C) The sound pressure level difference between the sound pressure from the BC transducer applied at the artificial mastoid and the ear canal sound pressure when the BC transducer is positioned at the ipsilateral mastoid (blue line). The red line shows the same difference when adjusted for the effect of the ear canal and pinna based on the data in Shaw (1974).
between 0.1 and 12.5 kHz, the sensitivity of the ER7C microphone was obtained based on the measurements obtained from the Brüel & Kjær type 4189 microphone.

2.6. Transducer airborne sound radiation

The output force of the BC transducer is generated by vibration of the transducer’s mass causing a reactive force. The motion of the transducer mass itself results in an airborne sound around the transducer that is transmitted to the open ear canal. Consequently, this is a secondary sound pathway generating an ear canal sound pressure that does not involve the BC pathway in the skull. To investigate if this airborne sound influenced the ear canal sound pressure measurements in the current study, the sound pressure generated by the transducer in the air was compared with the ear canal sound pressure measured in the ear canal.

The ear canal sound pressure obtained in the ear canal with ipsilateral BC stimulation in the participants as described above was used as reference. The sound radiated in the air from the BC transducer was assessed by placing the BC transducer on the artificial mastoid and placing the ER7C probe microphone at the approximate position of the ear canal opening. This sound pressure would be similar to the airborne sound from the transducer alone at the ear canal opening. The relation between the transducer radiated sound at the ear canal opening and the ear canal sound pressure is shown as the blue line in Fig. 3C. This analysis indicates that at frequencies below 1 kHz, the radiated sound is 15 to 30 dB below that obtained in the ear canal. This difference is reduced at higher frequencies and at frequencies of 6 kHz and above, the sound radiated in the air from the transducer is 1 to 2 dB below the ear canal sound pressure. However, the ear itself influences the sound pressure, both the ear canal (mainly resonances) and the fact that the BC transducer radiation is behind the pinna (high-frequency attenuation) changes the sound from the BC transducer to the eardrum. To adjust for these two effects, data from Shaw (1974) was used. The red line in Fig. 3C shows the relation between the transducer-radiated sound pressure and the ear canal sound pressure when adjusted for the effect of the ear canal (ear canal opening to the eardrum) and pinna (behind the ear relative to infront of the ear) in the radiated sound pathway. This result indicates that the effect of transducer radiation can affect the ipsilateral ear canal sound pressure at frequencies of 10 kHz and above.

2.7. Statistics

Most of the data had a non-normal distribution, and differences were investigated with non-parametric tests. For comparisons with average or median data (for example data provided in standards), one-sample Wilcoxon signed ranks test was used, while differences between AC and BC data obtained here were analysed with Wilcoxon signed ranks test. A p-value of 0.05 or less was considered significant and no correction for multiple tests was applied. All statistical analyses were conducted in IBM SPSS statistics ver 26.

3. Results

3.1. Air conduction stimulation

The sound pressure level in the ear canal at hearing thresholds with AC stimulation is presented in Fig. 4. The individual thresholds are shown with thin black lines while the median results of all participants’ thresholds are given as a thick blue line. Most individual thresholds are within 10 dB of the median. Also included in Fig. 4 are the ear canal sound pressures at thresholds for people with normal hearing given in Killion (1978) known as the minimal audible pressure (MAP, red line), and the threshold levels given in ISO 389-2 (1994) for an IEC 711 coupler.

The participants’ median hearing thresholds are within 5 dB of the MAP and ISO 389-2 normal data indicating that their hearing status is intact. The deviation for a few individuals at a few frequencies is not expected to affect the results in the current study. In addition, a one-sample Wilcoxon signed ranks test showed that the differences between the current participants and the MAP were only significant at 2 kHz (p = 0.022) and 12.5 kHz (p < 0.001).

3.2. Bone conduction stimulation

The individual and the median sound pressure levels in the ear canal when a BC stimulation of 1 N is applied on the mastoid behind the ear are shown in Fig. 5. The ear canal sound pressure with stimulation on the ipsilateral side is shown in Fig. 5A and with stimulation on the contralateral side is shown in Fig. 5B. The ipsilateral individual sound pressure levels are mainly within 10 dB of the median result at frequencies above 200 Hz, even if some data points fall outside this range, especially for sharp dips in the sound pressure. At the lowest frequencies, the ipsilateral variability is mainly in the 10 to 20 dB range with a few participants having results 30 dB below the median. The overall tendencies of the results are similar for ipsilateral (Fig. 5A) and contralateral (Fig. 5B) stimulation, even if the absolute values differ. The individual variability is slightly larger in the contralateral sound pressure levels compared with the ipsilateral results, especially at the highest frequencies.

Also included in Fig. 5A is the average ear canal sound pressure with ipsilateral BC stimulation obtained in cadaver heads (Stenfelt et al., 2003), where the stimulation has been adjusted from excitation at the skull bone to excitation at the skin according to Stenfelt (2006). The median sound pressure levels in the current study correspond well with the mean sound pressures from the cadaver heads except at the lowest frequencies, in the range below 400 Hz. At these low frequencies, the median ear canal sound pressures obtained here are around 20 dB greater than the sound pressure levels reported in Stenfelt et al. (2003). This deviation is partly a result of the force calibration of the BC transducer, where the mechanical load impedance of the human mastoid and the artificial mastoid differ (see appendix). On the human, the mechanc-
ical point impedance at the mastoid position is 1 to 5 dB lower (softer) than the impedance of the Brüel & Kjær artificial mastoid (Flottorp and Solberg, 1976; IEC 60318-6, 2007; Stenfelt and Häkansson, 1998). Such load difference affects the transducer’s resonance frequency at around 500 Hz (Fig. 3) shifting the resonance to a lower frequency thereby increasing the force output at frequencies below the resonance frequency and decreasing the force output at frequencies above the resonance frequency. This shift resulted in up to 10 dB difference between low and high frequencies in transducer output based on the simulation in the appendix.

The ear canal sound pressures with a BC stimulation of 1 N at the mastoid were obtained both with open ears and with a foam earplug inserted approximately 15 mm into the ear canal. The difference between the occluded and open sound pressure levels is often termed the objective occlusion effect. The objective occlusion effect is shown for the current participants in Fig. 6A with ipsilateral BC stimulation and in Fig. 6B with contralateral BC stimulation. The median results with stimulation at both sides are nearly identical with an increase of sound pressure with occlusion at frequencies below 700 Hz where the sound pressure increases at lower frequencies with a slope of approximately -30 dB/decade.

Also included in Fig. 6 is a model prediction of the ear canal sound pressure occlusion effect for an occlusion device 15 mm into the ear canal (red curve, (Stenfelt and Reinfeldt, 2007)). The model prediction is within a couple of dBs from the median objective occlusion effect values for both stimulation sides except at frequencies between 2.5 and 7 kHz for contralateral stimulation and above 8 kHz for ipsilateral stimulation. Here, differences of 5 to 10 dB between the model prediction and the measured median objective occlusion effects are found. One explanation for the contralateral difference between the model prediction and the current measured occlusion effect is that a contralateral BC stimulation excites the soft tissue to a lesser extent than ipsilateral stimulation and the reduction of sound pressure due to an earplug in the soft tissue part of the ear canal is less pronounced. The high-frequency deviation with ipsilateral stimulation can be a result of airborne sound radiation from the transducer influencing the high-frequency open ear canal sound pressure.

The hearing thresholds with ipsilateral BC stimulation are given in Fig. 7A. The thresholds are in force levels (dB re 1μN) according
to the calibration on the artificial mastoid. Most of the individual results are within 10 dB of the median threshold, but a few participants had BC hearing thresholds that deviated more. Two of the participants had a mid-high frequency threshold elevation that was also seen in the AC thresholds (Fig. 4). Compared to the reference equivalent threshold force levels (RETFL, the red curve in Fig. 7A) given in (ISO 389-3, 2016), the current median thresholds are 10 to 20 dB worse. Based on a one-sample Wilcoxon signed ranks test, all differences between the current thresholds and the RETFL were statistically significant (p<0.001) except for the two lowest frequencies, 250 and 315 Hz. The AC thresholds (Fig. 4) indicated nearly perfect hearing of the participants and the BC thresholds were expected to be close to the RETFL values. The origins of the deviations from the RETFL are multiple and include the impedance mismatch related calibration error described above as well as differences between the Radioear B71 (used to obtain the RETFL) and the transducers used here. An additional difference between the AC and BC thresholds is that the BC thresholds were obtained with contralateral masking which, due to central effects, elevates the thresholds by up to 10 dB (McDermott et al., 1990; Snyder, 1973a, 1973b; Zwislocki, 1972).

The hearing thresholds with contralateral BC stimulation are presented in Fig. 7B. The median threshold with ipsilateral stimulation also included in the graph illustrating that the contralateral stimulated thresholds are 5 to 15 dB worse compared to the ipsilateral thresholds with the smallest difference at around 1 kHz and the greatest difference at frequencies between 3 and 8 kHz.

The median threshold differences between AC ear canal sound pressure and MAP and the ipsilateral BC force levels and RETFL are shown in Fig. 8. The AC thresholds are close to those proposed by the MAP and are primarily within 5 dB. The deviation is greater at 12.5 kHz, the highest frequency tested. This deviation may, at least partly, originate in difficulties measuring the correct ear canal sound pressure at high frequencies where small changes in the position of the probe tube microphone can alter the measured sound pressure level significantly. At 12.5 kHz, a quarter wavelength is approximately 7 mm and the measurement position is within a few millimeters from an expected null in the ear canal sound pressure. As already indicated in Fig. 7A, the deviation between the BC thresholds and the RETFL are greater than between the AC thresholds and MAP, and is primarily in the 10 to 20 dB range (red line in Fig. 8). The difference between AC ear canal sound pressure thresholds and BC force level thresholds (difference between blue and red curves) are shown in the black curve in Fig. 8. This computation indicates that the BC thresholds are 10 to 20 dB off. As discussed previously, the difference is attributed to the calibration offset by the artificial mastoid and the effect of contralateral masking present for BC but not for AC stimulation. Computations of the effect from the calibration on the artificial mastoid indicate that BC threshold force levels presented in the current study should be increased by up to 5 dB below 500 Hz and decreased by 2 to 5 dB at frequencies above 500 Hz. That means that the effect of contralateral masking worsens the BC sensitivity by 5 to 15 dB which is in line with data in the literature.

The ear canal sound pressures at BC thresholds were computed from the open ear canal sound pressure measurements with BC stimulation (Fig. 5) and the BC hearing threshold measurements (Fig. 7) with the assumption of linearity. Based on these computations, the ear canal sound pressures at BC hearing thresholds are shown with BC ipsilateral and contralateral stimulation in Fig. 9A and B, respectively. With ipsilateral BC stimulation (Fig. 9A), the
median ear canal sound pressure is between 10 and 25 dB SPL at thresholds. The median sound pressure at thresholds with BC ipsilateral stimulation is close to, or slightly greater than, the median sound pressures obtained at thresholds with AC stimulation (red curve in Fig. 9) at frequencies below 6 kHz. At higher frequencies, the median BC-generated sound pressure levels are lower than the median AC sound pressure at thresholds with a difference of 25 dB at 12.5 kHz. It should be noted that the AC thresholds are obtained at fewer frequencies than the BC thresholds, and the lines indicating the thresholds between test frequencies are linear interpolations.

The sound pressures at thresholds when the BC stimulation was at the contralateral mastoid are shown in Fig. 9B together with the BC ipsilateral median sound pressures and the median AC sound pressures at thresholds. The median BC generated ear canal sound pressures at thresholds are nearly identical at frequencies below 8 kHz irrespective of stimulation side. This indicates that the vibration at the ear canal wall that generates the ear canal sound pressure and the vibration at the inner ear that is assumed responsible for the BC perception is influenced equally by a change in BC stimulation position. The small deviation at the highest frequencies, amounting to 10 dB, can be from airborne sound radiation of the BC transducer which influences the ear canal sound pressure from the ipsilateral but not from the contralateral BC transducer (Fig. 3C).

Fig. 10 shows the difference between ipsilateral stimulated BC ear canal sound pressures at threshold and AC stimulated ear canal sound pressures at thresholds. At frequencies between 250 and 500 Hz, the median BC-generated sound pressures are approximately 5 dB greater than the median AC-generated sound pressures while at frequencies above 500 Hz and up to 2 kHz the two modalities result in median ear canal sound pressures at thresholds that are nearly identical. At frequencies between 2.5 and 4 kHz, the median BC-generated ear canal sound pressures are again around 5 dB greater than the AC generated medians. At frequencies above 4 kHz, the median difference between the two ear canal sound pressures increases with AC-generated sound pressures at thresholds as the dominant. At the highest frequency, 12.5 kHz, median AC-generated ear canal sound pressure is 25 dB greater than median BC-generated sound pressure. A Wilcoxon signed ranks test was applied to analyze the differences at the frequencies where both AC and BC thresholds were measured. This showed that it was only at 4 kHz \((p = 0.014)\) and 12.5 kHz \((p < 0.001)\) that the differences were statistically significant. Again, it should be noted that AC thresholds were only obtained at octave frequencies while the BC thresholds were obtained at 3rd octave frequencies and the in-between test-frequency results are based on linear interpolation.

The hearing thresholds in the current study facilitated computation of the interaural separation also termed transcranial attenuation. It is here done in two ways, both as the difference between the force threshold levels for ipsilateral and contralateral BC stimulation, and as the difference in ipsilateral and contralateral ear canal sound pressures. The differences between ipsilateral and contralateral stimulated BC thresholds are shown in Fig. 11A. As reported in several other studies on transcranial attenuation, the individual variability is large with around 40 dB variation at a single frequency. The median results show 5 to 10 dB attenuation at frequencies up to 800 Hz, close to 0 dB at 1 kHz that increases with frequency up to 20 dB at 8 kHz, and a slight decrease at the highest frequencies tested drop-
est frequencies that decreases to around 0 dB at 1 kHz and increases to around 20 dB at the highest frequencies is similar to the force threshold data from Fig. 11A; these data are also included in Fig. 11B. Transcranial attenuation from Reinfeldt et al. (2013) shows less attenuation at the lowest frequencies but greater attenuation at the highest frequencies.

4. Discussion

4.1. Importance of the ear canal sound pressure for bone-conducted sound

The aim of the current study was to investigate the importance of the ear canal sound pressure for hearing BC sounds. This pathway has been suggested as important in some studies (Huizing, 1960; Khanna et al., 1976) but also been suggested to be of low importance to BC hearing in a normal ear in others (Stenfelt, 2016, 2020). The methodology here was to investigate the sound pressure level in the ear canal generated by BC stimulation at hearing thresholds and compare that with sound pressure levels generated by AC stimulation at hearing thresholds. The rationale for this method is that it is here assumed that the AC sound stimulates the ear through the ear canal alone, and the sound pressure level in the ear canal at the hearing threshold is the lowest level able to excite a hearing response. Consequently, when comparing the ear canal sound pressure levels generated by BC stimulation at threshold with those generated by AC at threshold, the difference in terms of how much lower the ear canal sound pressure is by BC compared to AC is an estimation of how much lower the outer ear pathway is compared to other contributors for BC hearing. This comparison is done in Fig. 10 where the BC generated ear canal sound pressure is equal or greater than the AC generated ear canal sound pressure at threshold for frequencies up to 4 kHz. According to our assumption, the BC-generated ear canal sound pressure at threshold should not be able to be higher than the AC threshold ear canal sound pressure. However, the same finding of larger ear canal sound pressure at threshold for BC than for AC stimulation was reported by Huizing (1960) at frequencies below 500 Hz.

In the current study, the possibility of a larger ear canal sound pressure at threshold for BC than AC stimulation is attributed to the use of contralateral masking that was used for BC but not for AC threshold testing. Contralateral masking was used to ensure that the response was from the left ear (right ear was masked) when BC stimulation was provided, but it also results in up to 10 dB elevated thresholds. It is not exactly known how much the BC thresholds were elevated due to the masking procedure in the current study, but one estimate can be obtained from Fig. 8 that shows the difference between the AC and BC thresholds based on MAP and RETFL. The AC thresholds are close to the MAP which seems reasonable based on the participants with a mean age of 33 years and no self-reported hearing problems. The only deviation of the median AC thresholds and the MAP was found at the highest frequency tested, 12.5 kHz, where a statistically significant higher sound pressure level at threshold was seen in the current study compared to the MAP. This deviation can partly be explained by a larger uncertainty of the sound pressure measurement at these high frequencies due to a quarter wavelength minima close to the measurement position. But the explanation can also be that the participants had a general worse hearing at the highest frequency tested.

The BC thresholds deviate more in relation to the RETFL than the AC thresholds in relation to the MAP. The RETFL curve used is for a B71 BC transducer on the mastoid, and the current study used a vibration unit from the Radioear B81. It has been shown that the Radioear B81 matches the Radioear B71 sensitivity (Jansson et al., 2015). The stimulation interface here was the
same as for the Radioear B71 and B81 (a circular plate with a diameter of 15 mm) but the stimulation position was at the mastoid around 2 cm superior of the usual Radioear B71 position. According to Dobrow et al. (2016), the sensitivity for BC stimulation is within 1 to 2 dB between the typical B71 position and the position used here, so the effect from using a different stimulation position is negligible.

The major difference between the BC transducer used in the current study (Fig. 1) and the Radioear B71/B81 transducers is the housing which gives a resonance at around 4 kHz that is not present in the current transducer (Fig. 3). Consequently, the deviation from the RETFL due to the current BC transducers should mainly appear at frequencies around 4 kHz and the greatest deviation between the current BC thresholds and the RETFL is found at 4 kHz. The issue related to the calibration of BC transducers on an artificial mastoid where the impedance mismatch between the artificial mastoid and the human mastoid changed the vibration force output from the transducer due to a shift in the transducer’s resonance frequency (here at 500 Hz), is similar for the current transducer and the Radioear B71/B81 transducers. This would therefore not affect the current BC thresholds in relation to the RETFL, and should not significantly affect the AC-BC difference curve in Fig. 8. However, the BC re RETFL curve in Fig. 8 indicates a different response above and below 500 Hz, and it is plausible that the resonance at 500 Hz in the current BC transducer (Fig. 3A) is influenced differently than a Radioear B71 transducer between application to a human mastoid and an artificial mastoid used for calibration (see appendix).

Based on the above, the results in Fig. 8 suggest that the effect of contralateral masking worsens the BC thresholds 5 to 15 dB. This means that for a correct comparison between AC and BC generated ear canal sound pressure at threshold, the difference curve in Fig. 10 should be decreased 5 to 15 dB to interpret the influence of the ear canal sound pressure for hearing BC sound. This is done in Fig. 12 where the estimated contribution from the ear canal sound pressure to BC hearing is computed by adjusting the difference in Fig. 10 by 5 and 10 dB, and the real contribution most probably falls between these two lines. A Wilcoxon signed ranks test were conducted on the shifted data and its results are shown at the top of Fig. 12, where the red and blue colors relate to the red and blue curves, and a filled square indicates a significance level of \( p < 0.01 \) and an unfilled square a significance level of \( p < 0.05 \). Consequently, with a 5 dB shift (blue curve) it is nearly only at the highest frequencies (\( f \geq 6.3 \) kHz) that the differences are significant, while with a 10 dB shift (red curve) nearly all frequencies (except 315 and 400 Hz) show a significant difference.

According to such modification, the interpretation is that the ear canal sound pressure is similar to other contributors of, or even dominating, the BC hearing at frequencies between 250 and 500 Hz as well as in the 2.5 to 4 kHz region, while the ear canal sound pressure is 5 to 10 dB below other contributors at frequencies between 500 Hz and 2 kHz. At higher frequencies, above 4 kHz, the ear canal sound pressure during BC stimulation seems to have less importance for perceiving BC sound, and its importance declines with frequency indicating around 30 dB below other contributors at 12.5 kHz. This result is based on the measurements with stimulation at the ipsilateral mastoid. However, Fig. 9B shows nearly identical results of ear canal sound pressures at thresholds for contralateral as for ipsilateral BC stimulation. Consequently, the importance of the ear canal pathway for BC hearing is similar for ipsilateral as for contralateral BC stimulation.

The similarities between ipsilateral and contralateral results indicate reliability and reproducibility of the current data. The similarities in the occlusion effect for ipsilateral and contralateral BC stimulation with the model predictions in Stenfelt and Reinfeldt (2007) show that the origin for the occlusion effect is primarily the change in radiation impedance of the ear canal opening. The similarity of the median ear canal sound pressure at threshold for ipsilateral and contralateral BC stimulation (Fig. 9B) indicates that, even if there is a great spread in individual data, the median results are stable and can be used to investigate the small differences found in the current study.

4.2. Transducer radiation of air-borne sound

The similarities between the AC and BC ear canal sound pressure levels at thresholds could be a result of airborne sound radiation from the BC transducer itself, and not caused by the vibrations of the skull. It was here investigated in Fig. 3C showing that it is only at the highest frequencies that the BC transducer airborne radiation can affect the measurements. An additional analysis is the inspection of the occlusion effects in Fig. 6 where the median occlusion effects corresponded well to the model predictions. Also, there were similarities between the ipsilateral and contralateral stimulated occlusion effects ruling out an effect of sound radiation from the transducer with ipsilateral stimulation. There are discrepancies at frequencies above 7 kHz that can be a result of transducer radiation at the highest frequencies, but it could also be an effect of inaccuracies in the model predictions at the highest frequencies (Stenfelt and Reinfeldt, 2007). However, there is also a deviation between the ipsilateral and contralateral stimulated ear canal sound pressure at threshold in Figs. 9B at 10 and 12.5 kHz, indicating that transducer sound radiation can have affected the measurements at the highest frequencies. As the ear canal sound pressure did not influence the perception of BC sound at those high frequencies (above 7 kHz), the airborne radiation from the transducer is not believed to influence the analysis of the ear canal as a contributor to BC hearing.

4.3. Comparison to previous estimations

The current findings indicate that the ear canal sound pressure with BC stimulation is 0 to 10 dB below other contributors for
hearing BC sound in the frequency range 0.25 to 4 kHz. At frequencies above 4 kHz, the influence from the BC-generated ear canal sound pressure on the perceived BC sound diminishes and becomes approximately 30 dB below other contributors for BC sound at 12.5 kHz. This is illustrated by the thick red and blue lines in Fig. 12. Also included in Fig. 12 are the estimates of the importance of the BC ear canal sound pressure on the perceived BC sound from other studies. In the study of BC generated ear canal sound pressure in cadaver heads in Stenfelt et al. (2003) the umbo velocity was related to the ear canal sound pressure caused by AC as well as BC sound (Fig. 10 in that study). If the umbo velocity with AC stimulation is used as a reference, the BC-generated sound pressure at the same umbo velocity can be used as an estimation of the importance of the BC-generated ear canal sound pressure. This is shown as the magenta line in Fig. 12. This estimate is close to the 10 dB adjusted curve (thick red line) at frequencies between 0.5 and 8 kHz. At 250 Hz, this estimate suggests the ear canal sound pressure to be 15 dB below other contributors for BC sound, which is significantly less than that found in the current study. At the highest frequencies, the estimate based on the Stenfelt et al. (2003) study indicates less decline in the ear canal sound pressure contribution but is still close to the values suggested by the 5 dB adjusted curve. The caveat in this analysis is that the umbo velocity is not the same as the cochlear excitation for BC stimulation, and Stenfelt (2006) suggested that the umbo velocity for BC stimulation is around 5 dB below the cochlear excitation at frequencies below 3 kHz. According to that, the magenta line should be down-shifted approximately 5 dB in Fig. 12.

In the current study and in the study of Stenfelt et al. (2003), there is an increase in the ear canal sound contribution at 2.5 kHz. This increase originates in the resonance of the open ear canal at around this frequency. Consequently, for BC sound in the ear canal, the resonance increases the sound pressure induced by the vibrations of the ear canal walls, and the ear canal sound pressure becomes nearer to other contributors. The opposite can be seen at around 5 kHz where there is a small dip in the estimated ear canal sound pressure contribution due to the anti-resonance at this frequency.

Another estimate of the ear canal contribution with BC sound comes from Röösli et al. (2012), shown as a cyan line in Fig. 12. This estimation is based on the ear canal sound pressure levels at thresholds provided in that study and adjusted for the air-bone gaps given. Based on these computations, Röösli et al. (2012) suggest the BC generated ear canal sound pressure to be 10 to 15 dB below other contributors for BC hearing at frequencies up to 2 kHz. Above this frequency, the contribution from the BC generated ear canal pressure increases with frequency and peaks with 10 dB above the other contributions at 4 kHz. This is a violation of the assumptions in the current study where the contribution cannot be higher than 0 dB. Based on the information provided in the study by Röösli et al. (2012), we cannot find a good explanation for this discrepancy.

Khanna et al. (1976) made measurements of the sound pressure at the ear canal entrance when a non-occluding box was covering the ear. This measurement set-up was not causing an occlusion effect while attenuating airborne sound, the same technique was used by Reinfeldt et al. (2010). In the Khanna et al. (1976) study, cancellation of AC and BC sound was accomplished in the frequency range between 500 Hz and 6 kHz when the stimulation was at the forehead through a 16 mm circular interface. The AC and BC generated sound at cancellation was reported at the ear canal entrance (Fig 22A in that study) and is here included as a dashed black line in Fig. 12. The results suggest that the ear canal sound pressure is important for BC hearing at frequencies below 800 Hz but is 15 to 20 dB below other contributors for BC hearing at frequencies above 1 kHz. The caveat is that the sound pressure at the ear canal entrance is used in the Khanna et al study and how that relates to the sound pressure at the eardrum for BC stimulation is unknown. Moreover, the forehead was used in the Khanna et al study while the mastoid was used for BC stimulation in the present study. Consequently, the results from the Khanna et al study should be interpreted with care in relation to the current results. Even so, the predictions from Khanna et al. (1976) at frequencies above 1 kHz follows the predictions in the current study but around 5 dB lower than the 10 dB adjusted estimates.

The two other estimates of the ear canal contribution to BC hearing are from the model predictions in Stenfelt (2016) (blue dashed line) and the updated version of the model ((Stenfelt, 2020), red dashed line). These model predictions are based on an impedance model of the inner ear, BC wave propagation, and experimental data from the literature (see Stenfelt (2016) for more details). The ear canal component predictions in these models are based on the ear canal sound pressure measurements in Stenfelt et al. (2003) presented with the magenta line in Fig. 12. The predictions from the model suggest that the ear canal component is 10 to 20 dB below other contributors at frequencies up to 2.5 kHz above which fall down to ~40 dB at 5 kHz and above. Compared to the other estimates, the models under-estimates the contribution from the BC ear canal sound pressure by 10 to 20 dB. This was unexpected since the model predicts inner ear fluid sound pressure from both AC and BC excitation in accordance with experimental data, and predicts sensitivity changes from ear pathologies in line with clinical results (Stenfelt, 2016, 2020). However, the current results indicate that the model needs revision, at least for the prediction of the BC contribution from the ear canal sound pressure.

Several studies have reported differences in the BC-generated occlusion effect when estimated by the sound pressure change in the ear canal and when it is estimated by threshold elevations (Goldstein and Hayes, 1971; Huizing, 1960; Reinfeldt et al., 2013). This difference can be used as an estimation of the contribution of the BC-generated ear canal sound pressure for hearing BC sound at low frequencies. For example, the dataset from Reinfeldt et al. (2013) suggest the ear canal sound pressure to become more important with frequency from around ~10 dB at 250 Hz to dominate the BC perception at 1 kHz. This is different from the estimates in Fig. 12 that suggest that the ear canal sound pressure is 5 to 20 dB below other contributors at 1 kHz. One possible explanation for this finding is that low-level noise was present during the threshold estimation that slightly masked the open-ear thresholds but not the occluded thresholds. However, the noise spectrum of the test room was not provided in the Reinfeldt et al. (2013) study and this is merely a possible explanation.

To summarise, the current and previous estimates of the contribution of the ear canal sound pressure on BC hearing suggest that the contribution is around 10 dB below other contributors at frequencies at and below 2 kHz, is close to other contributors at around 2.5 kHz due to the ear canal resonance, and becomes less important at frequencies of 3 kHz and above falling with frequency.

4.4. Transcranial attenuation

The transcranial attenuation, i.e. the difference in level between the ipsilateral and contralateral side of the head was estimated both by hearing thresholds (Fig. 11A) and by sound pressure in the ear canal (Fig. 11B). If the BC sound influences the vibration of the bone encapsulating the inner ear similar to the bone at the ear canal, the transcranial attenuation should be similar for hearing thresholds and for ear canal sound pressure. According to Fig. 11B, where both medians for estimating the transcranial attenuation are shown, this seems to be the case. Both median curves
have the same morphology and are at most frequencies within 5 dB. The threshold-based transcranial attenuation (Fig. 11A) is about 5 dB higher than that reported by Reinfeldt et al. (2013) and up to 10 dB higher than that reported by Stenfelt (2012). The measurement in Stenfelt (2012) was conducted on participants with unilateral deafness where several had undergone ear surgery. That means that the stimulation position was slightly altered to fit on the head and the surgery itself can impact the BC sound sensitivity (Prodanovic and Stenfelt, 2020). The setups here and in the Reinfeldt et al. (2013) study were similar but used different stimulation positions, BC transducers, and threshold estimation methods. These differences can, at least partly, be responsible for the slight differences in transcranial attenuation found in Fig. 11A. Even if the median transcranial levels appear higher in the current study compared to the Stenfelt (2012) and the Reinfeldt et al. (2013) data, other reports of threshold-based transcranial attenuation are similar to what is reported here. For example, Nolan and Lyon (1981) reported average transcranial attenuation that was around 10 dB for the entire frequency range between 250 Hz and 4 kHz, and Snyder, (1973) reported transcranial attenuation that was on average 8 dB at frequencies below 1 kHz and 11 to 13 dB in the 2 to 4 kHz frequency range. Moreover, the wide spread in individual transcranial attenuations is similar in range to that reported by others.

4.5. Clinical implications

The focus of the current study is on the importance of the ear canal sound pressure for the perception of BC sound. Even if it is important to understand the mechanisms generating BC sound perception, this may be regarded as clinically irrelevant. However, the current study has revealed topics of clinical importance. One such topic is BC stimulation at frequencies up to 12.5 kHz. Normally, BC threshold testing with the Radioear B71/B81 is limited to frequencies of 6 kHz and below. The reason for this limitation is the fast decline in output level with frequency at frequencies above 4 kHz (see Fröhlich et al. (2018)). Compared to the Radioear B71/B81 BC transducer, the BC transducer in the current study provided 20 dB more output at the high frequencies facilitating hearing testing well above 10 kHz. This can seem odd since the BC transducer here used the same motor unit as in the Radioear B81 transducer as well as the same stimulation interface area. The difference is that the current transducer design (Figs. 1 and 3) applied the vibrations directly to the skin whereas, in the Radioear B71/B81, the motor unit is attached to the backside of the housing thereby creating a resonance at around 4 kHz. As a consequence, the output sensitivity of the Radioear declines fast at high frequencies. One solution to this issue would be to flip the Radioear B71/B81 transducer and excite the backside. Unfortunately, this would also result in a different interface area and the transducer does not conform with the standard for BC testing. Others have also shown the possibility to provide high-frequency stimulation through a BC transducer facilitating hearing testing at frequencies up to 16 kHz (Popelka et al., 2010).

There have been suggestions that the threshold level for BC testing at 4 kHz given in the standard (ISO 389-3, 2016) is erroneous leading to false air-bone gaps (Margolis et al., 2013; Margolis et al., 2016). When investigating the offset between AC and BC thresholds in Fig. 8, this was not seen here but the differences, believed to be caused by contralateral masking during BC testing, are similar at frequencies between 0.5 and 4 kHz, except at 1.3 kHz. Consequently, the results here do not suggest a general problem with the threshold reference value at 4 kHz. The caveat is that the current testing was not done with a Radioear B71/B81 transducer that is specified in the standard and that a slightly different threshold estimation procedure is used compared to the conventional ascending method. However, the testing procedure was automatic and the lack of a specific air-bone gap at 4 kHz cannot be explained by tester-bias. As the reported error appears at 4 kHz, where the Radioear B71/B81 transducer has a resonance, it is likely an effect of calibration and differences between artificial mastoid impedance and human mastoid impedance, resulting in a resonance shift and altered output levels.

Such an effect of resonance shift on the BC transducer was seen at low frequencies in the current study. The BC transducer used in the current study has a resonance at around 0.5 kHz (Fig. 3). Due to the difference in mechanical point impedance between the human mastoid and the Brüel & Kjær artificial mastoid that was used for calibrations, the resonance frequency of the BC transducer is shifted to lower frequencies when applied to the human compared to the artificial mastoid (see appendix). This downshift in resonance frequency leads to higher output levels below the resonance frequency and lower output levels above the resonance frequency. According to computations of the transducer output for the two loading conditions (artificial mastoid and human mastoid) similar to computations done in Håkansson et al. (2020) and Chang and Stenfelt (2019), the output level on the human was up to 5 dB higher at frequencies below 500 Hz and 2 to 5 dB lower at frequencies above 500 Hz when the transducer was applied to the human compared to the artificial mastoid. This partly explains the difference in the ear canal sound pressure levels between the current study and the levels from Stenfelt et al. (2003) shown in Fig. 5. Consequently, there are differences in the output of BC transducers applied to humans and that obtained during calibration, and the levels can differ between calibration devices even if they conform to impedance levels specified in the standard (IEC 60318-6, 2007; Pollard et al., 2013; Stenfelt and Håkansson, 1998).

5. Conclusions

Based on ear canal sound pressure measurements with AC and BC stimulation, the importance of the outer ear pathway for hearing BC sound was estimated. Accordingly, the outer ear pathway was estimated to be within 10 dB of other contributors for BC hearing at frequencies up to 2 kHz. At frequencies between 2 and 4 kHz, the contribution from the ear canal sound pressure for hearing BC sound was estimated to be similar to other important contributors in this frequency range. A reason for the apparent importance in the 2 to 4 kHz frequency range is the resonance for the open ear canal in this frequency region. At frequencies above 4 kHz, the ear canal pathway seems to have a low influence on hearing BC sound. The current findings were similar when the stimulation was applied to the opposite ear, indicating that the findings were not limited to BC stimulation at a specific position.

The study also revealed problems of estimating the stimulation force from a BC transducer when calibrated on artificial mastoids. Due to impedance mismatch between the standardized artificial mastoid impedance and the impedance of a human mastoid, the BC transducer resonance will shift causing different outputs below and above the resonance frequency.

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Appendix

The effect of BC transducer calibration on an artificial mastoid in relation to the output force generated at the skin covered mastoid on a human was investigated using the method proposed in Flottorp and Solberg (1976) and Håkansson et al. (2020). A lumped-element impedance model (Fig. A1) of the BC transducer used in the current study was derived based on the model in Chang and Stenfelt (2019). The parameter values of the BC transducers modeled in Chang and Stenfelt (2019) differ from the BC transducer modeled in the current study. The output of the transducer (\( F_{\text{Out}} \)) in Fig. A1A is applied to the transducer load (\( Z_{\text{LOAD}} \)), in this case either the artificial mastoid or the skin on the human mastoid. The load impedance of the skin is taken from Flottorp and Solberg (1976) who suggested a RLC series network to simulate the skin impedance (\( m = 0.6 \, \text{g}, \, C = 4.7 \, \mu \text{m/N}, \, R = 20 \, \text{Ns/m} \)). The Brüel & Kjær Artificial Mastoid Type 4930 that was used to calibrate the transducers in the current study consists of a butyl rubber layer on a layer of silicone rubber with a mass in-between, and the entire rubber part rests on a brass mass of 3.5 kg. This was here modeled as two damped mass-spring systems in sequence ending with a 3.5 kg mass (Fig. A1B).

The impedance level of the skin covered mastoid according to Flottorp and Solberg (1976) for a 175 mm\(^2\) circular interface is shown in Fig. A2A together with the modelled artificial mastoid impedance from Fig. A1B and the impedance levels for an artificial mastoid as stated in IEC 60318-6 (2007), Fig. A1A shows that the impedance network and associated parameter values in Fig. A1B predict the impedance stated in IEC 60318-6 (2007) well and the artificial impedance model can be used for the current simulations. Fig. A1A also shows that the impedance level of an artificial mastoid is 1 to 5 dB higher than the average impedance of the skin covered mastoid for a circular interface of 175 mm\(^2\).

Fig. A2B shows the transducer model output together with the output from the BC transducer used in the current study (same as in Fig. 3A). The model predicts the behaviour of the BC transducer within 1 dB except in the 1.7 to 2.4 kHz frequency range, where the model output was around 2 dB below the BC transducer’s output as measured on the artificial mastoid. When the model transducer is loaded with the skin impedance instead of the artificial mastoid impedance, the modeled transducer output changes. The low-frequency resonance frequency with the modeled transducer connected to the artificial mastoid impedance is 490 Hz; this is lowered to 440 Hz when connected to the skin impedance. This resonance shift leads to a higher output force at frequencies below 490 Hz with the modeled transducer on the skin compared to when the modeled transducer is connected to the artificial mastoid. At frequencies above 490 Hz, the output of the modeled transducer loaded with the skin is lower than the output from the modeled transducer loaded with the artificial mastoid.

The difference in output force from the modeled transducer between skin loading and artificial mastoid loading is displayed in Fig. A2C. At 400 Hz, the shift is around 4.5 dB that diminishes at lower frequencies and becomes close to 0 dB at frequencies below 200 Hz. At frequencies between 0.5 and 2.5 kHz, the output on the artificial mastoid is 1 to 2.5 dB higher than that on the skin. This difference increases with frequency at higher frequencies and becomes 5 dB at 10 kHz. Consequently, the model simulations indicate a higher force output at low frequencies when a BC transducer is attached to the skin compared to when it is calibrated on an artificial mastoid, and lower force output at higher frequencies for a BC transducer on the skin compared to the artificial mastoid. This finding was illustrated in Stenfelt and Håkansson (1999) where there was an unexplained 5 to 8 dB improvement in skin applied force thresholds compared to direct skull bone applied force thresholds at 250 and 500 Hz. In that study, the

![Fig. A1.](image)

**Fig. A1.** (A) A lumped-element impedance model of the BC transducer used in the current study. The left-hand side of the model depicts the electrical part of the transducer and the right-hand side depicts the mechanical part of the transducer. (B) An lumped element impedance model of the Brüel & Kjær Artificial Mastoid Type 4930 comprising two damped mass-spring systems simulating the two rubber layers connected to a mass simulating the mass of a human head.
force output from the BC transducer on the skin was estimated by an artificial mastoid and the low frequency deviation is attributed to the mismatch in force output on the skin and on the artificial mastoid.

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