Research Article

Intracochlear Sound Pressure Measurements in Normal Human Temporal Bones During Bone Conduction Stimulation

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ABSTRACT

Bone conduction (BC) is heavily relied upon in the diagnosis and treatment of hearing loss, but is poorly understood. For example, the relative importance and frequency dependence of various identified BC sound transmission mechanisms that contribute to activate the cochlear partition remain unknown. Recently, we have developed techniques in fresh human cadaveric specimens to directly measure scalae pressures with micro-fiberoptic sensors, enabling us to monitor the input pressure drive across the cochlear partition that triggers the cochlear traveling wave during air conduction (AC) and round-window stimulation. However, BC stimulation poses challenges that can result in inaccurate intracochlear pressure measurements. Therefore, we have developed a new technique described here that allows for precise measurements during BC. Using this new technique, we found that BC stimulation resulted in pressure in scala vestibuli that was significantly higher in magnitude than in scala tympani for most frequencies, such that the differential pressure across the partition—the input pressure drive—was similar to scala vestibuli pressure. BC (stimulated by a Bone Anchored Hearing Aid [Baha]) showed that the mechanisms of

sound transmission in BC differ from AC, and also showed the limitations of the Baha bandwidth. Certain kinematic measurements were generally proportional to the cochlear pressure input drive: for AC, velocity of the stapes, and for BC, low-frequency acceleration and high-frequency velocity of the cochlear promontory. Therefore, our data show that to estimate cochlear input drive in normal ears during AC, stapes velocity is a good measure. During BC, cochlear input drive can be estimated for low frequencies by promontory acceleration (though variable across ears), and for high frequencies by promontory velocity.

Keywords: bone conduction, normal hearing, human temporal bone, intracochlear pressure, fiberoptic pressure sensors, laser Doppler velocity measurements

Abbreviations: AC – Air conduction; Baha – Bone anchored hearing aid; BC – Bone conduction; CR – Correction ratio; LED – Light-emitting diode; P – Pressure; $P_{EC,ST,SV}$ – Pressure in the ear canal, scala tympani or scala vestibuli; Ref_{ST} , Ref_{SV} – Reference measurements for scala tympani and scala vestibuli; RW – Round window; stap – Stapes; ST – Scala tympani; SV – Scala vestibule; V – Velocity; $V_{stap,prom,RW}$ – Velocity of stapes, promontory or round window; V_{RMS} – Voltage (root mean square); P_{Diff} – Cochlear input pressure drive (differential pressure) Christof Stieger and Xiyang Guan contributed equally as first authors.

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INTRODUCTION

Bone conduction (BC) is utilized in clinical settings for both diagnosis and treatment. The standard audiogram relies on the BC response to quantify sensorineural hearing loss. However, sound transmission during BC can be affected by macro-mechanical disturbances of the middle and inner ear despite a normal sensorineural system, resulting in diagnostic uncertainties. For example, immobilization of the stapes by otosclerosis can result in a dip in the audiometric BC threshold curve known as Carharts notch (Carhart 1950). Another example is the increase in BC response at low frequencies due to a fistula in the otic capsule, superior canal dehiscence (Cox et al. 2003; Mikulec et al. 2004). BC is also used in hearing aids to treat conductive and mixed hearing loss (Snik et al. 2005), and sometimes for single-sided deafness (Hol et al. 2010; Kompis et al. 2011; Pfiffner et al. 2009; Wazen et al. 2008). BC is also associated with acoustic trauma due to high vibrations such as in fighter jets (McKinley 2009; Berger et al. 2003). However, the extent and frequency dependence of various mechanisms underlying BC are not well understood. A limitation has been the difficulty in studying the mechanisms of how BC sound is transmitted and produces a measurable response in the human cochlea.

Experimental investigations of BC in human cadaveric specimens have been limited to certain types of mechanical measurements (e.g., velocity and acceleration) of structures surrounding the cochlea, such as the bony otic capsule, ossicles and round window (RW) membrane (Homma et al. 2009, 2010; Stenfelt and Goode 2005b; Stenfelt et al. 2000; Stenfelt et al. 2004b). We overcome the limitations of these measurements by direct measurement of the intracochlear pressures in scala vestibuli (SV) and scala tympani (ST). These pressures determine the input pressure drive that starts the cochlear traveling wave (Nakajima et al. 2009). The *cochlear input pressure drive* (complex difference in pressure between SV and ST at the base) has been shown to be nearly identical in magnitude and phase frequency response to neurophysiological measurements (cochlear microphonic) recorded near the same cochlear location during air-conducted (AC) stimulation in animals (Dancer and Franke 1980; Lynch et al. 1982). AC-evoked cochlear input pressure drive in fresh cadaveric specimen is also affected similarly to clinical AC audiograms by structural perturbations such as ossicular discontinuity and superior canal dehiscence (Nakajima et al. 2009; Niesten et al. 2015; Pisano et al. 2012).

Intracochlear pressure measurements combined with velocity measurements allows us to quantify the various impedances and volume velocities of sound transmission. This provides increased understanding of the fundamental mechanisms of sound transmission to each scalae for different types of stimulation (Nakajima et al. 2009; Stieger et al. 2013). Intracochlear pressure measurements in fresh human cadaveric preparations can thus provide important information that is not even obtainable from neurophysiological measurements in live preparations. For example, sound pressure plus velocity measurements can determine not only the input pressure drive across the partition that starts the traveling wave (similar to information gained by neurophysiological measurements in live animals), but can also determine the mechanisms and the paths of volume velocity (paths of sound transmission). However, the vigorous shaking produced by BC stimulation poses challenging conditions, complicating the ability to make accurate intracochlear pressure measurements.

To address the challenging conditions imposed by BC stimulation, we have developed a new method to measure intracochlear pressures during BC stimulation. We detail (1) how we mitigate measurement artifacts that can be induced by the motion of the entire bone, enabling the desired pressure measurements to be made; (2) how one can check the calibrations of the pressure sensors during our new recording method to ensure accurate pressure measurements; (3) our analyses of magnitude and phase of scalae pressures and extracochlear motions during BC; and (4) how BC compares to AC.

METHODS

General methodologies used for pressure and velocity measurements are in common with our previous published descriptions (Nakajima et al. 2009; Stieger et al. 2013). In this manuscript, we detail new experimental methods to enable intracochlear pressure recordings to be made without artifacts arising from the significant motion of the specimen during BC. We further introduce new calibration techniques suitable for these methods.

Air and Bone Conduction (AC & BC) Stimulation

A coupler for the speaker and probe-tube microphone was sealed to the bony external auditory canal. AC stimulation was presented as in past studies: a speaker (40-1377, Radio Shack) applied 74 logarithmically distributed 10 ms tones between 0.1 and

FIG. 1. Illustration demonstrating the different methods for securing the specimen. **a** The temporal bone was partially embedded in PlayDoh Putty, or **b** in a temporal bone holder, or **c** cemented to a brass bar. Movements of promontory (Prom), stapes, and round window (RW) were measured using laser Doppler vibrometry.

Experimental setup with fiberoptic microsensors for pressure measurements in scala vestibuli (P_{SV}) and scala tympani (P_{ST}) is shown. BC stimulation was performed with a bone anchored hearing aid system (Baha BP100) attached to a bone screw. AC stimulation sound was presented at the external auditory canal (EC)

20 kHz, with levels between 60 and 110 dB SPL with linearity established for all frequencies.

For BC, a Baha (Baha BP100, Cochlear, Australia) was anchored to a region near the ear canal having thick bone (Fig. 1). The exact location and relative direction of the anchoring screw varied slightly across experiments because the location of thick bone varied across ears. The BP100 processor was programmed with the conventional Baha fitting software in a linear mode. The Baha was driven with 40 mV_{RMS} voltage, confirmed to operate in the linear region (considerably lower than the 69 mV_{RMS} where distortions occurred).

Stimulus frequencies were identical to that used for AC stimulation. All response measurements to AC and BC stimulations were averaged 25 and 50 times, respectively.

Intracochlear Pressure and Vibration Measurements

Custom-made fiberoptic pressure sensors with a diameter of approximately 200 μm were used for intracochlear pressure measurements (Nakajima et al. 2009; Olson and Nakajima 2015; Stieger et al. 2013). Sound pressure results in the vibration of a gold-covered membrane on a capillary tip, modulating a

static reflected LED light. This modulation is detected with a photodiode (Chhan et al. 2016; Olson 1998; Olson and Nakajima 2015).

The sound pressure level at the ear canal (P_{EC}) was recorded with a probe-tube microphone (ER 7C, Etymotic) with the tip located 1–2 mm from the tympanic membrane near the umbo to monitor P_{EC} stability throughout the experiment. The velocities of the stapes, RW, cochlear promontory and sensors were measured with a laser vibrometer (Polytec CLV 1000). For the stapes measurements, the laser was aligned along the plane defined by the posterior and anterior crus, to measure mostly the piston-type motion and minimize measuring the pitch motion (anterior-posterior rocking along the long axis of the footplate). For promontory measurements and RW measurements, the laser was approximately perpendicular to the corresponding structure. We did not apply a cosine correction factor. Details of these methods have been described in earlier publications (e.g. Nakajima et al. 2009; Stieger et al. 2013).

Preparation of the Temporal Bone Specimen

Thirty-seven human temporal bones were harvested within 24 h post-mortem and either stored in saline

for experimentation or frozen for later use. The new BC preparation method was developed and refined in the first 14 specimens. Thereafter, reliable measurements for this study were made in 8 out of 23 specimens. Unreliable measurements or experiments were abandoned for various reasons. First, if unusual mechanics, e.g., unusually stiff or loose ossicles were observed during stapes velocity measurements. Second, if a specimen had evidence of intracochlear air or leak which was seen when the phase between stapes and round window movement was not a half cycle. Third, if any change occurred to the specimen during the experiment, determined by comparing repeated velocity measurements before and after pressure sensor insertions, and repeated pressure and velocity measurements throughout the experiment (detailed below). We also abandoned experiments with unstable pressure sensors.

The middle and inner ear were accessed through a facial recess opening. The cochlear promontory was thinned close to the oval and round windows to facilitate the drilling of the cochleostomies for the insertions of the fiberoptic pressure sensors in SV and ST.

Securing Methods for the Temporal Bone

In contrast to AC or RW stimulation in our previous studies, BC excitation with a Baha results in frequency-dependent complex 3-D motion of the entire temporal bone specimen (unlike the uniform motion of the entire specimen obtained when mounted on a shaker). Thus an adequate method for holding the specimen with Baha stimulation was investigated by comparing three holding methods: (1) the temporal bone was held in a temporal-bone holder (metal bowl with screws to hold the specimen) placed on top of a rubber ring stand; (2) a metal rod was firmly glued with dental cement to the posterior aspect of the temporal bone, and this rod was firmly held in place by metal hardware; and (3) the temporal bone was placed in malleable putty (Play-Doh®, Cincinnati, Ohio, USA) that cushioned the specimen (Fig. 1). (The vibration comparisons of these holding methods are described in the Results section below.) The bone, the BC stimulator, and the intracochlear pressure sensors were all mounted to an optical board placed on an air table. Figure 1 also illustrates the Baha BC stimulator attached near the bony ear canal and the fiberoptic sensors inserted into SV and ST.

Preventing Potential Artifacts in Pressure Measurements Evoked by BC

As in previous studies (Nakajima et al. 2009), 210–220 μm diameter cochleostomies in SV and ST at the base of the cochlea were created by hand with a sharp

pick with the cochlea submerged under normal saline (with fluid in the middle-ear cavity) to prevent air from entering the cochlea. While the region of the cochleostomy continued to be under fluid, a micro-manipulator (X330, Narashige, Japan) was used to insert a sensor through the cochleostomy until the tip of the sensor was in perilymph, 100–150 μm below the inner surface of the bony capsule. With the sensors in place, the cochleostomies were hermetically sealed to the outer bony surface with dental impression material (Jeltrate®, Milford, Delaware, USA), which cures in wet environments to a rubbery consistency. However, because Jeltrate is compliant, the otic capsule can vibrate differentially from the sensor during BC stimulation, resulting in measurement artifacts, where relative motion between the vibrating bone (including the intracochlear fluid) and the sensor may artificially *increase* the measured pressure. Another problem is that BC vibration can disrupt the soft Jeltrate seal of the sensor at the cochleostomy, resulting in a pressure release and a *decrease* in pressure measurement. We did not encounter such problems in our previous studies with AC and RW stimulation, but the vigorous motion during BC stimulation can result in significant inaccuracies in the pressure measurements (detailed in the Results section).

In order to reduce relative motion between sensor and bone, we used a new method to secure the sensor. We initially applied a minimal amount of Jeltrate to hermetically seal the cochleostomies around the sensor. Once the Jeltrate was set, its outer surface and the surrounding area were completely dried, after which dental cement (Durelon, 3 M Corp) was applied over the Jeltrate to tightly fix the sensor to the bony surface. (Dental cement cures to a much firmer consistency than Jeltrate, helping to mitigate differential movement between the sensor and the bone.) The sensor was then released from the micromanipulator, enabling the sensor to vibrate together with the otic capsule.

Accurately Determining the Sensitivity of the Pressure Sensor

Determining the absolute intracochlear pressures with accuracy is important in calculating the input pressure drive (complex difference between the scalae pressures), but the sensitivity (i.e., gain magnitude) of a pressure sensor can change during the experiment. In previous studies (Nakajima et al. 2009; Niesten et al. 2015; Pisano et al. 2012), we were able to confirm stability of the sensitivity by comparing the calibrations between the initial calibration (before sensor insertion) and final calibration (after sensor removal at the end of the experiment). In this study, final post-experiment calibration was not possible

because the firmly-glued sensor tips were often traumatized during the removal from the otic capsule. To enable accurate pressure measurements, we have developed an in situ method of determining the sensitivity of the sensors while the sensors are in the cochlear fluid. This multi-step method is described below.

A flow chart of our calibration method is shown in Fig. 2. The fiberoptic pressure sensors were initially calibrated using a shaker (4290, Bruel & Kjaer,

Fig. 2. Flow chart demonstrating our experimental procedure and intracochlear calibration of pressure sensors. AC: air conduction; BC: bone conduction; P_{SV} : pressure in scala vestibuli; P_{ST} : pressure in scala tympani; V_{stap} , V_{RW} , V_{prom} : velocities of the stapes, round window, promontory, respectively. After passing the dashed loop because AC response is stable before and after BC measurements, we can be confident that our absolute values for pressure measurements during BC are accurate. If AC pressure magnitude responses before and after BC measurements have a frequency-independent shift, sufficient information is available for calibration correction and measurements are repeated (loop of dashed line) to obtain accurate BC responses

Denmark) with attached water vial (C1 in Fig. 2) ((Nakajima et al. 2009; Nedzelnitsky 1980; Schloss and Strasberg 1962). If sequential calibrations were stable (within 0.5 dB in magnitude and no change in phase) across all frequencies, sensors were inserted into the cochlea. The phase of the sensor sensitivities was stable throughout experiments. After sealing the sensors with only Jeltrate (a soft dental impression material), the AC-evoked stapes and RW velocities were compared to initial velocity measurements made prior to making cochleostomies and sensor insertions. We checked the stability of the magnitude response as well as the half-cycle difference in phase between RW and stapes motion. This checked that: the mechanics of the inner ear and its mobile surroundings did not change owing to sensor insertion, the sensors were hermetically sealed at the cochleostomies, and air likely did not leak into the cochlea. We found that imperfect seals or introduction of air can easily occur, altering velocities of the stapes and RW and their phase relationship (Merchant et al. 2016; Frear et al. 2018). The intracochlear pressure responses to AC sound in both scalae were measured simultaneously with velocity measurements (AC1 in Fig. 2). Under fluid, the sensors were then removed from the Jeltrate-sealed cochleostomies and recalibrated in the shaker-mounted vial (C2) at least twice to confirm stable (<0.5 dB) sequential recordings. If this calibration (C2) and the initial calibration (C1) were stable (<2 dB difference across all frequencies), we used the AC intracochlear pressure responses (AC1) as a reference for future AC pressure responses ($Ref_{SV} = P_{SV_AC1}/V_{stap_AC1}$; $Ref_{ST} = P_{ST_AC1}/V_{stap_AC1}$) under normal middle- and inner-ear condition. The sensors were again inserted into the cochlea, sealed with minimum amounts of Jeltrate, and then firmly cemented in place as described above.

The middle ear cavity was then soaked with saline to prevent changes due to drying. After the saline was removed, V_{stap} and intracochlear pressures were measured in response to AC stimulation (AC2). To determine if air entered the cochlea during reinsertion of the sensors, both V_{stap} and V_{RW} were again measured to examine stability and half-cycle phase relationship. Usually, AC pressure responses measured with rigidly-cemented sensors were the same as those recorded previously with only the Jeltrate seal. If a constant pressure magnitude shift across all frequencies occurred, then correction for the pressure sensor sensitivity change could be performed.

After AC2 measurements, the intracochlear pressures and the velocities of the stapes, RW, and promontory produced in response to BC stimulation were recorded. The vibration during BC may change pressure sensor sensitivity and the integrity of the seals, resulting in inaccurate BC response measure-

ments. To ensure that the sensors and seals were stable during BC stimulation, AC responses were repeated to make certain that AC-evoked velocities and pressures before and after BC recordings were stable (within <0.5 dB for all frequencies). A shift in AC pressure magnitude response that was independent of frequency was deemed likely due to a shift in sensor sensitivity, while a change at only low frequency could be due to a leak between sensor and bone at the cochleostomy. More cement at the cochleostomy site determined if this low-frequency shift was resolved. AC, BC and then AC measurements were repeated if necessary until the AC responses were the same before and after BC measurements—in which case we knew the sensitivities of the pressure sensors, enabling accurate pressure measurements during BC.

A frequency-independent shift in pressure magnitude with stable phase indicated a sensor sensitivity shift that we quantified by observing the shift between the initial AC1 measurements and subsequent AC measurements. Corrections for sensitivity shifts were performed as follows: Based on the initial intracochlear AC1 measurement and the stable AC measurement (ACi), we defined a general correction factor for each sensor (CR_{SV} for SV sensor and CR_{ST} for ST sensor) to compensate for shifts of the sensor's sensitivity.

$$\begin{aligned} CR_{SV} &= Ref_{SV} / (P_{SV_ACi} / V_{stap_ACi}) \\ &= (P_{SV_AC1} / P_{SV_ACi}) * (V_{stap_ACi} / V_{stap_AC1}) \\ CR_{ST} &= Ref_{ST} / (P_{ST_ACi} / V_{stap_ACi}) \\ &= (P_{ST_AC1} / P_{ST_ACi}) * (V_{stap_ACi} / V_{stap_AC1}) \end{aligned}$$

To accurately compensate for the effect of sensor sensitivity shift, the intracochlear pressures recorded during BC were multiplied by the correction factor (which ranged from 0 to 4.3 dB in the eight specimens reported).

If an observed sensitivity change during AC is independent of frequency (unlike a leak which results in low-frequency decrease), we use a correction factor that accounts for the sensitivity change. Additionally, all of our BC measurements that are considered accurate require AC responses to be identical before and after the BC recording. If the AC responses (pre and post BC-stimulation run) are not identical, it can be due to instability at the sensor-cochleostomy site or change in sensor sensitivity that occurred during BC stimulation, resulting in unreliable and inaccurate BC response measurements.

RESULTS

Status of the Middle and Inner Ear Assessed with AC Response

Each specimen's middle ear motion was compared to established norms. Stapes velocity (V_{stap}) magnitude from this present study of eight ears (colored lines in Fig. 3a) were within the 95 % confidence interval about the mean of the of the ASTM standard (Rosowski et al. 2007 plotted with gray shading in Fig. 3) for most frequencies. One specimen #162 was slightly low for a wide frequency range (designated with orange dashed lines).

All specimens were checked for possible air in the inner ear or unusual leak of the inner-ear fluid by determining the phase relationship between the stapes and RW velocities in response to AC. If the low-frequency (<500 Hz) phase relationship deviated from $\frac{1}{2}$ cycle, either air was introduced into the inner ear or a significantly abnormal volume velocity leak existed (due to pathology or inadvertent damage during temporal-bone extraction, storage or preparation: [Merchant et al. 2016; Nakajima et al. 2009;

FIG. 3. Status of temporal bone specimens. **a** Magnitudes of stapes velocity referenced to ear-canal pressure are plotted with colored lines. The gray area designates the range of ASTM standard for normal ears ($\pm$ 95 % CI) of stapes velocity. One measurement, #162 displays slightly

lower velocities than the standard, designated with a dashed line. **b** Phase difference between stapes and round-window velocities are 0.5 cycle at low frequencies

Stenfelt et al. 2004a].) Figure 3b shows that the inner ears used in this study had $\frac{1}{2}$ cycle differences in phase at low frequencies. Experiment #162 had slight deviation from exactly $\frac{1}{2}$ cycle below 200 Hz, possibly because the signal-to-noise was low; however, if air was introduced or if there was a third window, the deviation from $\frac{1}{2}$ cycle would have been significantly larger and would have extended towards higher frequencies than that exhibited by #162 (thus we included #162 in our analyses). In the case of a significant deviation from $\frac{1}{2}$ cycle in phase, the experiment was halted (imperfect $\frac{1}{2}$ cycle occurred in 8 out of 23 specimens, as can happen with inner-ear air or fistula (Chien et al. 2007). After the sensors were cemented to the otic capsule at the cochleostomies, the stapes and round-window velocity magnitudes and phase relationships were confirmed to be stable compared to earlier measurements recorded before sensors were introduced.

Relative Movement Between Otic Capsule and Pressure Sensor During BC

Vigorous shaking from BC stimulation resulted in a challenging environment for intracochlear pressure measurements because the pressure sensor can move differently from the otic capsule, producing artifacts. To determine relative movement between sensor and otic capsule, the laser vibrometer was focused on reflectors on the shaft near the tip of the pressure sensor as well as on the cochlear promontory bone near the sensor. Figure 4 shows the relative motion between sensor and promontory bone during BC stimulation in a representative example. In this example, the sensor vibrated less than the bone by 10–25 dB when using Jeltrate alone to seal the sensor at the cochleostomy (Fig. 4, dotted line). Across experiments and frequencies using Jeltrate alone, we saw differences in velocity as large as 30 dB between sensor and bone. However, when dental cement (which is hard when dried) was applied over dried Jeltrate (which is soft), the bone and pressure sensor vibrated with similar magnitude and phase (Fig. 4, solid line).

As for intracochlear pressure measurements, we found that elimination of the relative motion between sensor and otic capsule by applying cement over Jeltrate yielded more consistent bone-conducted intracochlear pressures across specimens. During BC, differences in pressure measured with Jeltrate alone and Jeltrate plus cement were highly variable across experiments. Sometimes Jeltrate alone resulted in pressure magnitude measurements that were up to 10 dB higher or 30 dB lower compared to the Jeltrate-plus-cement condition. These differences between

Fig. 4. Ratio of the velocities of the pressure sensor and nearby bony cochlear surface, during BC stimulation. The magnitude and phase of the velocity ratio after sealing the pressure sensor to the bone at the cochleostomy with Jeltrate alone is plotted with dashed lines; the sensor is vibrating much less than the bone. After sealing the sensor-cochleostomy site with cement over Jeltrate, the ratios plotted with solid lines show that the sensor's vibration was tightly tied to the bone's vibration

Jeltrate alone and Jeltrate plus cement were likely due to artifactual difference in fluid motion with respect to the sensor motion (causing higher pressure measurement), or as a result of a leak in the Jeltrate seal (causing lower pressure measurement). It is possible that in some cases, the pressure measurements between Jeltrate and Jeltrate-plus-cement were similar because both increasing and decreasing effects were taking place and canceling, or because the Jeltrate was stiff enough to lock the motion of the sensor to the bony cochlear surface. These measurements corroborate the importance of ensuring that the sensors are hermetically sealed and cemented rigidly to the bony otic capsule so that the sensor moves in unison with the cochlear bone and fluid during the shaking imposed by BC stimulation.

Fig. 5. Comparison of the velocities of the cochlear promontory while the specimen was held by various methods during BC. When the specimen was held with a brass bar or temporal bone holder, the frequency response had large fluctuations (dotted and dashed lines). If the specimen was partially embedded in putty (Play-Doh), the frequency response was smoother (solid line)

Effect of Mechanically Securing the Specimen

During BC stimulation, varying how the specimen was held in position changed the character of the temporal bone vibration. When the specimen was held by a temporal bone holder sitting on a rubber ring base (illustrated in Fig. 1b), as often used for experiments with AC and round-window stimulation, large frequency-dependent fluctuations in the velocity of the promontory bone velocity were observed (Fig. 5, gray dashed line). Owing to suspicion that complex motions resulted from relative vibrations between the metal bowl and rubber ring, the temporal bone was then held tightly by a metal rod glued to the temporal bone (Fig. 1c), but we saw similar frequency-dependent fluctuations in magnitude and phase (Fig. 5, black dotted line). The temporal bone was then placed in a soft material that could mold to its shape and yet provide support (Play-Doh® modeling compound). This material held and supported the specimen in place in a stable manner (e.g., promontory and ossicular velocities were stable even after many hours), and allowed smoother frequency dependence of the promontory vibrations (Fig. 5; solid line). Also, *in situ* the skull is not rigidly held by its attachments at the neck. Therefore, embedding the specimen in soft putty results in Baha-evoked vibration profiles that are likely to be closer to the skull of a living subject or entire anatomical whole head spec-

imen (Banakis Hartl et al. 2016; Greene et al. 2015). However, the modes of vibration differ because the load impedance for the Baha is lower in the isolated temporal bone setup and the modes of vibration are also different as compared to a whole head comprising the entire skull and soft tissues. Results from the eight temporal bones described in the following sections were from experiments with specimens held by Play-Doh.

Intracochlear Pressures During BC

Magnitude and phase of velocity measurements of the cochlear promontory (V_{prom} , located between the pressure sensor sites) and V_{stap} evoked by the Baha BP 100 (with 40 mV_{RMS}) are plotted for a representative experiment in Fig. 6a, b. The phase of the velocity is referenced to the voltage stimulating the Baha. The bandwidth of these measured magnitudes was approximately between 200 Hz to 8 kHz, reflecting the band-pass characteristics of the vibrations produced by the Baha BP 100. The velocity magnitudes differed by about 30–40 dB between 200 Hz to around 1 kHz. Generally, the magnitudes and phases of V_{prom} and V_{stap} were similar across most frequencies; however, near 1 kHz, $|V_{stap}|$ was about 6–8 dB higher than $|V_{prom}|$. The phases of the promontory and stapes velocity were similar. The phase for the velocities decreases linearly for frequencies below 2000 Hz, which was used to estimate group delay (approximately 3.4 ms) between voltage input to the Baha and stapes or promontory velocity.

The magnitudes and phases of the intracochlear pressures evoked by BC in the same specimen are plotted in Fig. 6c, d. As shown in Fig. 6c, the magnitude of P_{SV} was generally higher than P_{ST} by about 10 dB. The overall shapes of the frequency responses of P_{SV} and P_{ST} were similar, unlike what was observed during AC where the shape of the P_{ST} magnitude was quite different and much lower in magnitude than that of P_{SV} (Nakajima et al. 2009). The cochlear input pressure drive, also referred to as the differential pressure (P_{Diff}), is the complex linear difference between P_{SV} and P_{ST} , and is plotted with dashed black lines in Fig. 6c. The magnitude of P_{Diff} was similar to the magnitude of P_{SV} because P_{SV} was significantly larger than P_{ST} for most frequencies. The bandwidth of BC-evoked intracochlear pressures ranged from around 200 Hz to 7–8 kHz, similar to that seen in the velocity responses. However, between 200 Hz and 1 kHz the pressure magnitude increased about 60 dB/decade, which was much larger than the increase seen in velocity of about 20 dB/decade, thus resulting in lower pressure responses at low frequencies. The pressure phases, plotted in Fig. 6d, were referenced to the input voltage to the Baha minus the

FIG. 6. Responses of velocity and pressure from a representative experiment excited by a Baha stimulated by 40 mV_{RMS}. **a, b** Velocity magnitude and phase of the cochlear promontory and the stapes. The phase (**b**) is referenced to the electrical voltage presented to the Baha; thus, the phase frequency response is mostly shaped by the

electronics of the Baha. **c, d** Pressure magnitude and phase of scala vestibuli (P_{SV}), scala tympani (P_{ST}), and the cochlear pressure drive (P_{Diff}). The phases (**d**) are referenced to the Baha input voltage minus the estimated delay of 3.4 ms between the voltage input and the velocity of the abutment of the titanium screw attached to the Baha

estimated delay of 3.4 ms between input voltage to the Baha and the velocity of the titanium screw abutment attached to the Baha (similar to the delay from voltage input to V_{prom} and V_{stap} discussed above). The phase for

P_{ST} was near 0 cycles at low frequencies and fluctuated around $-1/2$ cycle above 1 kHz. Due to the fluctuation in magnitude and phase within narrow frequency regions, the unwrapping of phase was uncertain. Overall, at many frequencies, especially above 1 kHz, the pressure phases were similar. The frequency responses of the phase for P_{SV} and P_{ST} were generally similar to that found in Chinchilla (Chhan et al. 2016).

Figure 7 shows velocity measurements of the promontory and Fig. 8 shows pressure measurements evoked by BC across all specimens (each specimen is designated with a colored line and their means with thick black lines). Figure 7 shows that the frequency response of the promontory velocity is generally similar across specimen with larger variation at lower frequencies (< 2 kHz), and is similar to the representative example of Fig. 6. Figure 8a, b plots the magnitudes of P_{SV} and P_{ST} frequency responses; they have similar shapes with P_{SV} approximately 10 to 15 dB higher than P_{ST} . The range in pressure among ears was about 20 dB. In some specimens, P_{ST} magnitudes exhibited narrow-band notches at various frequencies.

Fig. 7. Promontory velocity magnitudes evoked with BC (with Baha BP100 stimulated by 40 mV_{RMS}) for all specimens

FIG. 8. Intracochlear pressures during BC and AC. **a, b** Scala vestibuli and scala tympani pressures (P_{SV} and P_{ST}) evoked by BC (40 mV_{RMS} to Baha). **c, d** Input pressure drive P_{Diff} across the partition for BC and AC stimulation. The horizontal blue line shows that both P_{Diff}

are matched to be 20 Pa at 1 kHz (evoked by 40 mV_{RMS} to the Baha for BC and by 100 dB SPL at the ear canal AC). Pressures from individual ears are plotted with colored lines and the means are plotted with thick black lines

Figure 8c plots BC cochlear pressure input drive P_{Diff} for the individual ears and the mean. At 40 mV_{RMS} stimulation level to the Baha, the magnitude in BC P_{Diff} reached the maximum at around 1 kHz with a value of 20 Pa, which was equivalent to the maximum P_{Diff} at around 1 kHz evoked with AC of 100 dB SPL at the ear canal (Fig. 8d), represented by the blue horizontal lines in Fig. 8c, d. The comparison between AC and BC frequency responses demonstrates that the BP 100 has an output of limited bandwidth. This Baha's output steeply rolls off at frequencies below 0.8 kHz. The P_{Diff} slope below 1 kHz was 60 dB/decade for the BC response to iso-voltage stimulus to the Baha, as compared to 20 dB/decade for the AC response to iso-pressure stimulus at the ear canal. The BP 100 sound processor has difficulty stimulating sound pressure without distortions to the inner ear at low frequencies. Figure 8 also demonstrates that the range in P_{Diff} across ears are generally similar between BC (8c) and AC (8d).

To demonstrate how intracochlear pressures relate to vibrating structures, Fig. 9a, b plot intracochlear pressures versus V_{stap} during AC stimulation for a representative experiment. During AC stimulation,

P_{SV}/V_{stap} and P_{Diff}/V_{stap} magnitudes were generally flat with frequency and the phase was near zero, thus showing mechanical characteristics represented by resistance. P_{ST}/V_{stap} magnitude decreased with frequency then increased above 600 Hz, where the phase transitioned from around $-1/4$ cycle at low frequencies to about $+1/8$ cycle at high frequencies. These magnitude and phase frequency responses are similar to that described in previous studies (Nakajima et al. 2009; Stieger et al. 2013) and were modeled in Nakajima et al. (2009) with compliance, resistance and mass.

For BC, Fig. 9c, d plot intracochlear pressures versus promontory velocity (V_{prom}), and e & f plot pressures versus acceleration (A_{prom}). P_{SV}/V_{prom} and P_{Diff}/V_{prom} were similar with low-frequency magnitudes generally increasing about 60 dB per decade and phase near zero up to about 1 kHz, where the magnitude generally leveled off while the phase generally decreased with increasing frequency. At frequencies above 700 Hz, the normalized magnitude had significant narrow-band increasing and decreasing transitions with corresponding phase shifts. Magnitude of P_{ST}/V_{prom} was about 10 dB lower for most

FIG. 9. Example of the pressures (P_{SV} , P_{ST} , P_{Diff}) for AC and BC stimulation referenced to motion of the stimulus. **a**, **b** AC pressures are normalized to stapes velocity (V_{stap}), **c**, **d** BC pressures are normalized to the promontory velocity (V_{prom}), and **e**, **f** promontory acceleration (A_{prom})

frequencies (except the lowest and highest frequencies) with similar frequency dependent fluctuations compared to P_{SV}/V_{prom} . The BC response with respect to the acceleration of promontory bone is shown in Fig. 9e, f. P_{SV}/A_{prom} and P_{Diff}/A_{prom} magnitudes generally increased approximately 40 dB per decade for frequencies below 1 kHz and decreased with 20 dB per decade at higher frequencies (due to the 20 dB per decade difference between velocity and acceleration). The phases with respect to acceleration were shifted 1/4 cycle compared to those plotted against velocity, as expected. The phase of P_{ST}/V_{prom} and P_{ST}/A_{prom} were generally similar to the phase of P_{SV}/V_{prom} and P_{SV}/A_{prom} (Fig. 9d, f).

Results for all eight experiments in Fig. 10 show transfer functions of intracochlear pressures relative to stapes velocity for AC stimulation (similar to the representative case of Fig. 9a, b). The mean is shown in thick black lines. These AC driven transfer functions have similar behaviour to those in earlier reports (Nakajima et al. 2009; Stieger et al. 2013). It is notable that the magnitude of P_{ST}/V_{stap} exhibits a wide range at low frequencies across specimens; this shows that RW compliance can vary up to 20 dB across ears. There is also a range of frequencies across specimens where the magnitude minima occur and where the phase transitions from -0.25 to 0.125 cycles (corre-

FIG. 10. Intracochlear pressure measurements normalized to stapes velocity (V_{stap}) during AC stimulation for eight experiments

sponding to the frequency where the impedance changes from being dominated by compliance to dominated by mass effects, Fig. 10c, d). AC driven pressure with respect to V_{stap} is related to impedance: P_{SV}/V_{stap} the cochlear input impedance (Z_C) (Elliott et al. 2016; Nakajima et al. 2009; Olson 2001; Puria 2003; Ravicz et al. 2010), P_{ST}/V_{stap} the round-window impedance (Z_{RW}) of compliance at low frequency and fluid mass at high frequency (Nakajima et al. 2009; Olson 2001; Ravicz et al. 2010), and P_{Diff}/V_{stap} the cochlear partition impedance (Z_{Diff}) (Nakajima et al. 2009), with the assumption that only two windows significantly contribute to sound transmission during AC. These measurements allow for the quantification of mechanical properties such as resistance, compliance and mass, and how they contribute to sound transmission (calculated and computationally modeled in Elliott et al. (2016) and Nakajima et al. (2009).)

Results from all eight experiments during BC stimulation are plotted in Fig. 11. Magnitudes of intracochlear pressures are referenced to those of

promontory velocity (top row) and acceleration (middle row). Phases of the pressures are referenced to those of promontory velocity (bottom row). Means of magnitudes (black thick lines) were plotted for each frequency if the measurements from all eight ears had greater than 6 dB signal-to-noise ratios, which was typically the case for frequencies higher than 300 Hz. Compared to AC transfer functions, BC transfer functions showed relatively large narrow-band fluctuations in magnitude and phase (peaks and notches), and the frequencies where these fluctuations occurred varied across ears. A contributing factor to these large narrow-band fluctuations in the transfer function is that the frequencies where the fluctuations occurred in the motion of the promontory did not match the frequencies where the fluctuations occurred in the intracochlear pressures (e.g., compare Fig. 6a–c). Thus, the large fluctuations seen in the BC transfer functions do not represent the signal that the cochlear partition experiences; the P_{Diff} with respect to the input voltage to the Baha (Fig. 8c) is smoother and better represents the input pressure that the

FIG. 11. Intracochlear pressure measurements of eight experiments during BC stimulation. The first and second rows show magnitude of individual and mean pressures normalized to velocity and acceleration of promontory bone, respectively. The third row shows phase of individual pressures referenced to velocity of promontory bone

cochlear partition senses. A likely reason for the large fluctuations seen in transfer functions P_{Diff}/V_{prom} and P_{Diff}/A_{prom} (Fig. 11g, h) is the non-uniform three-dimensional frequency-dependent modes of motion of the temporal bone during BC stimulation (Stenfelt 2005b). A single point on the promontory for a velocity measure cannot represent the complex three-dimensional Baha-induced motion of the bone (Reinfeldt et al. 2013). For both AC and BC stimulation, the normalized P_{Diff} are similar to the normalized P_{SV} in that $|P_{SV}| > |P_{ST}|$. As can be appreciated in Fig. 11, for low frequencies (<1 kHz), the average BC pressures normalized to velocity P_{SV}/V_{prom} and P_{Diff}/V_{prom} (Fig. 11a, g) increase with frequency, while the pressures normalized to acceleration P_{SV}/A_{prom} and P_{Diff}/A_{prom} (Fig. 11b, h) are somewhat flat in comparison. For high frequencies (>1 kHz), the average P_{SV}/V_{prom} and P_{Diff}/V_{prom} are nearly flat, while P_{SV}/A_{prom} and P_{Diff}/A_{prom} decrease with frequency. The flattest response is seen for P_{ST}/A_{prom} across most frequencies. The phase of the BC transfer functions seems to show that, generally small delays occur between stimulus bone vibrations and the intracochlear pressures, though unwrapping the phase correctly is made difficult by the narrow-band fluctuations.

The ratios of pressures across the cochlear partition, P_{SV}/P_{ST} , (magnitude and phase) are plotted for AC and BC stimulation in Fig. 12. As in previous figures, the mean (thick black line) is only shown when all eight experiments had P_{SV} and P_{ST} magni-

tudes of at least 6 dB signal-to-noise ratios and there are reasonable certainties of phase unwrapping (which is difficult for low-frequency ratios during BC). The means of the ratios (in black) have similar overall shapes between AC and BC; the magnitude rises with increasing frequencies to approximately 15 dB around 1 kHz and decreases for higher frequencies. For AC, the phase is slightly above 0 cycles at low frequencies and increases to about +0.2 cycle near 250 Hz, then decreases to about -0.15 cycle around 2 kHz then goes back up to 0 cycles at higher frequencies (12 b). For BC, the phase has a similar shape: it is also approximately 0 cycles around 1 kHz and dips slightly for higher frequencies. However, for frequencies below 600 Hz in BC, the individual phases make large narrow-band transitions, making it difficult to unwrap the phase data. Differences between AC and BC ratios are seen in the details: In AC, the range across ears in P_{SV}/P_{ST} magnitude and phase at any one frequency is generally smaller than for BC, with individual BC responses exhibiting large narrow-band fluctuations across frequency. In AC, the variation across ears for frequencies below 1 kHz is larger than at higher frequencies. In BC, the ratio variations across ears are larger than in AC across all frequencies, where the BC narrow-band fluctuations contribute to the variation. Like AC, BC ratio have greater variation across ears for low than high frequencies. Overall, AC and BC scalae ratios are similar (Fig. 12e and f).

FIG. 12. Ratio of P_{SV}/P_{ST} magnitude and phase during AC (a, b) and BC (c, d) stimulation for eight experiments. Magnitudes are shown in upper panels, phases in lower panels. The mean value is only calculated when more than five recordings are above 6 dB signal-to-noise ratio. For direct comparison between AC and BC stimulation, the mean ratios of P_{SV}/P_{ST} are shown in panels e and f

DISCUSSION

BC evoked intracochlear pressures in human temporal bones were compared to AC evoked responses. To measure BC intracochlear pressures, we made necessary modifications to our technique: (1) reduced the complex motion of the bony capsule with moldable putty; (2) hermetically sealed, then secured the pressure sensor at the cochleostomy site by firmly gluing the sensor to the bone with hard-drying adhesive (dental cement). Cement allowed the sensor to vibrate in synchrony with the local otic capsule, eliminating various measurement artifacts; 3) performed intracochlear sensor calibrations for accurate pressure values throughout the experiment.

Air Conduction (AC) Stimulation Response

For AC stimulation, we assume that the oval window motion is the stimulus for sound transmission to the cochlea, and the volume velocity at the oval window generally flows through the RW membrane (Stenfelt et al. 2004a). Thus, we reference the intracochlear pressures to V_{stap} (Figs. 9 and 10). The ratio P_{Diff}/V_{stap} is relatively constant around 10^5 Pa/m/s in magnitude and 0 cycle in phase between 100 Hz to 6 kHz. We can also calculate cochlear partition impedance, which includes the effect of the helicotrema: $Z_{Diff} = P_{Diff}/(V_{stap} * S_{stap})$, [S_{stap} is the area of the stapes footplate]. Z_{Diff} data can be modeled as an acoustic resistance of approximately $20 \text{ G}\Omega$ (Pa s/m³) (Nakajima et al. 2009; Frear et al. 2018). RW impedance can be estimated as $Z_{RW} = P_{ST}/(V_{stap} * S_{stap})$. Z_{RW} data has been modeled as compliance at low frequencies of about $9 \times 10^{-14} \text{ m}^3/\text{Pa}$ (Nakajima et al. 2009; Frear et al. 2018) in series with a resistance of $2.5 \times 10^9 \text{ Pa s/m}^3$ and a mass of $1.0 \times 10^6 \text{ Pa s}^2/\text{m}^3$ (Elliott et al. 2016; Frear et al. 2018). However, the phase in this simple model reaches $\frac{1}{4}$ cycle (instead of the data's $\frac{1}{8}$ cycle) at frequencies above 0.5–1 kHz. Thus (Nakajima et al. 2009; Frear et al. 2018) modeled a distributed effect near the RW with a frequency-varying equivalent resistance and mass with an iterated Foster network (Nakajima et al. 2009; Frear et al. 2018), achieving the phase relationship of $\frac{1}{8}$ cycle at higher frequencies. Previous publications also refer to the cochlear impedance, Z_C , which is defined as: $Z_C = Z_{Diff} + Z_{RW}$ (Nakajima et al. 2009; Puria and Allen 1991; Shera 2007).

With the stimulus input at the oval window during AC, P_{SV} and P_{Diff} are directly related to V_{stap} ; however, P_{ST} is greatly influenced by the impedance of the RW. During AC stimulation in normal ears, V_{stap} is proportional to the cochlear input drive, P_{Diff} , consistent with the assumption that V_{stap} is a good estimate

of hearing, independent of frequency. This relationship can be appreciated by the frequency-independent constant magnitude and 0 cycles in phase across most frequencies seen in the transfer function of P_{Diff}/V_{stap} in Fig. 10e, f, which is related to Z_{Diff} . Such a frequency response is indicative of a pure resistance (Nakajima et al. 2009). On the other hand, the transfer function P_{ST}/V_{stap} is frequency dependent (Fig. 10) and is related to Z_{RW} . The P_{ST}/V_{stap} frequency response is indicative of a compliance dominant impedance at low frequency, a mass dominant impedance at high frequency, and resistive at the frequency in-between. There is variation across ears in compliance (low frequency), and resistance and frequency where dominance of compliance changes to mass, while the mass (high frequency) are more similar across ears. Impedances of the cochlea obtained from cochlear pressures and velocities are detailed in Frear et al. (2018).

Bone Conduction (BC) Stimulation Response

During BC, the magnitude of P_{SV} is higher than P_{ST} across most frequencies (e.g. Figs. 6 and 9), thus the cochlear input drive is similar to SV pressure ($P_{Diff} \approx P_{SV}$) for most frequencies. The BC frequency responses of P_{SV} and P_{ST} have similar overall shapes (narrow-band fluctuations, Fig. 6), likely due to similar influences.

Unlike AC where oval and round windows are major influences in sound transmission, during BC, sound is transmitted in a complex manner. The major input mechanism in BC is not at the oval window (although the effect of middle-ear inertia appears to contribute slightly around 1–2 kHz as shown in Fig. 6). Unlike AC, the volume velocities of the oval and round windows are not equivalent during BC (Stenfelt et al. 2004a). Other likely BC sound transmission mechanisms are the inertia of the inner-ear fluid, compression of the bony surroundings, and/or sound transmission through soft-tissue channels (e.g., vestibular and cochlear aqueducts, and/or other channels coursed by nerves and vasculature) (Chhan et al. 2013, 2016; Roosli et al. 2016; Stenfelt and Goode 2005a). We found that the BC magnitudes of P_{SV} and P_{ST} frequency responses have similar shapes, but the magnitude of P_{SV} is greater than P_{ST} , likely due to the relative difference in impedances influencing each scala. These findings can contribute to develop or validate BC computational models. For example, Stenfelt presented a lumped BC model for compression and fluid inertia based on theoretical calculations of P_{SV} and P_{ST} . (Stenfelt 2015).

In analyzing BC responses, a meaningful excitation reference is difficult to identify because there are likely a number of frequency-dependent BC mecha-

nisms that contribute to intracochlear acoustic pressures. One reference option is the stimulation voltage of the transducer (Fig. 6c, d, Fig. 8a and b and c). These pressure frequency responses are greatly influenced by the electromechanics of the BP 100 transducer. The magnitude of P_{Diff} frequency response, referenced to the Baha's input voltage, increases rapidly around 60 dB/decade for frequencies below 1 kHz and gently falls around 20 dB/decade above 1 kHz (Figs. 6c and 8c). However, the velocity of the stapes and promontory increases by about 20 dB/decade between 0.2 and 1 kHz (Fig. 6a). Thus, the low-frequency cut-off is sharper for P_{Diff} than V_{prom} below 1 kHz.

Comparison of AC and BC (Fig. 8c and d shows that the low-frequency (<1 kHz) AC P_{Diff} magnitude slope (referenced to 100 dB SPL at the ear canal) is about 20 dB/decade and BC P_{Diff} magnitude slope (referenced to 40 mV_{RMS}) is about 60 dB/decade. This comparison in Fig. 8c and d, where the maximum P_{Diff} magnitudes are equivalent for both AC and BC, shows that the hearing obtained with an iso-voltage driven Baha drastically drops with decreasing frequency, thus is band-limited as compared to AC hearing with iso-intensity ear-canal pressure.

To better understand the relation between the BC motion and the pressures in the cochlea, we can reference BC intracochlear pressure to velocity, as we do in AC (where we reference to V_{stap}). However, there is no ideal choice of a dominant velocity reference point during BC because the overall motion of the otic capsule acts as the input (Eeg-Olofsson et al. 2013; Reinfeldt et al. 2013). When pressures are referenced to V_{prom} , the referenced pressure frequency responses have multiple narrow-band fluctuations due to the three-dimensional complex motion of the bone, which is not captured in our one-directional velocity measurements (Stenfelt and Goode 2005b). Furthermore, our isolated temporal bone preparation in putty may not attenuate these sharp vibrations relative to frequency as compared to a temporal bone in the natural environment of a whole head (Banakis Hartl et al. 2016; Farrell et al. 2017). The overall trend of the frequency response of P_{SV}/V_{prom} and P_{Diff}/V_{prom} increases 20 dB/decade below 1 kHz, while the response is flatter between 1 and 4 kHz (Fig. 11a, g). Thus on average, the velocity of the promontory between 1 and 4 kHz is proportional to BC hearing. This is consistent with Eeg-Olofsson et al. (2013) who reported correlation between the movement of the otic capsule and BC threshold for frequencies higher than 1 kHz in patients with radical cavity.

We can also reference BC pressure response to the promontory acceleration A_{prom} , which is related to force. Acceleration has been used by other investigators as an input estimate for BC experiments (Stenfelt

and Goode 2005b; Stenfelt et al. 2000). Figure 11h shows that on average P_{Diff}/A_{prom} is almost flat for frequencies below 1 kHz, while it decreases around 1–4 kHz and stays relatively flat for frequencies above 4 kHz. The relatively flat averaged P_{Diff}/A_{prom} response for frequencies below 1 kHz implies that the acceleration of the promontory may be used to predict BC hearing below 1 kHz. However, the large variation across individual measurements (up to 40 dB) as well as their individual shapes deviates from such a simple relationship. Measurement of three-dimensional motion of the cochlear promontory might be useful to clarify the relation between the motion of the bony shell and the intracochlear sound pressure.

Influence of Impedances at the Scalae

An indirect way to look at the relative influences of impedances at each scalae is to look at the ratios of the scalae pressures. Comparison between AC and BC pressure ratios across the partition, P_{SV}/P_{ST} , reveal similarities (Fig. 12). Overall, magnitudes of the ratios (P_{SV}/P_{ST}) with respect to frequency have similar shapes and values between AC and BC, with a general peak around 1 kHz (12e). The phases of the P_{SV}/P_{ST} ratios also have similar frequency dependence between AC and BC (Fig. 12f; however, the BC phase is difficult to unwrap below 600 Hz). At frequencies below 1 kHz, there is a noticeable difference in the averages of both magnitude and phase (Fig. 12e, f). Perhaps the slight differences in P_{SV}/P_{ST} magnitude seen around 400–1000 Hz are influenced by the different sound transmission pathways through differing impedances during AC and BC. Above 1 kHz, both AC and BC have similar ratios, thus possibly similar impedance influences for sound transmission.

During AC, the volume velocity that passes through the oval window (stapes) and RW are about the same. Within the cochlea, the volume velocity is divided longitudinally across the cochlear partition (as well as the helicotrema) before flowing through the RW. This volume velocity within the cochlea is lumped as U_{Diff} and the impedance as Z_{Diff} . We then have the relationship:

$$\begin{aligned} U_{Diff} &= P_{Diff} / Z_{Diff} = (P_{SV} - P_{ST}) / Z_{Diff} \approx P_{SV} / Z_{Diff} \\ &\quad (\text{because } P_{SV} > P_{ST}), \\ U_{RW} &= P_{ST} / Z_{RW}, \\ U_{Diff} &\approx U_{RW}, \\ \text{Therefore, } P_{SV} / P_{ST} &\approx Z_{Diff} / Z_{RW}. \end{aligned}$$

At frequencies below ~700 Hz, where a compliant component in Z_{Diff}/Z_{RW} appears to dominate, the AC driven magnitude of scalae ratio P_{SV}/P_{ST} increases with increasing frequency, but varies considerably

across specimens (Fig. 12a). This large variation in P_{SV}/P_{ST} ($\approx Z_{Diff}/Z_{RW}$) is also seen in P_{SV}/V_{stap} ($= Z_{RW} \cdot S_{stap}$, where S_{stap} is the surface area of the stapes footplate, plotted in Fig. 10c), likely due to large variation in RW compliance across ears. The variation seen might also be due to slight differences in position of the pressure sensors across experiments. Above ~ 700 Hz, where a mass component in Z_{Diff}/Z_{RW} appears to dominate, magnitudes of P_{SV}/P_{ST} decrease similarly across specimens (Fig. 12a), as does the increase in P_{ST}/V_{stap} (Fig. 10c).

During BC, the average ratio of P_{SV}/P_{ST} at high frequencies (> 1 kHz) is similar to that evoked by AC, although the individual ratios have large narrow-band fluctuations that vary in frequency across specimens (Fig. 12c). One possible explanation for the similar pressure ratios for AC and BC at high frequencies (> 1 kHz) is the similar mechanism responsible in generating intracochlear sound pressures. A possibility is that BC-evoked relative motion of the stapes with respect to the promontory and the impedances affecting this mode of transmission may contribute to P_{SV}/P_{ST} , similar to AC. On the other hand, P_{SV}/P_{ST} in BC at low frequencies (< 1 kHz) is different from that in AC (Fig. 12e and f), and may reflect differing sound transmission mechanisms between BC and AC. Unique mechanisms in sound transmission during BC, such as cochlear fluid inertia and compression of the surrounding cochlear bone, relies on the oval window and middle ear impedances (looking out from SV near the oval window, referred to as reverse middle-ear impedance (Puria 2003; Stieger et al. 2013). Other impedance channels, such as the vestibular aqueduct and cochlear aqueduct, also likely contribute to BC sound transmission (Stenfelt 2015; Stenfelt 2016).

Limitations

We used isolated fresh human temporal bones embedded in putty, resulting in effective mass that is lower than a whole head held by a compliant neck. Additionally, the Baha implanted on the outer bone of the temporal bone is physically closer to the otic capsule and not positioned in the same location as in patients. The different condition between our experimental temporal-bone setup and a whole head likely result in differences in frequency-dependent three-dimensional vibrational motion during BC.

CONCLUSION

To understand the mechanisms responsible for BC and to compare to AC, intracochlear pressures (P_{SV} ,

P_{ST} , and the cochlear input drive P_{Diff}) were measured in conjunction with ossicular and promontory motion. Appropriate methodologies were developed to prevent inaccuracies of intracochlear pressure measurements that can arise during BC stimulation. Dissimilarities found between AC and BC pressure responses reflect the different mechanisms of sound transmission during BC that are more complex compared to the simple mechanism during AC.

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COMPLIANCE WITH ETHICAL STANDARDS

Conflict of Interest The authors declare that they have no conflict of interest.

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