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3aEA6. In-situ calibrated sound signals and hearing sensitivity

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The acoustic input impedance of the ear provides data on the acoustico-mechanical function of the ear and can be used diagnostically to assess outer and middle ear function. Sound signals delivered to the ear canal can be quantified in-situ if the complex reflectance of the ear has been determined. Hearing sensitivity can be expressed in terms of the forward-going sound pressure wave, or after correcting for acoustic delay, the fraction of the forward-going sound pressure wave transmitted to the middle ear. The latter is an estimate of the signal the cochlea receives. Data is presented examining the virtues of expressing hearing sensitivity in terms of an in-situ calibrated sound signal.

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I. Introduction

The outer and middle ear filter sound so that the sound received by the cochlea is not the sound incident on the eardrum. The role of the outer and middle ear is to impedance match airborne sound to water. A perfect impedance match would deliver all of the sound incident on the eardrum to the cochlea with no reflections (and so there would be no standing waves in the ear canal). Reflection of sound constitutes an expression of the impedance mismatch and so the filtering of sound by the outer and middle ear. In the broader context of electromagnetic waves, Fresnel in 1823 derived the expression for the complex reflectance at an impedance discontinuity. For sound waves measured at the measurement microphone, the complex reflectance (R(f)) is given by

$$R_{ear}(f) = \frac{Z_{ear}(f) - Z_0}{Z_{ear}(f) + Z_0} \qquad (1)$$

where Z_{ear} is the acoustic input impedance of the ear measured at the microphone location and Z_0 is the characteristic impedance of the ear canal at the microphone location,

$$Z_0 = \frac{\rho c}{\pi r^2} \qquad (2)$$

where ρ is the density of air, c is the speed of sound, and r is the radius. Equation 1 is valid for plane waves only. Sound propagation up to about 6 kHz is predominantly in the form of plane waves, the ear canal being well described by a uniform cylinder terminated by the impedance of the middle ear (Stinson, 1985). Beranek (1993) suggests that for the propagation of sound down a tube to travel predominantly as a plane wave the wavelength must be greater than about 6 times the diameter of the tube ($\lambda > 6.d$).

The forward-going sound pressure for a plane wave (P_f) is given by

$$P_f = \frac{P_{mic}}{1 + R_{ear}} \qquad (3)$$

where P_{mic} is the sound pressure measured at the microphone in the ear canal. The fraction of this incident signal transmitted to the middle ear and cochlea (P_t) is given by

$$P_t = (1 - R_{tm}).\, P_f \qquad (4)$$

where

$$R_{tm} = R_{ear}.\, e^{\frac{2wl}{c}} \qquad (5)$$

i.e., the complex reflectance of the ear is corrected for the acoustic delay in the ear canal to provide the complex reflectance at the eardrum.

The transmitted pressure is the fraction of the forward-going sound pressure wave that is transmitted to the middle ear and cochlea at the eardrum and is an estimate of the signal the cochlea receives. To calculate the transmitted sound pressure requires the acoustic length and radius of the ear canal, the latter being required to calculate the characteristic impedance of the ear canal at the position of the microphone. The value used for the ear canal radius has basically, to date, been an educated guess. Rabinowitz (1981) and Voss and Allen (1994) assumed a constant diameter for the ear canal for the calculation of Z_0. Keefe and Simmons (2003) used eartip size to provide a value for the diameter of the ear canal. Keefe and Abdala (2007) derived ear canal radius from the frequency average of the real part of the input impedance however standing waves in the ear canal would seem to impact this estimate (see Figure 1).

Accurate determination of the ear canal acoustic radius and length is necessary to quantify the fraction of the sound signal incident on the eardrum transmitted to the middle ear and cochlea. This paper will present a method for the derivation of ear canal radius and length from the acoustic input impedance of the ear and examine the transmitted pressure as an estimate of the signal the cochlea receives.

II. Method

Twenty two adults, age 18 to 34 years, served as subjects for this study. Subjects had no significant history of noise exposure, no family history of hearing loss, no recent history of ear infections, and no reported hearing problems. Data for this study was drawn from a larger data set obtained as part of an NIH-funded Small Business Innovation Research grant. For inclusion in this study, subjects were required to have a valid measure of the acoustic input impedance of the ear, defined as no substantial difference in the acoustic input impedance of the ear calculated from sound pressure measurements made for stimulus delivery presented from each of two speakers (Miller et al., 2010).

Signal generation and data acquisition was computer controlled using a Mimosa HearID system with version R4 software module with a type II PCMCIA soundcard, coupled to an Etymotic Research 10CP probe assembly, the microphone signal amplified 40 dB and digitized at a rate of 48 kHz. Microphone sensitivity was 50 mV/Pa; sound pressure measurements were corrected in software for the frequency response of the microphone. Fourier analysis was performed with a 2048 point Fast Fourier Transform, data analysis restricted to 256 points and an upper frequency limit of 6 kHz. The Thevenin equivalent acoustic impedance and sound pressure of the probe assembly was determined using four cavities of known acoustic impedance and solving four simultaneous equations with two unknowns, Zs and Ps, the source impedance and sound pressure, visco-thermal effects being accounted for in determining cavity lengths (Allen, 1986; Voss & Allen, 1994). Cavity calibration to obtain Zs and Ps was performed prior to each day of data acquisition. The probe assembly was inserted in the ear canal with the goal of the distal end of the eartip being flush with the entrance to the ear canal, the eartip sized to each ear canal. The eartips used were either ER10C-14A or ER10C-14B, foam eartips that are compressed before insertion into the ear canal and then expand to the ear canal wall after insertion. The ear canal sound pressure frequency response was obtained from sound pressure measurements in the ear canal of one ear of each subject using a sweep frequency or chirp stimulus and the acoustic input impedance of the ear calculated by solving Equation 7.

$$Z_{ear} = Z_s \cdot \frac{P_{mic}}{(P_s - P_{mic})} \tag{7}$$

Hearing thresholds in dB SPL at the measurement microphone for pure tones were obtained using the Hughson-Westlake technique with a 5 dB step-size. Hearing thresholds were measured within the frequency range 250 to 6000 Hz.

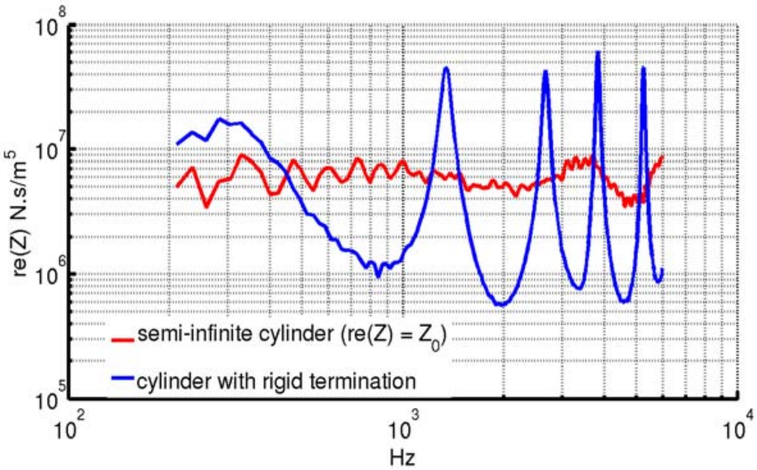

Figure 1 The real part of the acoustic input impedance calculated for a cylinder with a rigid termination (blue curve) and for a cylinder with the same radius but semi-infinite length. The red curve shows the characteristic impedance for both cylinders, the blue curve illustrating the effect of standing waves on the real part of the acoustic input impedance.

III. Results

III. (a) Calculation of Z_0

For a hard-walled cylinder with rigid termination, the characteristic impedance can be deduced relatively easily from the acoustic input impedance of the cylinder. The acoustic input impedance of this cylinder is given by

$$Z = -\frac{\rho c}{\pi * r^2} . \cot\left(\frac{2\pi f l}{c}\right) \tag{8}$$

Given Z, a nonlinear least squares fit to Z provides the radius for the calculation of Z_0. Deriving the radius of the ear canal and so the characteristic impedance is complicated by the middle ear being a non-rigid termination. The acoustic input impedance is dominated by the ear canal but the middle ear does play a role, altering the acoustic length. At low frequencies (less than 1 kHz) the middle ear is stiffness-dominated and so, over this frequency range, the ear canal may be reasonably represented by a cylinder with a rigid termination. To examine the validity of deriving the radius of the ear canal by assuming the ear is a hard-walled cylinder with a rigid termination (restricting the fit to frequencies less than 1 kHz), an acoustic model of the ear was used to generate Z, with $Z_{earmodel}$ given by

$$Z_{EarModel} = Z_0 \frac{\left(Z_{me} + i Z_0 \tan\left(\frac{2\pi f l}{c}\right)\right)}{\left(Z_0 + i Z_{me} \tan\left(\frac{2\pi f l}{c}\right)\right)} \tag{9}$$

where $\frac{1}{Z_{me}} = \sum_{1}^{7} \frac{1}{Z_i}$

and Z_{me} is a bank of simple harmonic oscillators with a total input acoustic impedance similar to that of a young adult female middle ear.

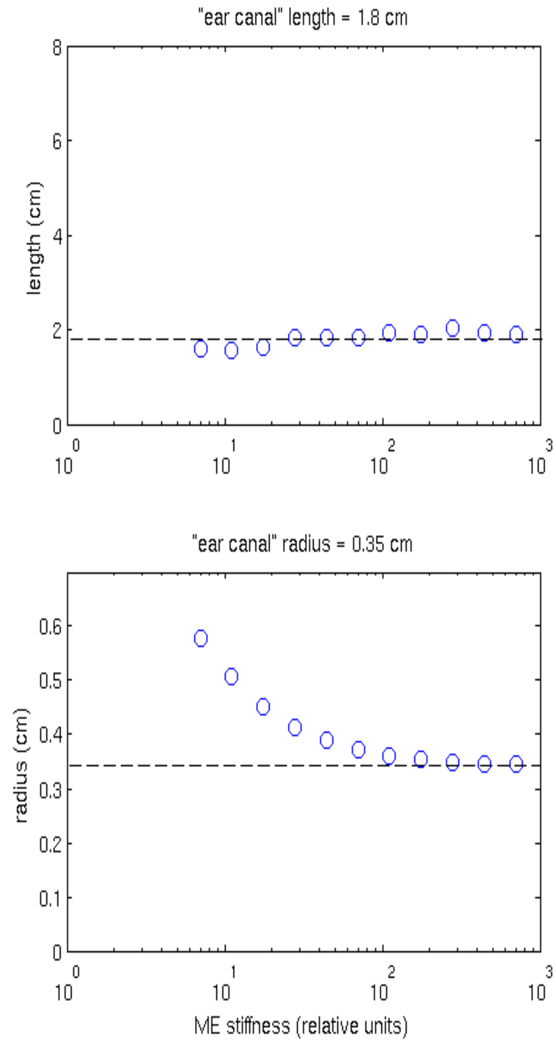

Figure 2 Acoustic length calculated from the SWF (panel A) and the radius calculated from the nonlinear LSF analysis with frequency restricted to less than 1 kHz and $Z_{earmodel}$ given by equation 9.

An impedance phase of zero cycles defines the one-quarter wavelength standing wave frequency (SWF) and so defines the acoustic length of the ear canal. With the length known, equation 8 was fit to $Z_{earmodel}$ (where the model ear canal had a physical length of 1.8 cm) for a range of model middle ear stiffness. Figure 2 shows the acoustic length calculated from the SWF and the radius calculated from the nonlinear LSF. The acoustic length calculated from the one-quarter wavelength SWF varies as stiffness is increased from a nominally normal middle ear to a rigid termination, ranging from about 1.6 to 2.1 cm. With the acoustic length defined by the one-quarter wavelength SWF, a nonlinear LSF to the model ear acoustic input impedance shows the physical radius of 0.35 cm to be reasonably determined with a rigid termination (highest stiffness) but to have an error that increases as the model termination stiffness approaches the nominal normal middle ear value, reaching a LSF estimate for the radius of 0.6 cm, 0.25 cm higher than the actual value. With the acoustic length defined and the radius the only unknown, the nonlinear LSF fails to provide a reasonable estimate of the radius.

Figure 3 shows the result of allowing both the radius and length to vary in fitting equation 8 to $Z_{earmodel}$. In essence, by not constraining the value of the length to be a fixed value, the estimate for the radius improves at the expense of the estimate of the length, but only where the stiffness parameter for the model ear approaches the nominal normal middle ear stiffness. While not shown, fixing the radius and fitting only the length parameter produces a similar result for length to Figure 3. In other words, a

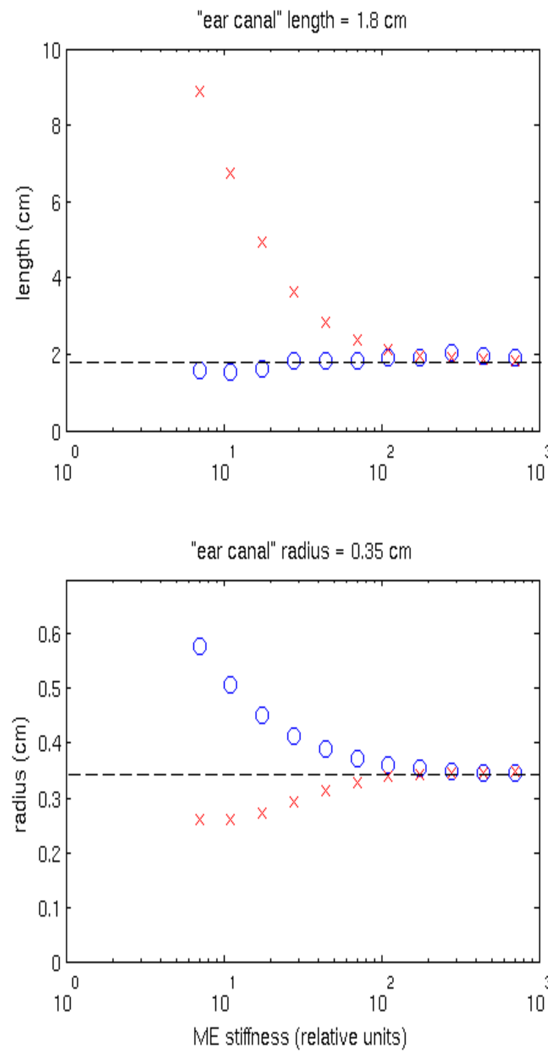

Figure 3 Acoustic length and radius (x) calculated from the nonlinear LSF analysis with frequency restricted to less than 1 kHz and $Z_{earmodel}$ given by equation 9. Also shown is Figure 2 result.

non-rigid termination (the middle ear) to a hard-walled cylinder (the ear canal), when treated as a hard-walled cylinder with a rigid termination, produces an increase in the length of the cylinder.

In Figure 3, for a nominally normal middle ear stiffness, the radius is under-estimated. The smaller the value of the radius, the larger Z_0 and the smaller the impedance mismatch at the eardrum. A value of the radius that is too low will not adequately capture the high-pass filtering of sound by the ear. On the basis that a non-rigid termination to a hard-walled cylinder acts primarily to increase the acoustic length when treated as a hard-walled cylinder with a rigid termination, the preceding analysis was extended to fitting equation 8 to $Z_{earmodel}$ (radius and length) with frequency not restricted to below 1 kHz but to include all frequencies (up to 6 kHz). Figure 4 shows the results presented in Figure 3 with the addition of the radius and length calculated from the nonlinear LSF analysis with frequency not restricted to less than 1 kHz. It is evident from Figure 4 that the estimate of radius and length is improved with the extension of the frequency range of $Z_{earmodel}$ included in the analysis (more data points) for the lowest stiffness values, although the error in length is still quite large. To further investigate extending the upper frequency limit for the LSF analysis from 1 kHz to 6 kHz, the nonlinear LSF estimate of ear canal length was compared with the acoustic ear canal length for the subset of the 22 subjects whose ear canal length was sufficiently long that the SWF was lower than 6 kHz (the upper frequency limit of this study). The acoustic length of the ear canal can be obtained from the one-quarter

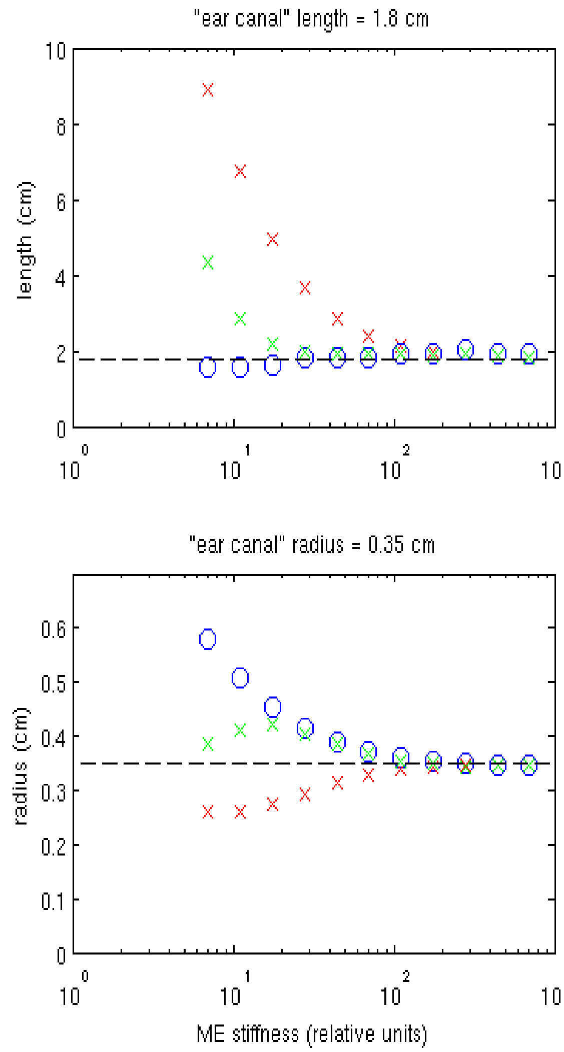

Figure 4 Acoustic length and radius (x) calculated from the nonlinear LSF analysis with frequency extended to 6 kHz and $Z_{earmodel}$ given by equation 9. Also shown is Figure 2 and Figure 3 results.

wavelength SWF and is not a measure of physical length but rather the acoustic length when a non-rigid termination (the middle ear) is attached to a cylinder (the ear canal). Figure 5(a) shows the LSF ear canal length estimate with the analysis limited to below 1 kHz (x) and with the LSF analysis extended to 6 kHz (x) versus the SWF estimate (o). The LSF analysis extended to 6 kHz gives a value of ear canal length that is in good agreement with the SWF value for the 18 subjects with a SWF lower than 6 kHz, within 0.5 cm for all but one subject. In contrast, the analysis with frequency limited to below 1 kHz showed large errors. The line of best fit, constrained to have an intercept of 0, is given by

$$ECL_{LSF} = 0.917\, ECL_{SWF} \qquad (10)$$

Figure 5(b) shows the corresponding radius estimates from the LSF analysis. Eartips used were either ER10C-14A or -14B, with a maximum radius of 0.65 and 0.5 cm respectively. The extended frequency analysis provides more realistic values for ear canal radius i.e., more consistent with eartip size, and more physiologically reasonable. In addition, the good agreement between the LSF prediction of ear canal length from the extended frequency analysis and the SWF value suggests that the other parameter in the fitting, the radius, is well quantified.

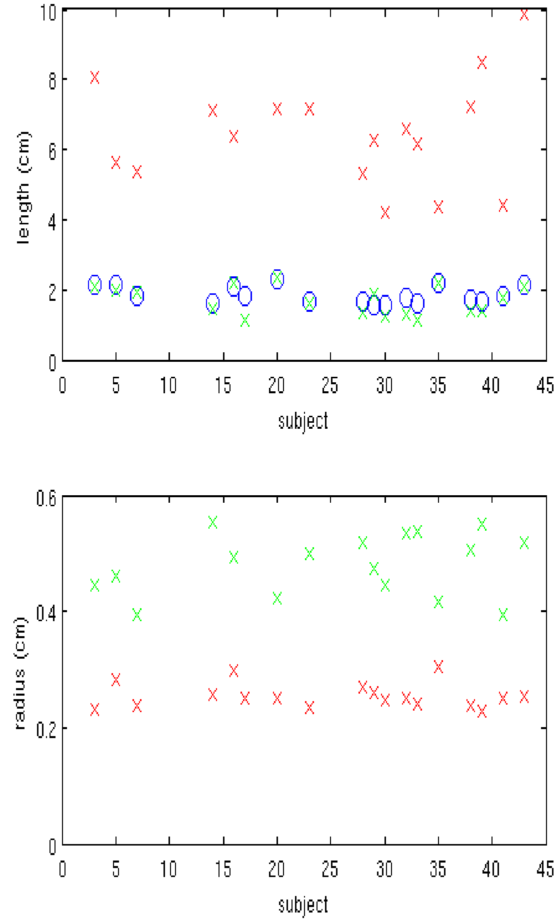

Figure 5 LSF ear canal length and radius estimate for 18 ears with the analysis limited to below 1 kHz (x) and with the LSF analysis extended to 6 kHz (x) versus the SWF estimate (o).

III. (b) Hearing thresholds in terms of $|P_t|$

The reflectance of the ear at the measurement microphone is given by equation 1. The acoustic input impedance of the ear (equation 7) is obtained by measuring sound pressure in the ear canal in response to an acoustically calibrated source, while the characteristic impedance (equation 2) is obtained by deriving the radius as described in section (a) of the Results. With the reflectance, R, determined, one can calculate the sound signal incident on the eardrum, the forward-going sound pressure wave (equation 3). Hearing thresholds can be expressed in terms of this forward-going wave (see Withnell et al., 2009; McCreery et al., 2009), but the outer and middle ear filter sound so that the sound received by the cochlea is not the sound incident on the eardrum. The fraction of the signal incident on the eardrum transmitted to the middle ear is a better estimate of the signal the cochlea receives (equation 4). Figure 6 shows audiograms obtained from 18 ears, hearing thresholds expressed in terms of $|P_f|$ and $|P_t|$ (in dB SPL), the magnitude of the forward-going sound pressure wave incident on the eardrum and the magnitude of the fraction of the forward-going wave transmitted to the middle ear. For frequencies greater than 2 kHz, $20 \log(|P_f|/0.00002) \sim 20 \log(|P_t|/0.00002)$, but below 2 kHz the difference between these two values emphasizes that the sound incident on the eardrum is not the signal the cochlea receives, the impedance mismatch between the ear canal (Z_0) and middle ear high-pass filtering the incident sound energy.

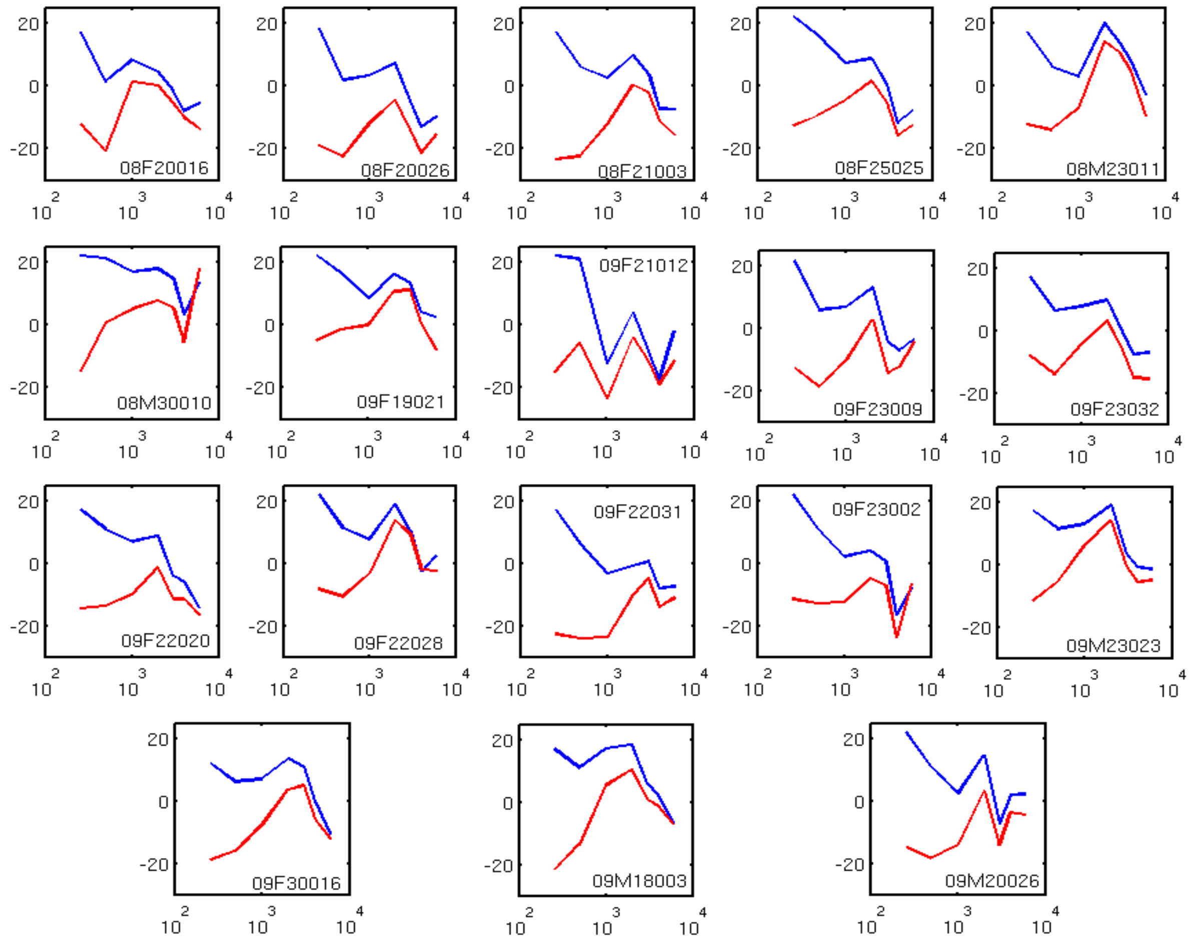

Figure 6 Audiograms for 18 ears, hearing thresholds expressed in terms of $|P_f|$ and $|P_t|$ (in dB SPL).

IV. Discussion

The complex reflectance of the ear characterizes the impedance mismatch between the ear canal (Z_0) and middle ear. This paper provides a method to calculate the physical radius and acoustic length of the ear canal based on the acoustic input impedance determined from sound pressure measurements in the ear canal. With knowledge of the physical radius at the microphone location in the ear canal, the complex reflectance can be derived. The forward-going or incident sound pressure wave can then be derived and, with knowledge of the acoustic length of the ear canal between the microphone and eardrum, the fraction of this incident wave transmitted to the middle ear determined. The latter is an estimate of the signal the cochlea receives, qualified by the fact that the middle ear is not lossless.

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