Original paper

Simulating cartilage conduction sound to estimate the sound pressure level in the external auditory canal

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Abstract

When the aural cartilage is made to vibrate it generates sound directly into the external auditory canal which can be clearly heard. Although the concept of cartilage conduction can be applied to various speech communication and music industrial devices (e.g. smartphones, music players and hearing aids), the conductive performance of such devices has not yet been defined because the calibration methods are different from those currently used for air and bone conduction. Thus, the aim of this study was to simulate the cartilage conduction sound (CCS) using a head and torso simulator (HATS) and a model of aural cartilage (polyurethane resin pipe) and compare the results with experimental ones. Using the HATS, we found the simulated CCS at frequencies above 2 kHz corresponded to the average measured CCS from seven subjects. Using a model of skull bone and aural cartilage, we found that the simulated CCS at frequencies lower than 1.5 kHz agreed with the measured CCS. Therefore, a combination of these two methods can be used to estimate the CCS with high accuracy.

Keywords: Aural cartilage, External auditory canal, Calibration, Sound pressure level, Head and torso simulator

1. Introduction

In 2004, Hosoi found that a specific type of transducer, gently placed on the aural cartilage (i.e., entrance of the external auditory canal), could be used to create clear audible sound [1,2]. Aural cartilage is found in the outer ear, and is distributed around the exterior half of the external auditory canal. This cartilage plays the role of a diaphragm, and the transducer functions as the voice coil of a loudspeaker. Bone conduction is a well-known method of sound transmission using vibration that can be achieved by pressing a transducer on a user's mastoid under strong pressure. Unlike through heavy skull bone, conduction through the light aural cartilage is likely to be driven by a slight touch and a powerless transducer. And in contrast to the auditory properties of sound waves that are externally generated in the air, cartilage conduction produces sound directly in the external auditory canal. Thus the concept and term of "cartilage conduction," which is difficult to categorize within the two previously known types of conduction, were proposed as shown in Figure 1b [1-3]. To examine the acoustical characteristics of cartilage conduction, we have developed a cartilage conduction transducer that is fitted on the entrance of the external auditory canal as shown in Figure 1a [4]. The ring-shaped transducer permits the user to hear without occluding the external auditory canal, thus avoiding the occlusion effect and enabling the user to hear unmodified sound via cartilage vibration. In recent studies, we reported that cartilage conduction can generate sound in the canal, particularly at frequencies lower than 3 kHz [5], and that only minimal cartilage conduction sound leaks from the canal regardless of whether the canal remains open [6].

Although cartilage conduction has potential applications in the speech communication and

music industries, no common methods for calibrating cartilage conduction-based devices exist because the sound transmission routes are different from the known air and bone conduction routes (Figure 1b). The calibration methods for air and bone conduction devices have been standardized [7–9]. The sound pressure produced by air conduction devices (e.g. telephone, headphone, and hearing aid) can be simulated using an ear simulator [10] embedded in a head and torso simulator (HATS) [11], while the vibration produced by bone conduction hearing aids can be simulated using an artificial mastoid [12]. Previous studies have reported that the subjective loudness produced by a cartilage conduction transducer corresponded to the sound pressure level (SPL) measured in the canal (i.e. the air and cartilage-air pathways in Figure 1b), and that the hearing threshold using cartilage conduction was not affected by the bone pathway [5, 13]. The sound passing through the air and cartilage-air pathways is defined as the "cartilage conduction sound" (CCS)†. It is important to estimate the CCS accurately to calibrate the transducer.

Thus, as a first step in establishing a calibration method for cartilage conduction, we assessed simulation methods that would enable us to estimate the CCS produced by a transducer located at the entrance to the external auditory canal. First, we measured the SPL by inserting a probe microphone into the auditory canals of seven subjects, henceforth called participants ("measured CCS"). Second, we used two different procedures to obtain a "simulated CCS". The first procedure was the calibration method used for air conduction devices (Figure 2a). The cartilage conduction transducer was placed on the pinna simulator of the HATS, and the response from the ear simulator was recorded. The second procedure was a model simulation using representations of

the skull bone and aural cartilage in the canal (Figure 2b). We used polyurethane resin to form a pipe to model the cartilage component of the external auditory canal. We then attached this model cartilage to an existing skull bone model. The transducer excited the model cartilage, and we recorded the response using a probe microphone. Finally, we compared the measured and simulated CCS values and evaluated the reproducibility of the two procedures.

During the comparison between the measured and simulated CCS, the tip of the probe microphone was inserted to a point 15 mm from the entrance of the external auditory canal of both the participants and the cartilage simulator. To verify the spatial distribution of the SPL in the canal, the CCS was measured shifting the several measurement points in the external auditory canal. For the measured CCS of seven additional participants, the allowable maximum depth was 15 mm because of pain felt upon deeper insertion, and the measurement points were shifted by 5-mm intervals along the canal direction (Figure 3a). For the model cartilage and skull bone, we were able to insert the microphone all the way to the end of the canal (30 mm), in addition to the more shallow measurements (Figure 3b).

2. Material and Methods

2.1. General methods

The ring-shaped cartilage conduction transducer was used for the stimulation experiments as shown in Figure 1a [4]. The transducer was composed of a piezoelectric bimorph covered with an elastic material that broadened the spectral range of the vibration. The piezoelectric bimorph is a

piezoelectric actuator that responds to electric fields by extending one layer and contracting the other. An acrylic ring-shaped component was attached to the upper part of the transducer. The vibration measured by the artificial mastoid (Type 4152; Brüel & Kjaer, Naerum, Denmark) was strongest at frequencies above 1 kHz [4].

The input signals of the transducer were sine waves (duration: 1 s) ranging from 125 to 16 kHz in 1/12 octave steps (input voltage: 0.5, 1, and 2 V). The measured sounds were digitized for subsequent analysis at a sampling rate of 44.1 kHz and at 16-bit resolution (UA-101 analog-to-digital converter; Roland, Hamamatsu, Japan). We performed all measurements three times after removing and replacing the transducer to confirm repeatability, and averaged the measured SPL values. Measurements were conducted in an ordinary sound-insulated test room. All methods involving human subjects were approved by the ethics committee of Nara Medical University.

2.2. Acoustical measurements on participants

We conducted acoustical measurements in the external auditory canal of seven participants (25–36 years old) who had no disorders of the outer ears. The participants were fitted with a cartilage conduction transducer at the entrance to their left auditory canal. Sound in the external auditory canal was measured using a probe microphone (type 4182; Brüel & Kjaer, Naerum, Denmark), with a metallic probe tube (length: 100 mm, diameter: 1.24 mm) that allowed sound pressure to be measured in a confined, narrow space. A rubber tube was placed on the distal end of

the probe for safety. The tip of the rubber tube was inserted to a point 15 mm from the entrance of the external auditory canal. Neither the probe nor the rubber tube was in contact with the transducer.

To determine the SPL distribution in the canal, the insertion depth of the microphone was adjusted in a separate cohort of seven participants (24-40 years old) who had no disorders of the outer ears. The measurement positions were at the entrance of the canal (0 mm) and at points 5, 10, and 15 mm away from it toward to the eardrum (Figure 3a). Two additional positions were measured at the external end of the ring-shaped fitting part (−8 mm) and inside the pinna (−13 mm), the farthest point. Neither the probe nor the rubber tube was in contact with the transducer. The input voltage for the transducer was limited to only 2 V.

2.3. Simulation methods

The transducer was fitted on the left pinna simulator of the HATS (Type 4128; Brüel & Kjaer, Naerum, Denmark), which contained the ear simulator (Type 4159; Brüel & Kjaer, Naerum, Denmark) as shown in Figure 2a and c. The pinna simulator was embedded in a silicon rubber base (width × height × thickness: 50 × 60 × 10 mm) as a unit, and had the approximate shape and dimensions of a median adult human pinna [10].

As shown in Figure 2b and d, the transducer was fitted onto the model cartilage, which was made of soft polyurethane resin designed to simulate the elasticity of human skin (human skin gel, Exseal Corporation, Mino, Japan). The model aural cartilage was shaped as a pipe (external diameter: 15 mm, internal diameter: 10 mm, length: 15 mm) with an internal diameter and length

that corresponded to those of the cartilage component of the external auditory canal. The model cartilage was attached to a skull bone model (A20, 3B Scientific, Hamburg, Germany) of an adult human head with rubber cement (Blu·Tack, Bostik Australia Pty. Ltd., Thomastown, Australia). The gap between the model cartilage and skull bone model was filled with rubber cement to avoid sound leakage. Although the skull bone model did not have an eardrum, it did have a 15-mm-deep external auditory canal. After attaching the model cartilage, the total length of the external auditory canal was 30 mm. The probe microphone was used to obtain acoustical measurements in the canal by the same methods that were used for the participants (i.e. the distal end of the probe had a rubber tube, and the tip of the rubber tube was inserted to a point 15 mm from the open end of the model cartilage). Neither the probe nor the rubber tube was in direct contact with the transducer.

To determine the distribution in the canal, measurements were made at insertion depths into the model cartilage of 30 (eardrum), 15, 10, 5, 0 (entrance of canal), −5, and −10 mm (Figure 3b). Neither the probe nor the rubber tube was in contact with the transducer. The input voltage for the transducer was limited to 2 V.

3. Results

Figure 4 shows the simulated CCS measured in the HATS (Figure 4a) and in the model cartilage (Figure 4b), which also includes the average measured CCS of the seven participants (dashed lines). The standard deviations (SD) of the measured CCS were very similar at all input voltages (4.8, 4.7, and 4.9 dB for 0.5, 1.0, and 2.0 V, respectively). The measured CCS showed that

the canal resonance produced a maximum peak around 2.5 kHz, and a second broad peak was found around 700 Hz, which was associated with cartilage conduction [14]. The average differences among the input voltages were 6.4 dB (0.5 vs 1.0 V) and 5.7 dB (1.0 vs 2.0 V).

The simulated CCS in the HATS corresponded to the measured CCS at frequencies above 2 kHz, which was the threshold for effective sound production (Figure 4a). Conversely, the simulated CCS in the model cartilage corresponded to the measured CCS at frequencies below 1.5 kHz, and the differences between the measured and simulated CCS values were distributed throughout the high frequency range (Figure 4b). The averaged differences for input voltages of 0.5, 1.0, and 2.0 V in the HATS were 4.8, 4.9, and 5.0 dB, respectively. Those for input voltages of 0.5, 1.0, and 2.0 V in the cartilage simulator were 7.0, 7.3, and 7.4 dB, respectively.

The CCS values obtained from each participant’s ear and the model cartilage in each measurement position are compared in Figure 5a and b. Focusing on the frequency range below 1.5 kHz, the measured and simulated CCS behaviors were very similar. The differences in SPL with measurement position in the ear canal (solid lines) were unremarkable, whereas the SPL decreased rapidly at positions farther away from the ear canal (dashed lines). When the measurement position was located at the end of the canal (○ in Figure 5b), the SPL curve also agreed with the other curves at the measurement positions in the ear canal. For the measured CCS (Figure 5a), the peak around 700 Hz disappeared when the measurement position was 5 mm or more from the ring, while for the simulated CCS (Figure 5b), the 700 Hz peak was observed at all measurement positions.

4. Discussion

The aural cartilage was subject to stronger vibrations at frequencies below 1 kHz [6], resulting in greater sound generation by the cartilage-air pathway (Figure 1b). Thus, the measured CCS was increased around 700 Hz. Sounds below 500 Hz were lost because of the leakage from the open auditory canal. A previous study on cadavers found that stimulation of the mastoid with a transducer, after removing the outer ear and soft tissue from the cadaver head, reduces the SPL in the auditory canal by 10 dB for frequencies below 1.5 kHz [15]. Therefore, aural cartilage is thought to be a good sound generator in the low frequency range.

The two simulation procedures used in the present study had opposite spectral characteristics. The simulated CCS using the HATS agreed with the measured CCS in the high frequency range, and the simulated CCS using the model cartilage agreed with the measured CCS in the low frequency range. One advantage of the HATS simulator is that the pinna component can be made to resemble the complex shapes and smooth curves of the human outer ear [10]. When a human ear is excited by a nearby sound source, a secondary mode occurs around 5 kHz due to the resonance of the concha, and higher order modes also occur due to the smaller groove gaps formed by the scaphoid fossa, tragus, and crus helicis [16, 17]. To observe the resonance effect of the pinna, we also measured the sound around the pinna simulator of the HATS. The SPL recorded around the pinna simulator (−13 mm from the entrance of canal) installed in the HATS wearing the transducer (as in Figure 2a) is shown in Figure 6. The measurement procedure was the same as that used for the human participants. The dashed line in Figure 6 indicates the measured CCS obtained from the

participants in the same position, which was extrapolated from Figure 5a. The simulated and measured CCS values have the same peaks at 2.5 kHz (resonance of the canal) and 5 kHz (resonance of the concha). Thus, it is possible that some of the resonances and the sound propagation associated with the pinna shape produced a simulated CCS that was closer to the measured CCS for high frequency sounds.

One advantage of the model cartilage is that aural cartilage is a large component of the external auditory canal. The vibration measured in the model cartilage had the spectral vibration characteristics of real aural cartilage in that the vibration decayed quickly for frequencies above 1 kHz (Appendix) [5]. While the large and heavy pinna simulator does not vibrate like real cartilage, the model cartilage can mimic the low-pass spectral shape of real cartilage vibrations, and thus generate sounds in the low frequency range similar to human aural cartilage.

When taken together, these results suggest that the canal cartilage and pinna cartilage contribute to the generation of the cartilage conduction sound in the low and high frequency ranges, respectively. Although the two simulation procedures are insufficient for representing the measured CCS individually, their spectral sensitivities are complementary. The simulated CCS produced in the HATS above 1587 Hz and the simulated CCS produced in the model cartilage below 1587 Hz are shown in Figure 7. The resulting combined simulated CCS chart agrees with the measured CCS. The average differences between the measured and simulated CCS levels were 3.6, 3.7 and 3.5 dB at input voltages of 0.5, 1.0, and 2.0, respectively. These differences are within the SD of the measured CCS among the participants. Thus, a combination of the two procedures is an appropriate

calibration method for the cartilage conduction transducer. To calibrate the CCS, we propose designing a new pinna simulator that can also simulate the vibration at the aural cartilage of the pinna and the external auditory canal when mounted to the HATS system. When the current pinna simulator is replaced with the new one, the HATS system will be able to estimate the CCS in the canal with high accuracy. The new pinna simulator will also simulate the sound in the external auditory canal when using an insertion-type earphone, which is likely to excite the aural cartilage.

Based on the differences in CCS at different microphone positions, the CCS is largely unchanged along the ear canal. In other words, the CCS is uniformly distributed in the canal. The first peak of the CCS is around 700 Hz, and this wavelength (0.49 m) is much longer than any dimension of the ear canal. Therefore, the sound generated by the aural cartilage vibration does not contribute to it. The peak at 700 Hz disappears when the measurement position is outside of the canal and farther than 13 mm from the canal entrance, while the simulated CCS in the model cartilage has a residual peak at 700 Hz in the same position. The model cartilage and the ring opening of the transducer lie along a straight line, as shown in Figure 3b, so that the CCS in the canal may be effectively transmitted, even to the outside of the canal. In contrast, the opening of the ring worn by the participants did not lie along a similar straight line (Figure 3a), and the ear tragus reflects the CCS coming from the canal. Therefore, the outer sound field of the canal may be also simulated more accurately using the pinna simulator (Figure 6).

Although the resonance frequencies that arise in the external auditory canal do not change, the dip frequencies associated with them have been shown to move depending on the position of the

microphone in the canal [16]. This previous study showed that the insertion depths of 20 and 15 mm from an eardrum are associated with dip frequencies of 5 and 7 kHz, respectively. For the model cartilage, the corresponding dips were observed for the measurements at 10 mm (i.e. 20 mm from the eardrum) and 15 mm (i.e. 15 mm from the eardrum) as shown in Figure 5b. Conversely, the measured CCS in the participants did not display such peaks and dips caused by resonance (Figure 5a). Although the resonances in the higher frequencies (around 9 and 12 kHz) could be observed in each individual participant, they were obscured in the average values because the resonance frequencies shifted slightly among the individuals. Therefore, it is difficult to simulate the higher frequency resonances using this unique representation system.

When the concept of cartilage conduction is applied to mobile phones, a robust signal above the environmental noise can be realized [18]. Cartilage conduction-based transducers are suitable for listening to music in noisy streets and train cars. In the case of cartilage conduction-based hearing aids, people with hearing loss, who have to use a hearing aid all day, would be much less stressed without the feeling of ear fullness [19-21]. In addition, unlike bone conduction hearing aids, cartilage conduction hearing aids do not result in pain and erosion at the contact face of the transducer. At present, we are developing a new cartilage conduction transducer driven by electromagnetic power [22]. When the proposed simulation method is available for different kinds of transducers, we plan to integrate our findings to produce a standardized method of calibration. For example, the output sound from the cartilage conduction smartphone can be defined using the calibration method. For the music player device, the cartilage conduction transducer must

limit the output SPL to the low risk range to prevent noise-induced hearing loss. This calibration method is also essential for consolidating all the information available on cartilage conduction hearing aids, whose performance should be adjusted according to the progression of hearing loss.

5. Conclusion

We used a head and torso simulator (HATS) and a model of skull bone and aural cartilage to simulate sounds generated in the external auditory canal by cartilage conduction. The HATS was able to reproduce the cartilage conduction sounds above 2 kHz, while the model cartilage was able to reproduce the cartilage conduction sounds below 1.5 kHz, including the spatial distribution of SPL in the ear canal. Thus, a combination of the two methods can be effectively used to estimate the measured cartilage conduction sound within the standard deviation observed among individuals. These results suggest that the pinna and external auditory canal, which both generate cartilage conduction sounds, may do so at different frequency ranges.

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Appendix: Vibration measurement for the model cartilage

To determine the reproducibility of the vibration of the model aural cartilage, we used the same measurement conditions to measure vibrations as for measuring CCS. The probe microphone was replaced with a subminiature charge accelerometer (type 4374; Brüel & Kjaer, Naerum, Denmark). The small accelerometer (diameter: 5 mm, height: 6.7 mm, weight: 0.65 g) was attached to the model cartilage with double-sided adhesive tape, and the vibration acceleration level (VAL) was calculated. The simulated VAL (Input voltage 2 V) averaged across three measurements is shown in Figure A1. The dotted lines indicate the measured VAL of real aural cartilage [6]. The spectral characteristics of the simulated VAL were similar to those of the measured VAL, and both VALs became higher at frequencies below 1 kHz. The average difference between the measured and simulated VALs was 8.2 dB. Because the model cartilage does not have a pinna or surrounding soft tissue components, it is likely that it has a larger vibration depth than a real ear.

Footnote

†) A previous study defined the cartilage conduction sound in a more limited sense, as the sound passing through only the cartilage-air pathway [5].

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FIGURE CAPTIONS

Figure 1. (a) Cartilage conduction transducer, and (b) the air, cartilage-air, and bone pathways during use.

Figure 2. Simulation with the (a) HATS and (b) model skull bone and aural cartilage, with cross-sections of the (c) HATS and (b) skull bone model. A probe microphone was inserted in the model cartilage.

Figure 3. Microphone positions in the (a) human participant ears and in the (b) model skull bone and cartilage.

Figure 4. Simulated CCS (solid line) obtained in (a) HATS and (b) the model aural cartilage. The dashed lines indicate the measured CCS obtained from the participants.

Figure 5 (a) Measured CCS obtained from the participants, and (b) the simulated CCS obtained in the model aural cartilage at different microphone positions.

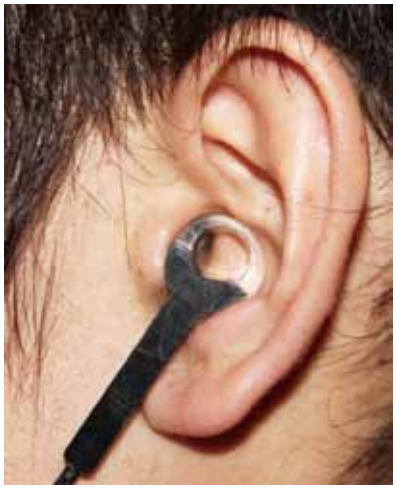
Figure 6. Comparison between the measured CCS in the participants and the simulated CCS in the pinna simulator at the measurement position located 13 mm laterally from the canal entrance.

Figure 7. Comparison between the measured CCS and the simulated CCS obtained by combining the outputs from the HATS and the model aural cartilage.

Figure A1. Measured VAL and simulated VAL obtained from the model aural cartilage. The measured VAL was averaged among the outputs at three points (the ear tragus, the scaphoid fossa, and behind the concha) in a real human outer ear [5].

Figure1

(a)



(b)

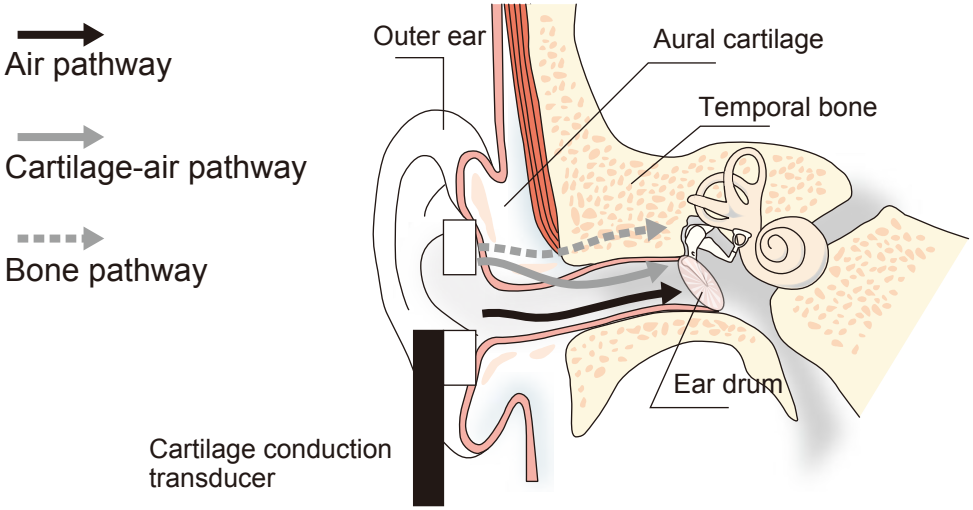


Figure 1, Shimokura et al.

Figure2

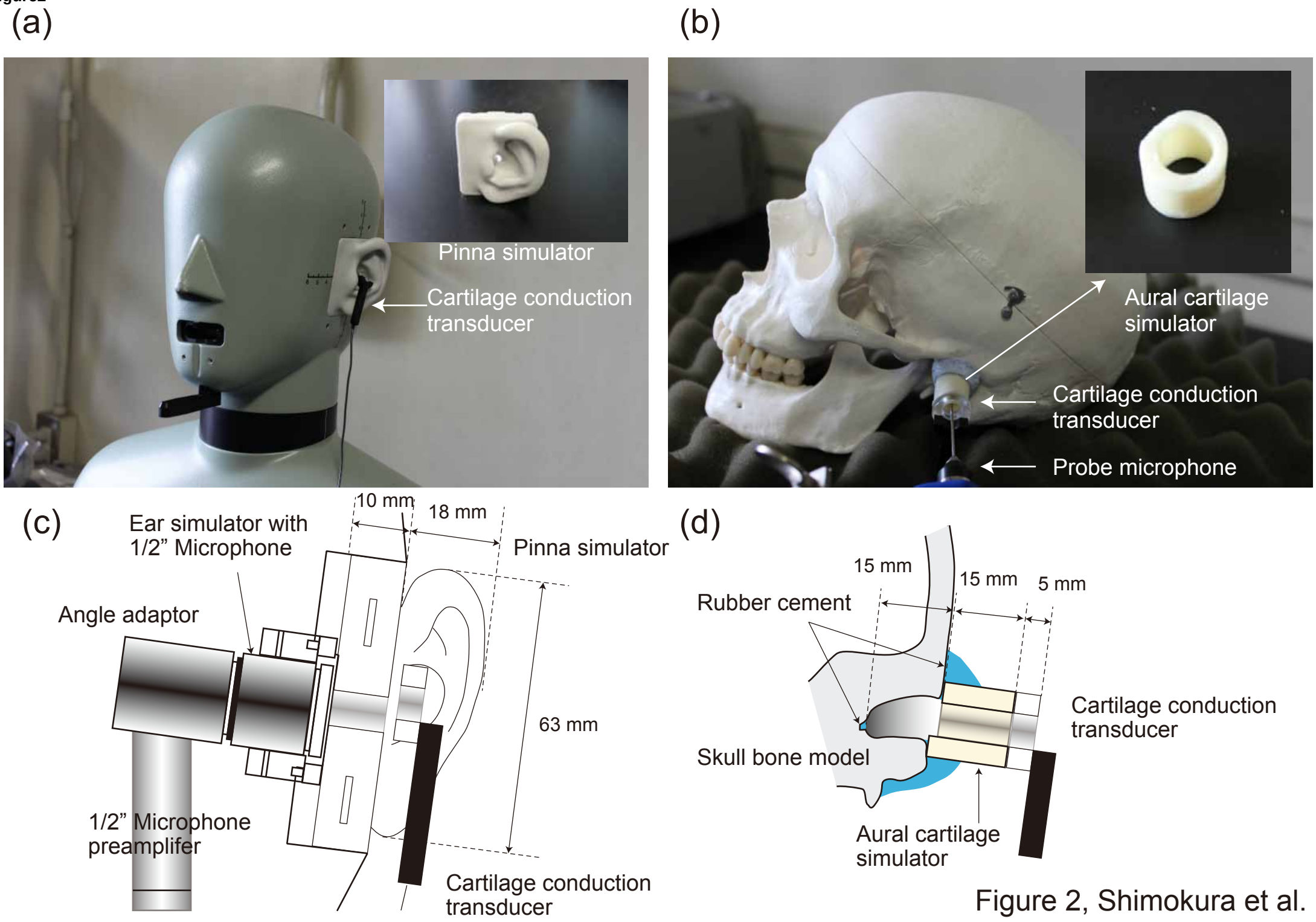


Figure 2, Shimokura et al.

Figure3

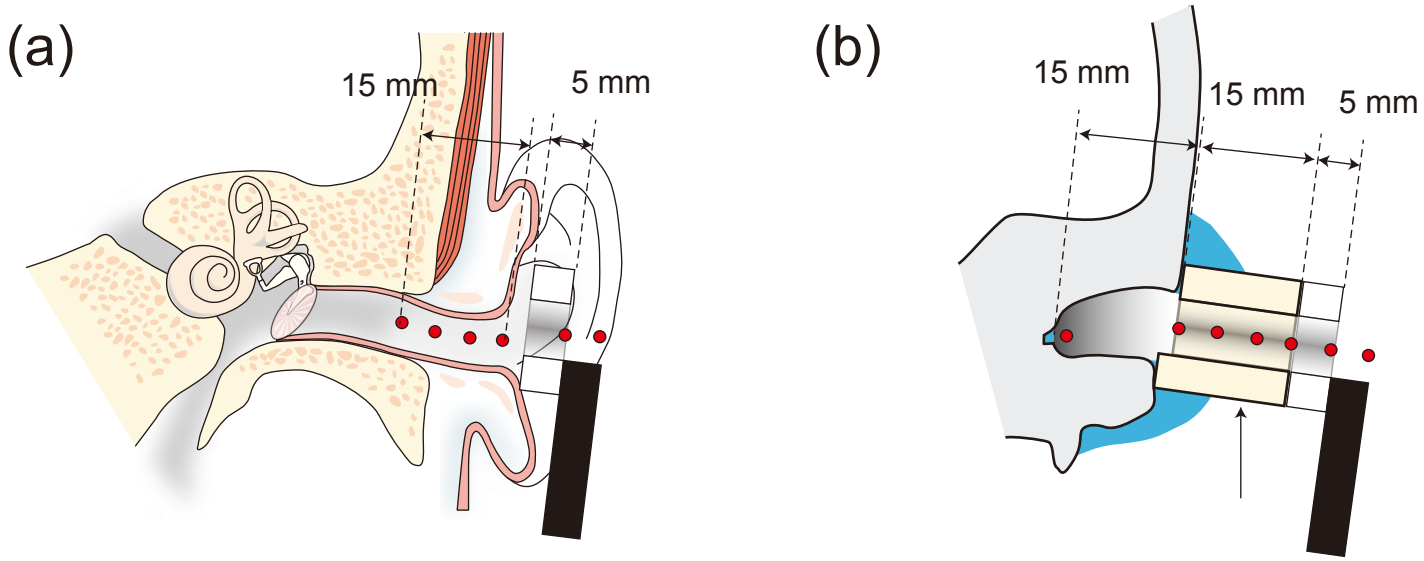


Figure 3, Shimokura et al.

Figure4

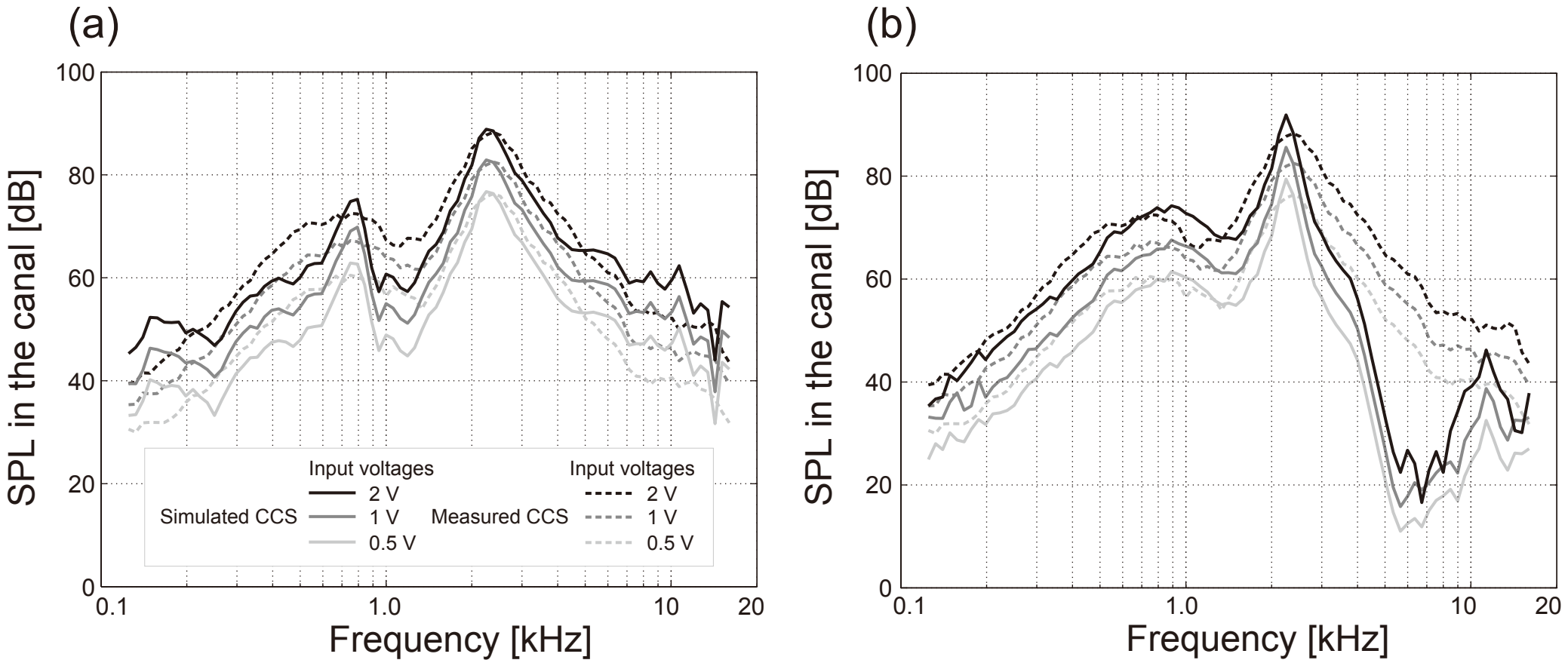


Figure 4, Shimokura et al.

Figure5

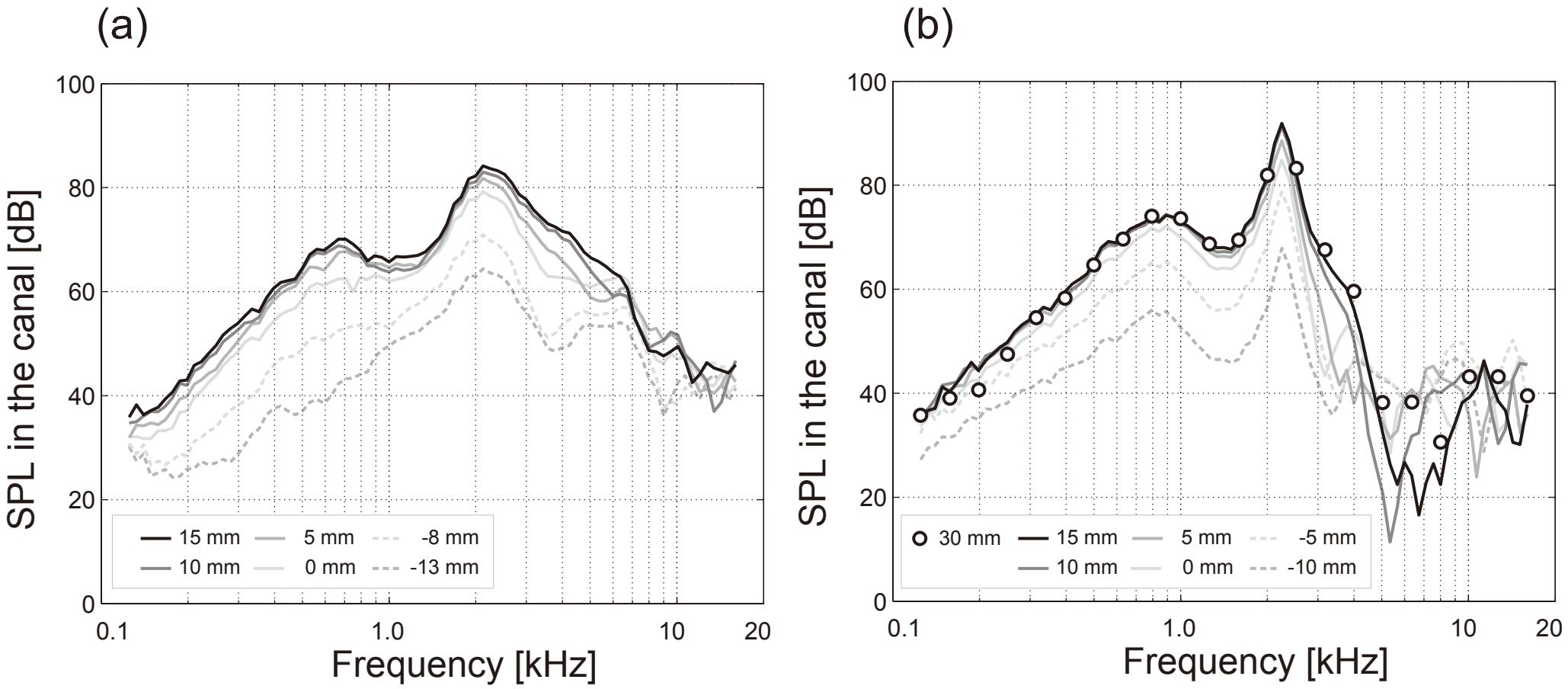


Figure 5, Shimokura et al.

Figure6

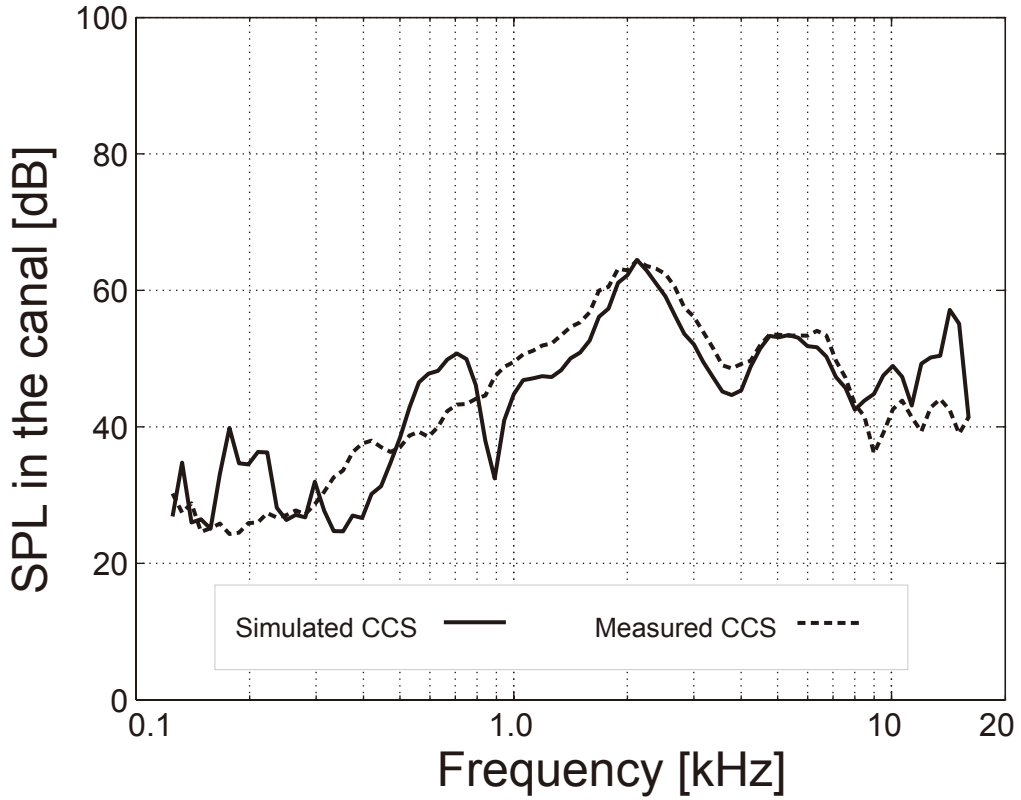


Figure 6, Shimokura et al.

Figure7

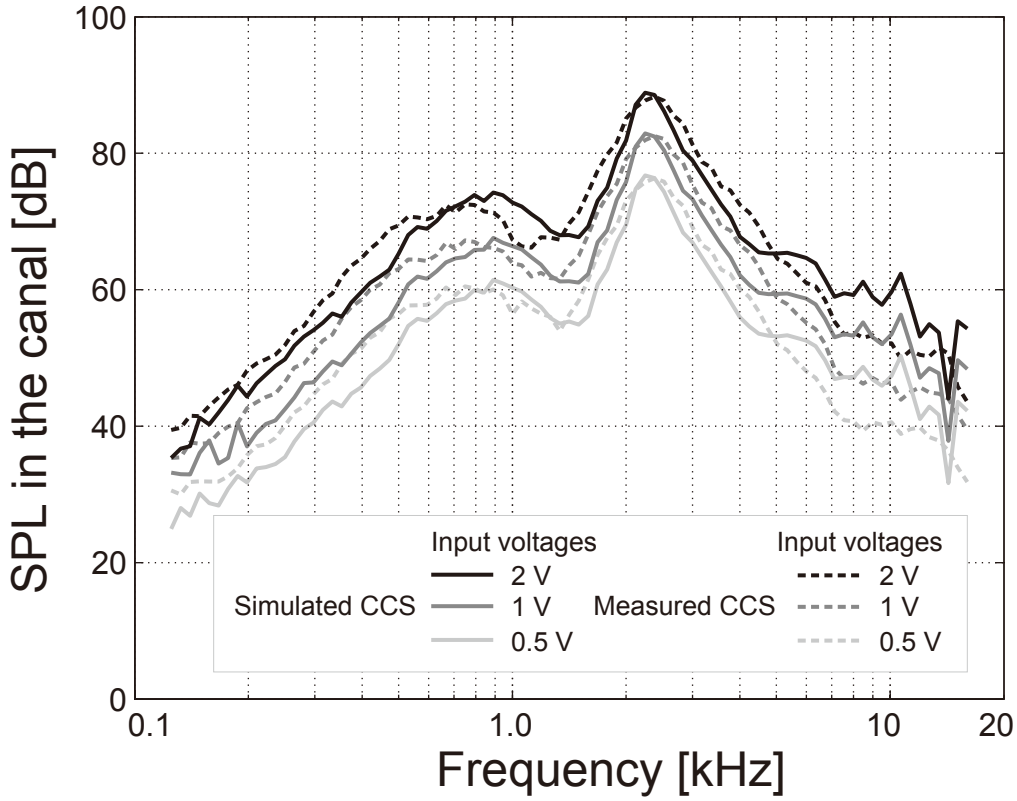


Figure 7, Shimokura et al.

FigureA1

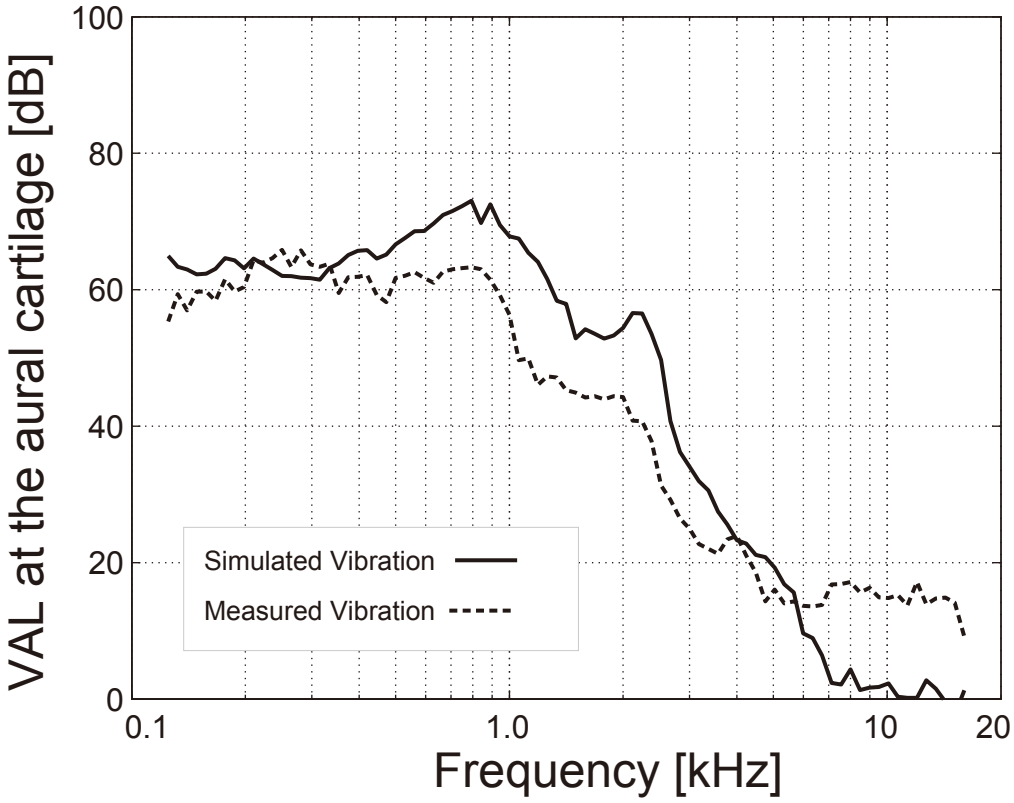


Figure A1, Shimokura et al.