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Comparison of forward (ear-canal) and reverse (round-window) sound stimulation of the cochlea

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Abstract

The cochlea is normally driven with “forward” stimulation, in which sound is introduced to the ear canal. Alternatively, the cochlea can be stimulated at the round window (RW) using an actuator. During RW “reverse” stimulation, the acoustic flow starting at the RW does not necessarily take the same path as during forward stimulation. To understand the differences between forward and reverse stimulation, we measured ear-canal pressure, stapes velocity, RW velocity, and intracochlear pressures in scala vestibuli (SV) and scala tympani (ST) of fresh human temporal bones. During forward stimulation, the cochlear drive (differential pressure across the partition) results from the large difference in magnitude between the pressures of SV and ST, which occurs due to the high compliance of the RW. During reverse stimulation, the relatively high impedance of the middle ear causes the pressures of SV and ST to have similar magnitudes, and the differential pressure results primarily from the difference in phase of the pressures. Furthermore, the sound path differs between forward and reverse stimulation, such that motion through a third window is more significant during reverse stimulation. Additionally, we determined that although stapes velocity is a good estimate of cochlear drive during forward stimulation, it is not a good measure during reverse stimulation.

Keywords

Round-window stimulation; cochlear pressure; scala vestibuli; scala tympani

1. INTRODUCTION

Acoustic energy is normally transmitted to the cochlea by “forward” stimulation, in which sound presented to the ear canal elicits vibration of the middle-ear ossicles, producing a pressure difference across the cochlear partition that results in a traveling wave (Dancer et al., 1980; Nedzeltnitsky, 1980; Lynch et al., 1982; Voss et al., 1996). Likewise, acoustic energy can be transmitted to the cochlear partition by “reverse” stimulation, in which

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mechanical vibration presented to the round window (RW) results in a pressure difference across the cochlear partition (Wever et al., 1948; Wever et al., 1950; Shera et al., 1992; Spindel et al., 1995; Voss et al., 1996; Nakajima et al., 2010b; Nakajima et al., 2010a).

RW stimulation utilizing modified middle-ear active transducers has improved various conductive and mixed hearing losses in patients; however, great variability in the efficacy of this approach has been reported (Martin et al., 2009; Baumgartner et al., 2010; Bernardeschi et al., 2011; Colletti et al., 2011; Mandala et al., 2011; Rajan et al., 2011; Iwasaki et al., 2012; Verhaegen et al., 2012). A number of studies have tried to address this concern by determining how well the RW actuator couples its motion to the RW membrane (Arnold et al., 2010b; Arnold et al., 2010a; Nakajima et al., 2010a; Pennings et al., 2010; Tringali et al., 2010; Schraven et al., 2011; Schraven et al., 2012). Schraven et al. (2011 & 2012) carefully studied various actuator tips and the need for pre-load to the RW and reported generally reproducible performance. Often in these studies, stapes motion is used as an estimate of cochlear drive.

Determination of the mechanism and characteristics of sound transduction for reverse RW stimulation as compared to forward ear-canal stimulation has important implications. Stapes motion is a good estimate of the input to the cochlear partition during forward sound stimulation. This relationship relies on the assumption that the cochlear shell is a rigid body filled with incompressible fluid, with the oval and round windows as the only major flexible components by which volume velocity can enter and leave the inner ear. Evidence for this two window hypothesis in forward stimulation was shown by measuring near-equal volume velocities of the oval and round windows (Kringelbotn, 1995; Stenfelt et al., 2004; Sim et al., 2012). However, it is unknown whether the two-window hypothesis holds during reverse stimulation, and this has important implications for whether or not stapes motion can be used as a good estimate of sound transduction in reverse stimulation. Differences in how the stapes velocity relates to intracochlear pressures and sound transduction in forward and reverse stimulation are possible because the impedances that the acoustic stimuli experience in the forward and reverse directions differ, as was shown by Puria (2003). Such differences in the impedance that the sound acts against as it flows through the ear in the forward and reverse directions influence sound flow, and differences in the relative impedances between possible paths may allow sound to follow different paths during reverse stimulation. Thus, whether the motion of the stapes is a good indicator of reverse stimulation is unknown.

The aim of this study is to understand the mechanics of how sound stimulates the inner ear in forward and reverse stimulation. We measure intra-cochlear pressures in scala vestibuli (P_{SV}) and scala tympani (P_{ST}) simultaneously, and calculate the differential pressure, which is the drive for the motion of the cochlear partition. We compare these pressure measurements with measurements of ossicular motion during forward and reverse stimulation. Additionally we use these measurements to determine whether or not there are only two significant windows (paths for volume velocity to enter or exit the inner ear) during reverse stimulation, as has been shown for forward stimulation.

2. METHODS

Most of the methods used in this study are similar to that described in previous publications (Nakajima et al., 2009; Nakajima et al., 2010b; Nakajima et al., 2010a). Therefore brief descriptions of most of the methods follow, with the exception of detailed descriptions of techniques new to this study.

2.1 Preparation of temporal bones

Sixteen human temporal bones were harvested within 24 hours of death, where 9 of the specimens were used fresh and 7 were used after having been frozen. Prior to specimen preparation, the specimens were stored at 4°C in 0.9% normal saline. The external auditory canal was shortened to approximately 1 cm and a mastoidectomy was performed with a wide opening of the facial recess, without entering the inner-ear fluid space or compromising anatomical structures such as the semicircular canals and the endolymphatic sac. The RW area was generously exposed by reducing the bony overhang around the perimeter to approximately 1 mm, facilitating the visualization of the coupling of the actuator to the RW membrane. Cochleostomies (0.2 mm diameter) were made (under fluid to prevent air from entering the cochlea) in scala vestibuli and scala tympani where the pressure sensors were inserted into the cochlear fluid approximately 100-150 µm from the bony surface. The fluid of the middle ear was then removed except locally around the pressure sensors, and the sensors were sealed to the surrounding bone of the cochleostomy sites with Jeltrate (L.D. Caulk Co.).

The anatomy of the RW membrane appears to vary considerably across ears. We have on occasion encountered a thin bony ledge adherent to the RW membrane encompassing 1/3-1/2 of the area of the RW membrane. We have also seen a similar thin bony ledge that encompassed the majority of the RW membrane area, and an attempt to remove it resulted in rupture of the RW membrane area, as it appeared that the thin bone either was adherent to or replaced the RW membrane.

2.2 Measurement Setup

2.2.1 Forward and reverse stimulation—Figure 1 illustrates our experimental system demonstrating the stimuli and recordings made in the inner and middle ear. For forward stimulation, sound was applied to the ear canal with a loudspeaker (40-1377, Radio Shack). Seventy-four logarithmically-distributed 10 ms tones between 0.1-20 kHz, with levels between 50 and 115 dB, were presented (responses averaged 50 times). The sound pressure level at the ear canal (P_{EC}) was recorded with a probetube microphone (ER 7C, Etymotic) with the tip located 1-2 mm from the umbo of the tympanic membrane. A coupler for the loudspeaker and microphone was sealed to the external auditory canal.

For reverse stimulation, two different types of actuators were used to mechanically stimulate the RW (Fig. 1). In the first set of measurements ($n=2$) a floating mass transducer (FMT) (Vibrant Soundbridge, MED EL) was coupled to the round window membrane via a soft-tissue interface (Colletti et al., 2006; Arnold et al., 2010a; Nakajima et al., 2010b). However, in order to study reverse stimulation reliably for frequencies below 1 kHz, a second set of experiments was performed with a custom-made piezoelectric actuator with a transparent glass coupling rod of 1 mm diameter. The transparent rod tip, smoothed around the edges and covered with a thin layer of clear nail polish, was directly coupled to the RW membrane or coupled with an interfacing transparent disk (1.5 mm diameter) punched out of an eye contact lens. The transparency of the structure(s) allowed visual feedback of when the actuator made contact with the RW membrane. After contact with the RW, the actuator rod was advanced another 20 to 50 µm towards scala tympani until increase in intracochlear pressures were no longer observed. This piezoelectric actuator with rod tip enabled linear and reliable reverse stimulation for frequencies between 100 Hz and 10 kHz. The actuator for reverse stimulation was driven at the same frequencies as the loudspeaker during forward stimulation. It is important to note that the velocity of the actuator is not necessarily the velocity of the RW membrane due to inefficient coupling of energy from actuator to RW membrane, thus the effective RW volume velocity cannot be accurately measured.

2.2.2 Pressure sensors—Intracochlear pressures in the SV and the ST were recorded with fiberoptic micro-scale 167 μm diameter pressure sensors (Olson, 1998) in the manner described in (Nakajima et al., 2009). The sensors were fabricated with hollow glass fibers terminated by a gold-coated diaphragm (1 μm thick) at one end, with a 'Y'-shaped optical fiber coupler melded to the open end. Light from a LED source transmitted through the optical fiber was reflected back to a photosensor by the gold-coated diaphragm. This diaphragm moves with changes in pressure, modifying the amount of reflected light returning to the fiber. Because sensor sensitivity was prone to changes during use, all sensors were calibrated just prior to cochlear insertion and immediately after removal from the cochlea. The sensitivity changes affected the calibration magnitude but the phase remained stable. Experiments with a drift in sensor calibration greater than 3.5 dB were excluded.

2.2.3 Velocity measurements—Simultaneous to the pressure recordings in scala vestibuli, scala tympani and the external auditory canal, vibrational velocities of the stapes (posterior crus), RW membrane, actuator or promontory were also measured with a single-point laser Doppler vibrometer system (Polytec CLV), with $\sim 0.1\text{-}0.2\text{ mm}^2$ reflective tape used to improve reflection. To enable similar alignment of the laser for each measurement, the position of the supporting microscope was marked and all microscope settings were held constant. 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) of the stapes. The phase difference between stapes and round window velocity was checked to be 0.5 cycles below 500 Hz after the sensors were inserted and sealed, to check whether air was introduced into the cochlea and whether the seals surrounding the sensors were intact (Kringelbotn, 1995; Stenfelt et al., 2004; Nakajima et al., 2009). The movement of the promontory was measured during forward and reverse stimulation to determine the level of noise vibration (artifact). Actuator velocity was measured at the side of the cylindrical glass tip or the side of the FMT with an angle ($\sim 20\text{-}30$ degrees) relative to the direction of the actuator motion.

2.3 Data acquisition and analysis

A four-channel data-acquisition board (PXI, 6122, National Instruments) was used for simultaneous recordings of the pressure in the external auditory canal, the scala vestibuli, the scala tympani and the velocity of one of stapes, RW membrane, cochlear promontory or actuator. For each stimulus frequency, the response of all channels was averaged 50 times prior to transformation into the frequency domain. The magnitudes and phases at the stimulation frequency were selected and assembled into one frequency response using custom written LabView® software.

Data were analyzed using Matlab (R2011a) software. All pressure measurements with signal-to-noise ratio (SNR) below 6 dB were excluded. The sensor noise was determined by measuring the pressures in the identical manner as during normal recordings, with the exception of disconnecting the electrical input for the stimulus: This noise measurement captured inadvertent electrical pickup at the fundamental frequency. We also excluded velocity measurements that were less than 6 dB greater than the velocity of the cochlear promontory (what we call artifact) for the corresponding stimulation (ear-canal sound or RW actuator). This occurred for example when we encountered round windows with a thin bony ledge (as mentioned above), which produced high artifact measurements during reverse stimulation. Furthermore, in experiment No. 106, nine data points between 260 and 480 Hz were interpolated for SV pressure recordings because of an unexplained narrow-band resonance of the pressure sensor recorded during calibration.

To calculate the input to the cochlea – the differential pressure across the cochlear partition – the pressures in SV and ST are subtracted as vectors (i.e., as complex numbers in the frequency-domain representation). Figure 2 shows two examples of vector representations of P_{SV} and P_{ST} phasor vectors and their differential pressures, $\Delta P = (P_{SV} - P_{ST})$ for forward and $\Delta P = (P_{ST} - P_{SV})$ for reverse stimulation. Interpretation of this vector representation will be described in the Results section.

Because there is uncertainty of the absolute intracochlear pressure measurement imposed by deviation of the pressure sensor sensitivity, we estimated the maximum possible error in the differential pressure calculation. We varied independently the magnitude of the vectors P_{SV} and P_{ST} in 9 step sizes within the range of their change in sensitivity. (The phase of the sensor calibrations were stable.) Then we calculated the differential pressure for each pair of possible combination of P_{SV} and P_{ST} pressures.

Due to pressure sensor instability and other complications faced during some experiments, several experiments were excluded from the study for analyses. A summary of all experiments including sensor stability, complications, and reasons for exclusion of any data is shown in table 1.

3. RESULTS

3.1 Pressures in scala vestibuli and scala tympani

Intracochlear pressures in SV and ST were measured simultaneously along with the velocity of the stapes or RW actuator to obtain the transfer function of the intracochlear pressures referenced to their stimulus velocities. Figure 3 shows comparisons of SV and ST pressures elicited by forward ear-canal stimulation and reverse RW stimulation for a representative experiment. The intracochlear pressures were referenced to stapes velocity (V_{Stap}) for forward stimulation, and RW actuator velocity (V_{Act}) for reverse stimulation.

During forward stimulation, the magnitude for SV was generally bigger than ST for a wide frequency range (between about 300 Hz and 4 kHz). The large difference in pressure magnitude was due in part to the high compliance of the round window membrane, which functions as a low-impedance path for pressure release, allowing for low ST pressure (Nedzelitsky, 1980; Lynch et al., 1982; Nakajima et al., 2009).

During reverse stimulation, the SV and ST pressure magnitudes were similar for frequencies below 1 kHz, but at higher frequencies, ST was of significantly higher magnitude (Fig. 3c). For reverse stimulation, the stimulus was now at ST and the load included the SV where the fluid encountered the high impedance of the middle-ear (as opposed to the highly compliant RW which acted as the load during forward stimulation). The pressures in the scalae evoked by reverse stimulation for a given actuator velocity were not generally higher for reverse stimulation as compared to forward stimulation, as would be expected with the high middle-ear load during reverse stimulation. This was likely caused by the difference in drive area with reverse stimulation as compared to forward stimulation, as well as the loss of drive due to inefficient coupling between the actuator and RW. The RW volume velocity cannot be measured accurately.

For a given type of stimulation, the responses in intracochlear pressures are generally similar across multiple ears, though greater variability was observed for reverse stimulation than for forward stimulation. Figure 4 shows the SV and ST pressures with respect to their stimulus velocities for 8 ears, where each colored line represents a specific ear. The black line represents the geometric mean and the gray shaded region is one standard deviation around the mean. For forward stimulation, the SV pressure magnitudes are nearly flat across

frequency (Fig. 4a) and the phase is near 0 for most frequencies (Fig. 4b), consistent with a resistance-dominated system. The ST magnitude decreases as frequency increases, then increases again at around 400–700 Hz (Fig. 4b). The frequency where the magnitude has a local minima and the phase makes a transition from -0.25 cycle to $+0.25$ cycle varies across ears. The phase of ST at low frequencies (below 300–600 Hz) is approximately -0.25 cycle, consistent with a compliance-dominated system, while above around 800 Hz the phase is approximately $+0.25$ cycle, consistent with a mass-dominated system.

During forward stimulation, P_{ST}/V_{Stap} magnitude and phase with its characteristic pattern (Figs. 3a,b & 4c,d) – a dip in magnitude and 0.5 phase shift around 300–700 Hz – were observed. This characteristic pattern was discussed in detail in Nakajima et al. (2009). In that study, we were able to closely model the measured round-window impedance, $Z_{RW} = P_{ST}/U_{Stap}$. We found that at low frequency (below the dip in magnitude), Z_{RW} behaves as a compliance (approximately $9 \times 10^{-14} \text{ m}^3/\text{Pa}$), and above the magnitude dip, Z_{RW} becomes dominated by inertia and resistance. However, we found that the higher frequency behavior was not well modeled with a simple lumped parameter model as Nedzelnitsky (1980) had attempted in cat. We were able to closely model the behavior by a distributed parameter system such that the effective resistance increases as the square root of frequency and the effective inductance/mass decreases as one over the square root of frequency, consistent with distributed inertia and energy dissipation of cochlear fluids.

During reverse stimulation (Fig. 4 e,f,g,h), the pressures elicited in SV and ST differed from those found with forward stimulation. Within each ear, the pressure magnitudes in SV and ST were more similar during reverse stimulation than they were in forward stimulation. This similarity was likely due to the high impedance of the middle ear faced by the fluid flow at the oval window, which maintains the SV pressure at a relatively high magnitude. The relatively higher variability of pressures measured in different ears during reverse stimulation was likely due in part to differing actuators and how the actuator couples to the RW membrane. For example, the 2 experiments utilizing the FMT actuator (dashed lines) with an interface of soft tissue at the RW produced larger pressure responses in the mid-to-high frequencies as compared to the pressures elicited by actuators with a 1-mm diameter glass rod tip (with and without a contact-lens interface). The SV and ST pressure phases were generally near -0.25 cycles (compliance dominated) for most experiments.

3.2 Differential pressure

Vector representation of intracochlear pressures are illustrated in Fig. 2, as described in the Methods section. For forward stimulation, P_{SV} was a much larger vector than P_{ST} , thus the magnitude of the differential pressure, $|\Delta P| = |(P_{SV} - P_{ST})|$, was similar to the magnitude of P_{SV} and was generally insensitive to the phase relationship between P_{SV} and P_{ST} . On the other hand, for reverse stimulation, the magnitudes of P_{SV} and P_{ST} were similar, and the magnitude $|\Delta P| = |P_{ST} - P_{SV}|$ depended greatly on the phase relationship between P_{SV} and P_{ST} .

Differential pressures with respect to the stimulus velocities are plotted in Fig. 5 for one representative ear. Across frequency, the magnitude of the differential pressure with respect to the stapes velocity, $\Delta P/V_{Stap}$ during forward stimulation was higher than the differential pressure produced by similar velocities of the RW actuator during reverse stimulation $\Delta P/V_{Act}$. The shaded area illustrates a region of uncertainty due to variations in the pressure sensor calibration. The calculations of these uncertainties were detailed in the Methods section. The example error analysis shows that the difference between forward and reverse stimulation for these calculations was credible across frequency. Only data obtained with stable sensors throughout the experiment and with resulting error analyses for calculated results that allowed for separation of effects between stimuli are presented (Table 1).

The relationships between differential pressures with respect to the stimulus velocities are shown in Fig. 6 for 8 ears. Figure 6 a & b plots the magnitude and phase of the differential pressure with respect to stapes velocity ($\Delta P/V_{Stap}$) during forward stimulation, and Fig. 6 c & d plots magnitude and phase of the differential pressure with respect to RW actuator velocity ($\Delta P/V_{Act}$) during reverse stimulation. The forward stimulation case is related to the forward differential impedance, and has magnitude that was generally flat across frequency and phase that remained near 0 cycles for most frequencies, suggesting a resistive system, similar to that reported in (Nakajima et al., 2009). The magnitudes of the differential pressure with respect to the stimulus velocities were higher for forward stimulation compared to reverse stimulation, and this comparison varied across ears and across actuators. This relationship likely depended on the difference in drive area, how the actuator couples to the RW, as well as third-window fluid leakage that was significant during reverse stimulation (explained in the Discussion section). Furthermore, the variability of the reverse stimulation case was greater than in forward stimulation, suggesting variability in the effective drive area of the RW, the coupling of the actuator to the RW membrane, and possibly variability of third-window effect across ears.

As can be appreciated in Figs. 5 & 6, $|\Delta P/V_{Act}|$ during reverse RW stimulation is generally lower than $|\Delta P/V_{Stap}|$ during forward ear-canal stimulation, sometimes by large amounts. With sufficient drive available from an actuator, however, sufficient differential pressure can be generated. Thus, this merely expresses the fact that the actuator must be able to deliver significantly more flow than that developed by stapes motion for similar levels of differential pressure.

3.3 Impedances of the cochlea and the middle ear

The cochlear impedance can be estimated by measuring SV pressure with respect to the stapes volume velocity, $Z_C = P_{SV}/U_{Stap}$ during forward stimulation. The stapes volume velocity was calculated as $U_{Stap} = V_{Stap} * A$, where $A = 3.2 \text{ mm}^2$ (Aibara et al., 2001) is the area of the footplate. Likewise, the middle-ear impedance can be estimated by SV pressure with respect to stapes volume velocity, $Z_{ME} = P_{SV}/U_{Stap}$ during reverse stimulation. Z_C and Z_{ME} were also defined and estimated by Puria (2003) by a different method. Figure 7 plots both Z_C and Z_{ME} for several of our experiments and Puria's data (dashed dark gray line), showing that the middle-ear impedances vary across ears. Five of our experiments had a 6 dB signal-to-noise ratio or better for stapes velocity across a wide frequency range during reverse stimulation. Similar to Puria (2003), magnitudes of Z_{ME} were generally 10-30 dB higher than Z_C at low frequencies, and were similar above about 2 kHz.

3.4 Relationship between stapes velocity and the cochlear input (differential pressure)

Figure 8 plots the stapes velocity with respect to the differential pressure during forward and reverse stimulation across ears. Five experiments had over 6 dB SNR for stapes velocity across a wide frequency range during reverse stimulation. The plot shows that the magnitude of $V_{Stap}/\Delta P$ for forward stimulation is higher than the magnitude of $V_{Stap}/\Delta P$ for reverse stimulation. That is, for the same cochlear drive (differential pressure across the partition ΔP), reverse stimulation results in less stapes velocity than during forward stimulation. This difference in magnitude is likely due to a third window in SV that becomes significant in the reverse stimulation case, as will be explained in the Discussion section. This relationship also varies across ears possibly due to variations in the effect of the SV third window. Because we reference to the differential pressure across the partition, the comparison is independent of the coupling of the RW actuator.

4. DISCUSSION

4.1 The cochlea and middle-ear system for forward and reverse stimulation

4.1.1 Forward Stimulation—An illustration representing the cochlear system during forward stimulation is shown in Fig. 9a. During forward stimulation, the volume velocities at the oval window (U_{Stap}) and RW (U_{RW}) have been measured to be nearly equal (Stenfelt et al., 2004). Therefore, the volume velocity through the basilar membrane, meaning through the whole partition, (U_{BM}) is also about the same as through the oval window and the RW; that is, $U_{Stap} \approx U_{RW} \approx U_{BM}$. Alternate sound paths, such as physiological third windows (paths for leakage flow), may exist within SV and ST but have low volume velocity and we may assume that $|U_{Lk_ST}/U_{BM}| \ll 1$ and $|U_{Lk_SV}/U_{BM}| \ll 1$. Third windows could include structures such as neuro-vascular channels, vestibular aqueduct, cochlear aqueduct, etc. Hence the impedance of the RW is much lower than the leakage impedance of any ST third-window, $|Z_{RW}| \ll |Z_{Lk_ST}|$, and the cochlear impedance is much lower than the impedance of the SV leakage impedance; that is, $|Z_C| = |Z_{BM} + Z_{RW}| \ll |Z_{Lk_SV}|$.

4.1.2 Reverse Stimulation—Figure 9b is an illustration representing the system during reverse RW stimulation. For reverse stimulation, volume velocity through a RW interface leakage path, U_{Lk_RW} , can occur at the RW membrane area due to inefficient coupling of the RW actuator to the RW membrane, as well as outward volume velocity through deformation of the compliant RW membrane where the actuator is not touching the membrane (that is, $|U_{Act}| > |U_{BM}|$) (Schraven et al., 2011). Furthermore, because the flow of sound during reverse stimulation faces a different set of impedances as compared to forward stimulation, third-window leakages at ST (U_{Lk_ST}) and SV (U_{Lk_SV}) may be significant. Our experimental measurements can describe the leak at SV (U_{Lk_SV}) but not the leak at the RW and the ST (U_{Lk_RW} and U_{Lk_ST}).

4.2 Comparison of intracochlear pressures evoked by forward and reverse stimulation

During forward stimulation (Figs. 3 & 4), a large pressure difference between SV and ST occurs due to the low impedance of the RW membrane, $|Z_{RW}|$ relative to $|Z_{BM}|$. Thus, the magnitude of the differential pressure ΔP depends largely on the magnitude of P_{SV} owing to the large difference in magnitudes between SV and ST pressures. During reverse stimulation (Figs. 3 & 4), the magnitudes of the pressures in SV and ST can be similar, especially at low frequencies, due to the large middle-ear impedance that the flow of sound faces ($|Z_{ME}| \gg |Z_C|$), where Z_{ME} is measured with reverse stimulation and Z_C is measured with forward stimulation, and $Z_C = Z_{BM} + Z_{RW}$, as shown in Fig. 7 and described in the Results section). The similarity in amplitude between P_{SV} and P_{ST} for reverse stimulation makes their difference depend critically on the phase difference between the two pressures.

Differential pressure with respect to stapes velocity during forward stimulation is significantly larger than the differential pressure with respect to the RW actuator velocity during reverse stimulation (Fig. 5). The low differential pressure with respect to stimulus velocity during reverse stimulation is likely due to the difference in the area of the effective drive that the actuator imparts on the RW and the area of the stapes (e.g. the area of the stapes can be about 3 times the area of the glass rod actuator), as well as the inefficient coupling of the actuator to the RW membrane, as depicted as a volume velocity leak, U_{Lk_RW} , in Fig. 9b. However, significant third-window fluid motion during reverse stimulation also appears to contribute to the low reverse differential pressure. We were able to show this effect by focusing the analysis of the system to only the section from the partition impedance Z_{BM} , where we measure ΔP , to the stapes where we measure its motion (detailed in the following section).

4.3 Is there a third window leak? Is stapes motion a good estimate of cochlear drive?

The question of whether there is a significant fluid flow through a physiological third window is also related to the question of whether stapes motion can be used as an estimate of cochlear drive. In this analysis, we focus on the system between the cochlear partition and the stapes where we reference our analysis to ΔP , thereby compensating for differences between the velocity of the actuator and the stapes as well as the coupling of the actuator to the RW. During forward stimulation (Figure 9a), we assume $|Z_{Lk_SV}| \gg |Z_{BM} + Z_{RW}|$, $|U_{Stap}| \approx |U_{BM}|$, making stapes velocity proportional to the volume velocity and differential pressure across the partition.

We also assume that the impedance across the partition, Z_{BM} , is the same during both forward and reverse stimulation (see Fig. 9). Therefore, during both forward and reverse stimulation, the differential pressure across the partition $\Delta P = (P_{SV} - P_{ST})$ is proportional to the volume velocity across the partition U_{BM} with the same proportionality constant Z_{BM} . Thus, differential pressure ΔP measurements for forward and reverse stimulation can be used as a relative measure of U_{BM} .

During reverse stimulation, the flow across the partition U_{BM} faces the middle-ear impedance, Z_{ME} (Fig. 9b). The magnitude of Z_{ME} (measured during reverse stimulation) at frequencies less than 1 kHz is much larger than the magnitude of the impedance $Z_C = Z_{BM} + Z_{RW}$ (measured during forward stimulation), faced by the flow across the partition U_{BM} , as shown in our measurements plotted in Fig. 7. Thus, $|Z_{ME}|$ may approach the level of $|Z_{Lk_SV}|$, resulting in the flow across the partition U_{BM} dividing between Z_{ME} and Z_{Lk_SV} , according to their relative impedances, resulting in a significant leakage flow (U_{Lk_SV}) at the third-window in SV during reverse stimulation.

Because we measure differential pressure, which is directly related to the flow across the partition, we can focus our analyses of the system in Fig. 9 to that between Z_{ME} and Z_{BM} , where U_{Stap} and U_{BM} flow. This enables us to ignore the uncertainty of the relationship between U_{Act} and U_{BM} (during reverse stimulation) due to differences in actuator area as well as inefficiencies in actuator coupling, U_{Lk_RW} , and/or leakage at ST, U_{Lk_ST} .

As described above, during reverse stimulation the flow U_{BM} can significantly divide between the path through the stapes (represented by Z_{ME}) and a third window leakage path (represented by Z_{Lk_SV}). Thus unlike forward stimulation, during reverse stimulation the flow U_{Lk_SV} is significant, and U_{Stap} does not have the same flow as that across the partition U_{BM} .

We show evidence of this dissimilarity between forward and reverse stimulation by looking at the relationship between U_{Stap} relative to U_{BM} during forward stimulation and reverse stimulation. In each case, we utilize differential pressure ΔP across the partition as a relative measure of differential flow U_{BM} (with the same proportionality constant Z_{BM} in each case, where $U_{BM} = \Delta P / Z_{BM}$). We also utilize V_{Stap} as a relative measure of U_{Stap} for forward and reverse stimulation (because $U_{Stap} = V_{Stap} * A$, where A is the stapes footplate area). This comparison is shown in Fig. 8, and $V_{Stap} / \Delta P$ is generally found to be larger during forward stimulation than reverse stimulation, likely due to the significant SV leakage flow, U_{Lk_SV} , during reverse stimulation. This inequality is most significant at frequencies less than 1 kHz and varies across ears, suggesting that the flow U_{Lk_SV} through a SV third window varies across ears. Therefore, measurements of V_{Stap} cannot be used as a good measure of the input to the cochlea during reverse stimulation because it is not related to the differential pressure across the partition, especially across ears.

Stated another way, while we may not know the leak at the RW during reverse stimulation, we can exploit knowledge of our measurement of ΔP . Because we assume that Z_{BM} remains the same during forward and reverse stimulation, the relationship, $U_{BM}/\Delta P = 1/Z_{BM}$ should be the same for forward and reverse stimulation. During forward stimulation, we know that $U_{BM} \approx U_{Stap}$, we can calculate $U_{Stap}/\Delta P = 1/Z_{BM}$ and $V_{Stap}/\Delta P = 1/(A \cdot Z_{BM})$.

If, during reverse stimulation, there was no significant leakage out of the SV third window, U_{Lk_SV} , then $U_{BM} \approx U_{Stap}$ and we would expect that the relationship of $V_{Stap}/\Delta P = 1/(A \cdot Z_{BM})$ would likewise hold true for reverse stimulation. Because $1/(A \cdot Z_{BM})$ is considered the same for both forward and reverse stimulation, $V_{Stap}/\Delta P$ should be the same for both forward and reverse stimulation. However, as shown in Fig. 8, they are not equal ($U_{BM} \neq U_{Stap}$) because $V_{Stap}/\Delta P$ is lower for reverse as compared to forward stimulation. This confirms that there is a significant flow U_{Lk_SV} for reverse stimulation (as illustrated in Fig. 10). This makes sense, because during reverse stimulation, the reverse flow divides among the large impedances of both the middle ear and the SV third window.

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Highlights

Forward stimulation: cochlear drive depends on scalae pressure magnitude difference

Reverse stimulation: cochlear drive depends on scalae pressure phase difference

Third-window sound flow is significant for reverse stimulation

Stapes velocity is not a good estimate of cochlear drive during reverse stimulation

Figure 1.

Illustration demonstrating the stimuli and recordings made in the inner and middle ear. A loud speaker (LS) introduces sound pressure to the ear canal (P_{EC}) which is measured by a probetube microphone. Intracochlear pressures are measured in both scalae simultaneously (P_{SV} and P_{ST}). Velocities are measured on the cochlear promontory (V_{Prom}), stapes (V_{Stap}), and round window (V_{RW}). Round window stimulation is produced by either an FMT or piezoelectric actuator and their velocities are measured (V_{Act}).

Figure 2.

Vector representations of P_{SV} and P_{ST} during forward and reverse stimulation. In this example, during forward stimulation, P_{SV} is much larger than P_{ST} , resulting in $\Delta P = (P_{SV} - P_{ST})$ that is similar in magnitude to P_{SV} magnitude. During reverse stimulation, P_{SV} and P_{ST} have similar magnitudes, thus the magnitude of $\Delta P = (P_{ST} - P_{SV})$ is greatly affected by the difference in phase between P_{SV} and P_{ST} .

Figure 3.

Comparison of intracochlear pressures elicited by forward and reverse stimulation in a representative experiment. Left column (a & b) plots magnitude and phase of SV and ST pressure with respect to stapes velocity for forward (ear canal) stimulation. Right column (c & d) plots magnitude and phase of SV and ST pressure with respect to actuator velocity for reverse (RW) stimulation. Only data with SNR of at least 6 dB are plotted.

Figure 4.

Intracochlear pressures with respect to stimulus velocities across 8 ears and the means (black lines) and standard deviations (gray shaded areas). For forward stimulation, SV and ST pressure with respect to stapes velocity (P_{SV}/V_{Stap} & P_{ST}/V_{Stap}) magnitudes and phases are plotted (a,b,c,d). For reverse stimulation, SV and ST pressure with respect to actuator velocity (P_{SV}/V_{Act} & P_{ST}/V_{Act}) magnitudes and phases are plotted (e,f,g,h). Solid lines are measurements using a piezoelectric actuator, dashed lines are with the FMT. Only data with SNR of at least 6 dB are plotted.

Figure 5.

The differential pressure magnitude with respect to the stimulus stapes velocity produced by forward stimulation is higher than the differential pressure produced by similar velocity of the RW actuator produced by reverse stimulation in a representative ear. The shaded areas represent the maximum error possible due to recorded change in pressure sensor sensitivities.

Figure 6.

Differential pressure with respect to stimulus velocity for forward and reverse stimulation for 8 ears and their means (black lines) and standard deviations (gray shaded areas). a & b plots the magnitude and phase of the differential pressure ΔP with respect to stapes velocity (V_{Step}) for forward stimulation. c & d plots the magnitude and phase of the differential pressure ΔP with respect to actuator velocity (V_{Act}) for RW reverse stimulation. Only data with SNR of at least 6 dB are plotted.

Figure 7.

(a & b) Magnitude and phase of cochlear impedance ($Z_C = P_{SV}/U_{Stap}$) measured during forward stimulation and (c & d) of middle-ear impedance ($Z_{ME} = P_{SV}/U_{Stap}$) calculated during reverse transduction for 5 ears with their means (black lines) and standard deviation (gray shaded areas). The dashed dark gray line represents similar measurements performed with different techniques by Puria (2003). Only data with SNR of at least 6 dB are plotted.

Figure 8.

Magnitudes of stapes velocity with respect to the differential pressure across the cochlear partition for 5 ears and their means (black lines) and standard deviation (gray shaded areas). The relationship between stapes velocity and differential pressure differ between forward (a) and reverse (b) stimulation, and the differences vary across ears. Only data with SNR of at least 6 dB are plotted.

Figure 9.

Schematic diagram of the system for forward stimulation (a) and reverse stimulation (b). Volume velocities are represented with arrows for the stapes (U_{Stap}) and across the cochlear partition (U_{BM}) for forward and reverse stimulation. Leakage of volume velocities are represented at SV (U_{Lk_SV}) and ST (U_{Lk_ST}). Pressures measured in SV and ST for forward and reverse stimulation are represented by P_{SV} and P_{ST} . Differential impedance across the partition (Z_{BM}), RW impedance (Z_{RW}), middle ear impedance (Z_{ME}), and impedances for leakages at SV (Z_{Lk_SV}) and ST (Z_{Lk_ST}) are illustrated with boxes. During forward stimulation, sound stimulus is represented by the speaker's stimulus pressure (P_{stim}) and source impedance (Z_{source}) and the ear canal pressure as P_{EC} . During reverse stimulation, the volume velocity of the RW due to the actuator is represented by U_{Act} , with accompanying leakage volume velocity (U_{Lk_RW}) and impedance (Z_{Lk_RW}) at the RW.

Figure 10.

(upper illustration) During forward stimulation, $U_{Stap} \approx U_{BM}$ without significant leakage at a SV third window, U_{Lk_SV} . (lower illustration) During reverse stimulation, $U_{Stap} < U_{BM}$, because U_{BM} divides between two high impedances – middle-ear impedance and SV third-window – resulting in a significant leakage at a SV third window, U_{Lk_SV} . Therefore, although U_{Stap} can give a good estimate of the cochlear drive responsible for U_{BM} during forward stimulation, U_{Stap} is not a good estimate of cochlear drive during reverse stimulation.

Table 1

Summary of experiments

Experiment Number	Sensor stability (dB) @ 1kHz: Diff. Initial vs. Final calibration (dB)		Used for analysis	Complication	Stimulator
	SV	ST			
036	<0.5	<0.5	Yes		FMT
041	1	2	Yes		FMT
097	2	1.5	No	Low P _{SV} & P _{ST}	Piezo
101	1	0.5	No	SV sensor shifted half cycle	Piezo
102	1	3	No	Low V _{Snip}	Piezo
103	11	0.5	No	Sensor unstable	Piezo
104	1.5	<0.5	Yes		Piezo
106	0.5	3.5	Yes		Piezo
109	1.5	1	No	Low P _{ST}	Piezo
111	NA	NA ^	No	Air in cochlea	Piezo
112	5	<0.5	No	Sensor unstable	Piezo
113	NA	NA	No	Possible air in cochlea	
116	<0.5	<0.5	Yes		Piezo
119		1.5	Yes		Piezo
129	1	1.5	Yes		Piezo
131	0.5	3	Yes		Piezo