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Performance Considerations of Prosthetic Actuators for Round-Window Stimulation

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Abstract

Round window (RW) stimulation has improved speech perception in patients with mixed hearing loss. In cadaveric temporal bones, we recently showed that RW stimulation with an active prosthesis produced differential pressure across the cochlear partition (a measure related to cochlear transduction) similar to normal forward sound stimulation above 1 kHz, when contact area between the prosthesis and RW is secured. However, there is large variability in the hearing improvement in patients implanted with existing modified prosthesis. This is likely because the middle-ear prosthesis used for RW stimulation was designed for a very different application.

In this paper we utilize recently developed experimental techniques that allow for the calculation of performance specifications for a RW actuator. In cadaveric human temporal bones (N=3), we simultaneously measure scala vestibuli and scala tympani intracochlear pressures, as well as stapes velocity and ear-canal pressure, during normal forward sound stimulation as well as reverse RW stimulation. We then calculate specifications such as the impedance the actuator will need to oppose at the RW, the force with which it must push against the RW, and the velocity and distance by which it must move the RW to obtain cochlear stimulation equivalent to that of specific levels of ear canal pressure under normal sound stimulation. This information is essential for adapting existing prostheses and for designing new actuators specifically for RW stimulation.

Keywords

round window; middle ear; cochlea; scala vestibuli; scala tympani; prosthesis

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INTRODUCTION

Stimulation of the cochlea in a reverse manner, by driving the round-window (RW), has been shown to be an effective mechanism for hearing transduction. In animal studies, evoked responses such as cochlear potentials, auditory-nerve potentials and auditory brainstem potentials were found to be similar between reverse RW stimulation and normal forward oval-window stimulation (Wever and Lawrence, 1950; Dumon et al, 1995; Spindel et al, 1995; Voss et al, 1996). In human cadaveric experiments, RW stimulation has been shown to produce the intracochlear sound pressure difference across the partition necessary for hearing transduction (Nakajima et al, 2009b). Clinically, the RW has been stimulated by an electromechanical prosthetic device, the floating mass transducer (FMT) made by MED-EL, Soundbridge. Various publications (Colletti et al, 2006; Kiefer et al, 2006; Wollenberg et al, 2007; Beltrame et al, 2009; Linder et al, 2009) have reported improved speech perception due to FMT-RW stimulation. However, these clinical tests of RW stimulation have been made across a range of pathologies, and there has been great variation in the hearing results. The FMT device was originally designed and FDA approved as a middle-ear implant that latches onto the incus for sensorineural hearing loss. Thus, it has not been specifically designed for RW stimulation, and there are likely limitations to the design that contribute to the reported variation when adapted to stimulate the RW.

Cadaveric temporal bone preparations allow for controlled experiments to assess FMT-RW stimulation. Nakajima et al (2009b) compared the differential pressure across the partition of the cochlea (which is the stimulus that generates cochlear traveling waves) for FMT-RW stimulation and normal ear-canal sound-evoked ossicular stimulation of the oval window in cadaveric temporal bones. The results show that the FMT stimulation of the RW can result in similar pressure differentials to normal ossicular-oval-window stimulation for frequencies above 1 kHz, but functions poorly at low frequencies, yielding high distortion: This poor low-frequency response is likely due to inherent limitations of the FMT actuator.

We also studied how the FMT device can be best coupled to the RW (Nakajima et al, 2009b). The FMT diameter is similar in dimension to the flexible RW membrane diameter, but the window is usually surrounded by a narrower bony overhang. Even after drilling to reduce the overhang, the FMT appears to overlap the bony perimeter around the round window, which may inhibit good coupling of the device to the RW. Adding a 0.5 mm layer of fascia between the FMT and the RW, both to fill in any space between the FMT and the RW, and buffer any contact between the FMT and the bony perimeter, seems to improve the coupling between the motion of the FMT and the RW. Another consideration is that the “free” end of the FMT (that is not interfaced to the RW) needs to be stabilized, preventing positional shifts between the FMT and RW and reducing relative motions between the FMT and RW. We found that bracing the free end of the FMT to the hypotympanic bony wall with a soft, pliable material helped stabilize the FMT in position, yet allowed for the FMT to move freely. However, even after using fascia to improve coupling to the RW and stabilizing the free end of the FMT, we found there was greater variability in differential-pressure response produced by FMT-RW stimulation compared to the response due to natural sound-stimulation in the forward direction (Nakajima et al, 2009b). Part of the increased variability within the FMT-RW response may arise because the FMT has been designed to meet the needs of a different application.

In this paper, we investigate the design considerations of electromechanical actuators specifically for RW stimulation. An important first step is to quantify the performance requirements necessary for such an actuator. Now that we are able to make simultaneous pressure measurements in scala vestibuli and scala tympani as well as stapes velocity and ear-canal pressure measurements, under normal sound stimulation as well as with RW stimulation, we can calculate the performance specifications necessary to design an actuator that stimulates

the RW in a manner to produce “normal” hearing. We can now calculate (1) the acoustical or mechanical impedance that would be loading the RW actuator, (2) the force required by the actuator at the RW and (3) the actuator velocity and stroke, or distance the actuator needs to traverse, to achieve a pressure differential across the partition corresponding to a desired level of hearing across a specified frequency range.

METHODS

Experimentation on human cadaveric preparation has been approved by the Institutional Review Board of Massachusetts Eye and Ear Infirmary. A detailed description of the temporal bone preparation in which we make simultaneous scala vestibuli and scala tympani intracochlear pressure measurements with micro-scale pressure transducers (developed by Olson, 1998) is provided in Nakajima et al (2009a). The descriptions of how the temporal bone experiments were performed in which we simultaneously measured scala vestibuli and scala tympani intracochlear pressures, stapes velocities and ear-canal pressures during RW stimulation can be found in Nakajima et al (2009b). In this paper, we present data from three temporal bones with complete measurements of pressure and velocity with forward sound stimulation and reverse RW stimulation. Presented data are at least 10 dB above measured noise levels and are generally linear.

The load impedance for the actuator

We first consider the acoustic impedance that loads a RW actuator by considering the load at scala tympani, $Z_{ST(R)}$, under the condition of reverse RW stimulation (R), where

$$Z_{ST(R)} = \frac{P_{ST(R)}}{U_{RW(R)}}. \quad (1)$$

The intracochlear pressure, $P_{ST(R)}$ is measured in scala tympani while the RW is stimulated. For the purpose of estimating RW volume velocity, we assume that the fluid in the cochlea can be approximated as incompressible and that the volume velocities of the stapes at the oval window and round window are equal in magnitude but opposite in sign (as was shown for normal forward ossicular stimulation by Stenfelt et al, 2004):

$$U_{RW} = -U_{Stap}, \quad (2)$$

as defined in Fig. 1, making

$$Z_{ST(R)} = \frac{P_{ST(R)}}{-U_{Stap(R)}}, \quad (3)$$

where the volume velocity at the RW is calculated from the product of the measured stapes velocity with RW stimulation and the area of the stapes footplate, i.e.

$$U_{Stap(R)} = V_{Stap(R)} \cdot A_{Stap}. \quad (4)$$

Thus,

$$Z_{ST(R)} = \frac{P_{ST(R)}}{-\left(V_{Stap(R)} \cdot A_{Stap}\right)}, \quad (5)$$

where the area for the stapes footplate, A_{Stap} , is approximately 3.2 mm^2 (von Békésy, 1960; Aibara, 2001).

We are able to obtain $Z_{ST(R)}$ because we can measure both $P_{ST(R)}$ and $V_{Stap(R)}$ while stimulating the RW (Nakajima et al 2009b). To get the complete impedance, $Z_{Total(R)}$, seen by the actuator at the input of the RW, we should also account for the impedance presented by the RW membrane, $Z_{RW(R)}$:

$$Z_{Total(R)} = Z_{RW(R)} + Z_{ST(R)} \quad (6)$$

To generalize, “actuator” may include the physical transducer and any interposing material between the RW membrane and the transducer (such as the FMT with fascia), or in some cases just the transducer if no interposing material is used. We can estimate $Z_{RW(R)}$ based on measured RW impedance during forward (F) sound stimulation, $Z_{RW(F)}$. We have $Z_{RW(F)}$ data for the three temporal bones presented here as well as several more shown in Nakajima et al. (2009a). As will be seen, the contribution of the RW to the total impedance is not large.

The acoustic impedance $Z_{Total(R)}$ is readily related to mechanical impedance based on the area of the round window (A_{RW}) and the shape of the actuator surface that interfaces to the RW membrane. The round window generally has a slight oval shape with an average size of $2.3 \text{ mm} \times 1.87 \text{ mm}$ (Proctor et al., 1986). Thus for our calculation, we will estimate an area of $A_{RW} = h \cdot w \cdot \pi / 4 = 3.4 \text{ mm}^2$ (based on an ellipse where h and w are the height and width). These calculations can be modified depending on the shape and size of the actuator that would determine the volume displacement of fluid at the RW. Generally, it is best to make a conservative estimate based on the most demanding case to determine design parameters.

Force required by the actuator

The force that the actuator needs to generate for a desired level of perceived sound can now be calculated. The stimulus for sound transduction in the inner ear corresponds to the difference in pressure across the cochlear partition (Dancer and Franke, 1980; Nedzelnitsky, 1980; Nakajima et al, 2009a). In Nakajima et al (2009a) the differential pressure across the partition relative to the ear-canal pressure was found for normal forward sound stimulation. First, we consider the pressure across the partition corresponding to a certain level of sound at the ear canal under normal forward stimulation. Ideally, the RW actuator should be able to produce pressure differentials across the partition similar to those produced by normal forward-conducted ear-canal sound pressure across a wide range of frequencies. From a first set of pressure measurements under forward stimulation, we calculate the transfer function,

$$|H_1(\omega)| = \frac{|P_{SV(F)} - P_{ST(F)}|}{|P_{EC(F)}|}, \quad (7)$$

where $P_{SV(F)}$, $P_{ST(F)}$, $P_{EC(F)}$ are the pressures in scala vestibuli, scala tympani and ear-canal pressure during forward sound stimulation. To achieve a cochlear response equivalent to the forward stimulation of $P_{EC(F)} = X$ across frequency, we need a differential pressure magnitude

$$|P_{Diff}| = \frac{|P_{SV(F)} - P_{ST(F)}|}{|P_{EC(F)}|} \cdot X = |H_1(\omega)| \cdot X. \quad (8)$$

From other pressure measurements with RW stimulation, we calculate the magnitude of the transfer function that defines the trans-cochlear pressure difference associated with a sound pressure in the scala tympani,

$$|H_2(\omega)| = \frac{|P_{ST(R)} - P_{SV(R)}|}{|P_{ST(R)}|}. \quad (9)$$

From $H_1(\omega)$ and $H_2(\omega)$ we calculate the required scala tympani pressure $P_{ST(required)}$ from reverse RW stimulation resulting in equivalent cochlear response due to forward sound stimulation with ear canal pressure X in normal hearing. Taking the ratio:

$$\frac{|H_1(\omega)|}{|H_2(\omega)|} = \frac{|P_{SV(F)} - P_{ST(F)}|}{|P_{EC(F)}|} \cdot \frac{|P_{ST(R)}|}{|P_{ST(R)} - P_{SV(R)}|}, \quad (10)$$

and setting $|P_{SV(F)} - P_{ST(F)}| = |P_{ST(R)} - P_{SV(R)}|$, substituting $P_{ST(required)}$ for $P_{ST(R)}$ and substituting X for $P_{EC(F)}$, we get

$$|P_{ST(required)}| = \frac{|H_1(\omega)|}{|H_2(\omega)|} \cdot X. \quad (11)$$

To calculate the equivalent pressure, $P_{Actuator}$, that must be produced by the actuator at the round window, we need to include the RW impedance, $Z_{RW(R)}$. We can define the pressure ratio between the actuator and the scala tympani as:

$$|H_3(\omega)| = \frac{|P_{Actuator}|}{|P_{ST}|} = \frac{|Z_{RW(R)} + Z_{ST(R)}|}{|Z_{ST(R)}|}. \quad (12)$$

Although $Z_{RW(R)}$ can differ slightly depending on the shape of the actuator's interface to the RW membrane, it can be approximated by the measured $Z_{RW(F)}$ due to forward sound stimulation. Therefore,

$$|H_3(\omega)| = \frac{|Z_{RW(F)} + Z_{ST(R)}|}{|Z_{ST(R)}|}. \quad (13)$$

As will be seen, $Z_{RW(F)}$ is small compared to $Z_{ST(R)}$, so $|H_3(\omega)|$ is well approximated as unity.

The pressure of the actuator acting on the RW during reverse RW stimulation that is equivalent to an ear canal sound pressure of X in forward stimulation is:

$$|P_{Actuator}| = \frac{|H_1(\omega)|}{|H_2(\omega)|} \cdot |H_3(\omega)| \cdot X. \quad (14)$$

To calculate the required force provided by the actuator at the RW (F_{RW}) to achieve a response equivalent to that of ear canal pressure X with normal hearing,

$$F_{RW} = |P_{Actuator}| \cdot A_{RW}, \quad (15)$$

$$= \frac{|H_1(\omega)|}{|H_2(\omega)|} \cdot |H_3(\omega)| \cdot A_{RW} \cdot X, \quad (16)$$

where $A_{RW} = 3.4 \times 10^{-6} \text{ m}^2$. This calculation assumes piston-type motion at the actuator-RW interface with the area of the whole RW driven by the actuator. However an optimum actuator would be designed to have an area that is slightly smaller than the RW membrane area. To calculate the maximum necessary force conservatively, an estimate based on the area of the whole RW membrane is calculated.

Velocity and stroke of the actuator

The required volume velocity at the RW ($U_{RW(required)}$) (and by extension, the required mechanical velocity of the actuator ($V_{RW(required)}$)) to produce differential pressure equivalent to an ear-canal sound level of X under forward stimulation can be directly calculated from the required pressure and $Z_{ST(R)}$:

$$|U_{RW(required)}| = \frac{|H_1(\omega)|}{|H_2(\omega)|} \cdot \frac{X}{|Z_{ST(R)}|} \quad (17)$$

$$|V_{RW(required)}| = \frac{|H_1(\omega)|}{|H_2(\omega)|} \cdot \frac{X}{|Z_{ST(R)}| \cdot A_{RW}} \quad (18)$$

In calculating the velocity we again assume piston-type motion encompassing the entire cross-sectional area of the RW. The stroke, or peak-to-peak displacement, of the actuator (Δx) is the time integral of the instantaneous RW velocity ($v_{RW}(t)$) over a positive half cycle, during RW stimulation. For a sine-wave velocity drive at angular frequency $\omega = 2\pi f = 2\pi/T$, we find a stroke Δx :

$$\Delta x = \int_0^{T/2} v_{RW}(t) dt \quad (19)$$

$$= |V_{RW(required)}| / \omega \cos(\omega t) \Big|_0^{T/2} \quad (20)$$

$$|V_{RW(required)}| / (\pi f) \quad (21)$$

Therefore,

$$\Delta x = \frac{|H_1(\omega)|}{|H_2(\omega)|} \cdot \frac{X}{|Z_{ST(R)}| \cdot A_{RW} \cdot (\pi f)} \quad (22)$$

for a sinusoidal input with frequency f (Hz). Again, the above calculations assume piston-type motion at the actuator-RW interface, with an actuator area that is just smaller than the RW membrane. This calculation can be modified if we assume other various actuator sizes and shapes that might better drive the RW membrane, as will be discussed.

RESULTS

The acoustic impedance at the RW during reverse RW stimulation, $Z_{Total(R)}$, can be obtained because both $P_{ST(R)}$ and $V_{Stap(R)}$ were measured while stimulating the RW (see equation 5 and 6). The RW impedance, estimated by $Z_{RW(F)}$ (discussed in Nakajima et al., 2009a), is very small compared to $Z_{ST(R)}$ (by more than a factor of ten), and does not significantly contribute to the load at the RW. The calculated $Z_{Total(R)}$ versus frequency is plotted in Fig. 2 with units of $G\Omega$, where one mks Ω is equivalent to one $(Pa \cdot s/m^3)$. From the phase data, it can be appreciated that at low frequencies (<500 Hz) the phase is near -90° (dominated by compliance), in the mid frequencies (1–5 kHz) the phase is near 0° (dominated by resistance), and at high frequencies (>7 kHz) the phase is near $+90^\circ$ (dominated by mass). As a comparison, Fig. 3 plots the cochlear impedance at scala vestibuli during forward sound stimulation, $Z_{C(F)}$, which is defined as $Z_{C(F)} = Z_{SV(F)} = P_{SV}/U_{Stap(F)}$. Similar results for $Z_{C(F)}$ for several more temporal bones were presented in Nakajima et al. (2009a), and an average of this previous data is also shown in Fig. 3. Comparison of the impedance at the RW, $Z_{Total(R)}$ obtained by reverse RW stimulation, to the impedance seen by the stapes, $Z_{C(F)}$ obtained by forward sound stimulation, show that the $|Z_{Total(R)}|$ is approximately double that of $|Z_{C(F)}|$ and $\angle Z_{RW(R)}$ changes with frequency, as $\angle Z_{C(F)}$ is near zero degrees for all frequencies (although $\angle Z_{C(F)}$ is influenced by $\angle Z_{RW(F)}$, as was discussed in Nakajima et al. (2009a)). The difference between the two measurements of impedance is explainable by the difference in termination of the cochlea in the two cases. In the forward transmission case the RW with its small impedance terminates the non-driven end of the cochlea. In case of reverse RW stimulation the non-driven end of the cochlea is terminated by the middle-ear output impedance.

Figure 4 plots the transfer function magnitude $|H_1(\omega)| = |P_{SV(F)} - P_{ST(F)}| / |P_{EC(F)}|$ (equation 7) for normal forward sound stimulation in our three test ears as well as the averaged data from previous measurements (Nakajima et al 2009a). Figure 5 plots the transfer function magnitude $|H_2(\omega)| = |P_{ST(R)} - P_{SV(R)}| / |P_{ST(R)}|$ (equation 9) for reverse RW stimulation. The transfer function $|H_3(\omega)| = |Z_{RW(F)} + Z_{ST(R)}| / |Z_{ST(R)}|$ approximates 1 because $Z_{RW(F)}$ is significantly smaller than $Z_{ST(R)}$, as was described earlier regarding equation 13. The force (in units of N or $Pa \cdot m^2$) required of an actuator at the RW to produce a cochlear stimulus equivalent to that produced by 100 dB SPL (2 Pa) is plotted in Fig. 6, conservatively estimated by assuming a piston drive pushing against the entire RW membrane area, A_{RW} (as in equation 16). Thus, to obtain the force required for a desired X dB SPL equivalent hearing, one would shift the curve by $(X - 100)$ dB (shift up for positive and down for negative). It is interesting that below 1 kHz, the required force decreases somewhat with decreasing frequency. This is due to the influence of the differential pressure transfer function $|H_1(\omega)|$, seen in Fig. 4. Likewise, the required force decreases above approximately 2 kHz for the same reason.

The magnitude of the volume velocity required of a RW actuator, $|U_{RW}|$ (expressed in equation 17), is plotted in Fig. 7. The volume velocity requirement decreases with decreasing frequency below 1 kHz because of the combined effects of the transfer functions $|Z_{ST(R)}|$ and $|H_1(\omega)|$ below 1 kHz, as can be appreciated in Fig. 2 (where $|Z_{ST(R)}| \approx |Z_{Total(R)}|$ because $|Z_{RW(R)}| \ll |Z_{ST(R)}|$) and Fig. 4.

The peak-to-peak stroke distance required of an actuator at the RW for cochlear response equivalent to normal forward transduction with $P_{EC} = 100$ dB SPL is plotted in Fig. 8. Again, to obtain the distance required for a desired X dB SPL equivalent hearing, the curve is shifted $(X - 100)$ dB. At frequencies below 1 kHz, the required stroke does not vary greatly because of the counterbalancing influence of $|V_{RW}|$ and f at low frequencies. As can be seen in Fig. 7, U_{RW} (and hence V_{RW}) increases with frequency below 1 kHz, which is countered by the frequency term in the denominator of equation 21. At frequencies above 1 kHz, the required stroke decreases with increased frequency as might be expected.

DISCUSSION

The ability to measure simultaneous intracochlear pressures of P_{SV} and P_{ST} , as well as V_{Stap} , under forward sound stimulation and under reverse RW stimulation, allows us to define performance specifications for the design of a RW actuator that stimulates the cochlea in a reverse manner. In particular, we can define the level of RW stimulation required to produce a cochlear stimulus (differential pressure across the cochlea) that is equivalent to a stimulus produced in a normal ear by a given external-ear sound pressure P_{EC} . For example, if a RW actuator is to produce the equivalent hearing level of 100 dB SPL at 1 kHz at the ear canal (an example maximum output specification for a RW prosthesis): The force required of the actuator can be determined by Fig. 6, yielding a peak force of approximately 200 μ N at 1 kHz. This calculation of the force requirement is conservative in that the force estimate is based on a piston-type motion of the entire RW membrane area. Similarly, the peak volume velocity at 1 kHz would be approximately 0.7 mm^3/s for equivalent hearing level of 100 dB SPL at the ear canal (Fig. 7).

To calculate the velocity and stroke previously in equations 18 and 22, a piston-type motion at the RW displaced by an actuator approximately the size of the RW was considered. However, if the actuator had a very small diameter compared to the RW diameter, in the most extreme case, the shape of the RW deflections would look more like a cone. Likely the shape of the deformation would be between a cone and cylinder. To calculate a conservative (i.e. demanding) estimate of the requirement of the actuator in regards to velocity and stroke, the volume of the driven fluid can be calculated to be similar to that of a cone, 1/3 the volume of a cylinder as calculated above. Therefore equations (18) and (22) could be modified to conservatively estimate the velocity and stroke distance requirements of the actuator:

$$|V_{RW(required)}| = \frac{|H_1(\omega)|}{|H_2(\omega)|} \cdot \frac{X}{|Z_{ST(R)}| \cdot A_{RW}} \cdot 3 \quad (18a)$$

$$\Delta x = \frac{|H_1(\omega)|}{|H_2(\omega)|} \cdot \frac{X}{|Z_{ST(R)}| \cdot A_{RW} \cdot (\pi f)} \cdot 3 \quad (22a)$$

If a RW actuator is to produce the equivalent hearing level of normal sound of 100 dB SPL at the ear canal, the 1 kHz peak stroke distance would be approximately 200 nm assuming a cone deformation at the RW (three times the values plotted in Fig. 8).

The performance specifications defined in this paper will be useful both in better adapting existing actuator technology for RW stimulation, and for developing new actuators designed specifically for RW stimulation.

Simultaneous intracochlear measurements, P_{SV} and P_{ST} , for forward sound stimulation and reverse RW stimulation in cadaveric temporal bones have also allowed for the calculation of forward and reverse middle-ear gain as well as various transfer functions. Comparison of middle-ear gain and Z_C obtained from P_{SV} and V_{Stap} measurements by Puria (2003) showed similar results. Our measurements also allow for the calculation of differential-pressure impedances in the forward (sound) and reverse (RW) stimulation directions (Nakajima et al 2009a discussed the forward stimulation case). These types of measurements are useful in understanding the acoustics and generation of emissions. Our measurements of stapes and RW volume velocity were also checked against the published results of Stenfelt et al. (2004), confirming similar results.

The measurement of differential pressure is a very powerful tool because it is a measure of the stimulus to the cochlear partition and allows an understanding of different cochlear stimulations as well as the effects of various cochlear manipulations. This was difficult to understand with prior measurement techniques such as V_{Stap} , P_{SV} , V_{RW} and bone conduction response alone, because different cochlear stimulations and manipulations change the relationships between these single variables and predicted cochlear response. Thus not only will this technique allow us to calculate useful parameters for design and analysis, but will also allow us to quantify the performance of various prostheses that will mechanically stimulate the cochlea under various diseased conditions.

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Abbreviations

RW	round window
FMT	floating mass transducer
R	reverse round-window stimulation
F	forward sound stimulation
Z	impedance
ST	scala tympani
SV	scala vestibuli
U	volume velocity
$Stap$	stapes
v, V	velocity
A	area
Δx	stroke, displacement

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Figure 1.

Illustration of a stretched-out cochlea showing volume velocity vector of the stapes U_{Stap} at the oval window (OW), volume velocity vector U_{RW} of the round window (RW), with respect to the scala vestibuli and scala tympani pressures, P_{SV} and P_{ST} .

Figure 2.

Plot of the acoustic impedance, $Z_{Total(R)} = Z_{RW(R)} + Z_{ST(R)}$ for reverse RW stimulation for three ears. Units are in $G\Omega$ where 1 mks Ω equals 1 ($\text{Pa}\cdot\text{s}/\text{m}^3$).

Figure 3.

Plot of the cochlear input acoustic impedance, $Z_{C(F)} = Z_{SV(F)} = P_{SV(F)} / U_{Stap(F)}$ for forward sound stimulation in three ears. Units are in $G\Omega$ ($\text{Pa s/m}^3 \cdot 10^9$). The black dashed line is the mean ($N=6$) from a previous study (Nakajima et al 2009a).

Figure 4.

Transfer function magnitude $|H_I(\omega)| = |P_{SV(F)} - P_{ST(F)}| / |P_{EC(F)}|$, for normal forward sound stimulation in three ears. The black dashed line is the mean (N=6) from a previous study (Nakajima et al 2009a).

Figure 5.
 Transfer function magnitude $|H_I(\omega)| = |P_{SV(F)} - P_{ST(F)}| / |P_{EC(F)}|$, for reverse RW stimulation in three ears.

Figure 6. Magnitude of the force required of an actuator, F_{RW} (units dB re 1 micro Newton), for cochlear response equivalent to normal forward sound stimulation with $P_{ec} = 100$ dB SPL in three ears.

Figure 7. Magnitude of volume velocity, $|U_{RW}|$ (units dB re mm³/s) required of a RW actuator for cochlear response equivalent to normal forward sound stimulation with $P_{ec} = 100$ dB SPL in three ears.

Figure 8.
Magnitude of peak-to-peak stroke distance (in units of dB re nm) required of a RW actuator for cochlear response equivalent to normal forward sound stimulation with Pec = 100 dB SPL in three ears.