An *in situ* calibration for hearing thresholds

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Quantifying how the sound delivered to the ear canal relates to hearing threshold has historically relied on acoustic calibration in physical assemblies with an input impedance intended to match the human ear (e.g., a Zwislocki coupler). The variation in the input impedance of the human ear makes such a method of calibration questionable. It is preferable to calibrate the acoustic signal in each ear individually. By using a calibrated sound source and microphone, the acoustic input impedance of the ear can be determined, and the sound delivered to the ear calibrated in terms of either (i) the incident sound pressure wave or (ii) that portion of the incident sound pressure wave transmitted to the middle ear and cochlea. Hearing thresholds expressed in terms of these quantities are reported, these *in situ* calibrations not being confounded by ear canal standing waves. Either would serve as a suitable replacement for the current practice of hearing thresholds expressed in terms of sound pressure level calibrated in a 6cc or 2cc coupler.
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I. INTRODUCTION

Pure tone audiometry has been the standard for measuring hearing sensitivity since soon after the development of the first commercial audiometer (Fowler and Wegel, 1922). Quantification of the signal delivered to the ear (i.e., calibration of the audiometer) entails measuring the sound signal generated by the headphone with a microphone physically coupled to the headphone through a volume intended to match the volume enclosed by the headphone on a human ear (Burkhard and Corliss, 1954). But the acoustic output of a headphone, "as a function of electrical signal, depends on the acoustic load with which it is terminated" (Burkhard and Corliss, 1954, p. 679). The acoustic input impedance of the human ear has been found to vary significantly (Voss and Allen, 1994), raising serious doubt about the validity of the standard method of calibrating an audiometer. This obviously has implications for the accuracy of hearing tests (e.g., Voss *et al.*, 2000). *In situ* measurement of acoustic signals in the ear, with a microphone placed in the ear canal, provides an alternative method of calibration but standing waves confound sound pressure measurements (Stinson *et al.*, 1982; Siegel, 1994). The use of a microphone in clinical settings for recording sound pressure has been predominantly used for recording otoacoustic emissions and monitoring stimulus signal levels, and for obtaining the input impedance of the ear (e.g., tympanometry and wideband power reflectance). Due to the effect of standing waves on the monitoring of signal levels for generating otoacoustic emissions at high frequencies (Siegel, 1994), Neely and Gorga (1998) calibrated their sound stimuli in terms of sound intensity rather than sound pressure. However, as observed by Farmer-Fedor and Rabbitt (2002), the total acoustic intensity also forms standing waves in the ear canal. Farmer-Fedor and Rabbitt (2002) suggested quantifying the incident or forward-going acoustic intensity in the ear canal for the calibration of acoustic signals in the ear canal, it being a "valid measure of the stimulus input to the ear over the entire frequency range" (p. 617), devoid of standing waves, and so "could be used to extend the range of audiometric tests to high frequencies (>8 kHz)" (p. 617). Consistent with this suggestion Hazlewood *et al.* (2007), in consideration of how the ear processes the power it receives, concluded that hearing thresholds must be expressed in terms of the forward-going sound pressure (or intensity) wave, i.e.,

$$P_i^{\mathrm{th}} = \frac{P_m^{\mathrm{th}}}{1+R_m}, \tag{1}$$

where P_i^{th} is the forward-going or incident sound pressure wave at behavioral hearing threshold, P_m^{th} is the sound pressure at the measurement microphone at behavioral hearing threshold, and R_m is the complex reflectance of the ear at the measurement microphone in the ear canal. Pursuant to the clinical application of calibrating stimulus levels in terms of the forward-going sound intensity or pressure wave, Scheperle *et al.* (2008) investigated stimulus level variability in

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distortion product otoacoustic emission measurement with the stimuli calibrated in terms of the forward-going sound wave, and found that stimulus level calibrated in terms of the forward-going sound pressure wave was less variable than total sound pressure as a function of probe insertion depth in the ear canal, consistent with the forward-going or incident sound pressure wave not being contaminated by standing waves.

Sound propagation in the human ear canal at low frequencies occurs predominantly as plane waves. Up to about 6 kHz, the ear canal can be modeled as a uniform cylinder (Stinson, 1985) and so can be represented by a one dimensional transmission line or waveguide, terminated by the impedance of the middle ear. At higher frequencies, multiple higher-order modes may be present due to the shape and varying cross-sectional area of the ear canal. Farmer-Fedor and Rabbitt (2002) explored the contribution of these higher-order modes by examining nonplanar sound waves in the ear canal by making sound pressure measurements at multiple locations in an open ear canal; they concluded that quantifying the nonplanar traveling wave, while more accurate, was technically challenging. In contrast, studies that assume that sound propagates along the ear canal as a plane wave (e.g., Rabinowitz, 1981; Keefe *et al.*, 1993; Voss and Allen, 1994) have provided for the development of clinical instruments that quantify the reflectance of the ear by measuring sound pressure in a closed ear canal with a probe assembly that houses both a sound source (a speaker) and a microphone (e.g., Feeney *et al.*, 2003; Allen *et al.*, 2005).

The use of a single microphone/earphone combination in a closed ear canal to quantify the input impedance/admittance (and therefore reflectance) of the ear for plane waves and subsequent calibration in terms of the forward-going sound intensity or pressure wave rather than total sound pressure requires the prior determination of the acoustical characteristics of the sound source, i.e., the acoustic impedance and pressure of the microphone/earphone combination. The acoustic impedance (Z_s) and sound pressure (P_s) of the sound source can be determined by connecting the sound source to various cavities with known acoustic impedance and measuring the sound pressure in these cavities for a constant electrical source signal (Rabinowitz, 1981; Allen, 1986; Keefe *et al.*, 1992, 1993). With P_s and Z_s determined, the sound source can be inserted in an ear canal and the load admittance or the acoustic input admittance of the ear (Y_m) determined by measuring the sound pressure in the ear canal, i.e.,

$$Y_m = \frac{U_s}{P_m} - Y_s, \tag{2}$$

where Y_m is the input admittance of the ear at the microphone location, U_s is the volume velocity of the sound from the source ($U_s = P_s/Z_s$), Y_s is the admittance of the source, and P_m is the sound pressure at the measurement microphone. A significant, if not the primary, determinant of the setting of hearing sensitivity is the acoustic input impedance/admittance of the ear. The (normalized) acoustic input admittance of the ear (Y_m/Y_0) can also be expressed in terms of the reflectance, i.e.,

$$\frac{Y_m}{Y_0} = \frac{1-R_m}{1+R_m}, \tag{3}$$

where $Y_0 = A/(\rho c)$, A is the cross-sectional area of the ear canal at the location of the microphone, ρ is the density of air in the ear canal, c is the wave velocity of sound in the ear canal, and R_m is the reflectance of the ear at the measurement microphone in the ear canal. This expression derives from the French physicist Fresnel's work (1823) on electromagnetic wave propagation in different media (Born and Wolf, 1999). For sound waves, it assumes a one dimensional waveguide with plane wave propagation. The reflectance of the ear, R_m, includes the acoustic delay in the ear canal.

Sound delivered to the ear should be quantified in terms of the forward-going sound intensity or sound pressure wave,[1] the forward-going sound wave not being contaminated by standing waves in the ear canal (Farmer-Fedor and Rabbitt, 2002). To obtain the forward-going or incident sound pressure wave, P_i, we note that the sound pressure at the microphone (P_m), is given by

$$P_m = P_i + P_r, \tag{4}$$

where P_r is the reflected or backward-going sound pressure wave, and that

$$R_m = \frac{P_r}{P_i}, \tag{5}$$

the reflectance at the measurement microphone being the ratio of the reflected planar sound pressure wave and the incident planar sound pressure wave. From Eqs. (4) and (5) we obtain Eq. (1). For the measurement of hearing thresholds, $|P_m^{\text{th}}|$ is the hearing threshold level in pascals at the measurement microphone, a value that can alter with the location of the microphone in the ear canal, particularly at high frequencies due to ear canal standing waves. The quantity $|1+R_m|$ also varies with position along the ear canal due to standing waves. The result is that $|P_m^{\text{th}}|$ and $|1+R_m|$ covary with position along the ear canal so that $|P_i^{\text{th}}|$ remains constant, i.e., $|P_i^{\text{th}}|$ does not vary as a function of position along the ear canal. Note that the phase of P_i^{th} does depend on the location of the microphone in the ear canal due to the acoustic delay of sound propagation along the ear canal adding to the phase.

The magnitude of the incident (forward-going) sound pressure wave quantifies *in situ* the sound incident on the eardrum. The sound incident on the eardrum is a suitable calibration reference for calculating hearing thresholds but it is not the sound transmitted to the middle ear. The impedance mismatch between the ear canal and the middle ear means that some of the incident sound pressure wave is reflected at the eardrum; we will quantify this reflection of sound by the reflection coefficient *at the eardrum* (not at the microphone), designated R_{tm}. The reflected wave at the eardrum is then given by $R_{tm}P_i$ and that portion of the incident wave transmitted to the middle ear is then given by $(1-R_{tm})P_i$. Therefore, the magnitude of the fraction of the incident wave transmitted to the middle ear at hearing threshold is given by

$$|P_i^{\mathrm{th}}| = |(1 - R_{tm})P_i^{\mathrm{th}}|. \tag{6}$$

Equation (6) makes no assumptions about how sound is transmitted through the middle ear (e.g., the question of transmission losses through the middle ear) being only an expression of that portion of the incident wave in pascals *transmitted to the* middle ear. The quantity R_{tm} is obtained from R_m by correcting for the acoustic delay between the microphone and the eardrum. It would be nice if we could further quantify the reflectance at the stapes footplate but we do not know the acoustic delay through the middle ear.

Using a calibrated sound source and microphone, we can obtain the input admittance (Y_m) and reflectance (R_m) of the ear from the measurement of sound pressure in the ear canal. Behavioral hearing thresholds can then be expressed in terms of the incident planar sound pressure wave, a value uncontaminated by ear canal standing waves. Here we examine hearing thresholds in terms of (i) $|P_i|$, the magnitude of the incident or forward-going sound pressure wave, and (ii) $|P_t|$, the magnitude of the portion of the incident sound pressure wave transmitted to the middle ear. We compare these hearing thresholds with (i) hearing thresholds obtained in terms of the total sound pressure measured in the ear canal and (ii) hearing thresholds obtained in terms of the sound pressure measured in a Zwislocki (DB100) coupler.

II. METHOD

Thirteen females, aged 20–30, with no neuro-otological history, served as subjects for this study. This study was completed with the approval of the human ethics committee, Indiana University, Bloomington.

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 eartip was sized to each ear with eartip size providing the dimension for cross-sectional area, A, for the calculation of the characteristic impedance (for further discussion on ear canal area estimation, see Voss and Allen, 1994, and Keefe and Abdala, 2007). 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, Z_s and P_s, the source impedance and sound pressure, visco-thermal effects being accounted for in determining cavity lengths (Allen, 1986; Voss and Allen, 1994; Keefe, 1984). Cavity calibration to obtain Z_s and P_s 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 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 load admittance (Y_m) calculated by solving Eq. (2). Hearing thresholds in dB sound pressure level (SPL) at the measurement microphone for pure tones were obtained using a method of limits with a 1 dB step-size and six reversals. Hearing thresholds were measured within the frequency range 250–6000 Hz.

Equivalent threshold sound pressure levels (ETSPLs) were calculated from the voltage delivered to the ER10CP probe at each frequency at behavioral threshold multiplied by the sound pressure per volt measured at the behavioral test frequencies in a Zwislocki (DB100) coupler with a condenser microphone.

Data analysis was performed in MATLAB.

III. RESULTS

A. $|1+R_m|$, $|1-R_m|$, and the phase of R_m

Figure 1 shows $|1-R_m|$ and $|1+R_m|$, and the phase of R_m, as a function of frequency for eight subjects. $(1-R_m)$ and $(1+R_m)$ are the numerator and denominator terms of Eq. (3) for the normalized input admittance of the ear (Y_m/Y_0). In a hard-walled cylinder terminated at a right angle by a hard-walled boundary, the magnitude of the input admittance [Eq. (3)] is a maximum when $|1+R(f)|$ is a minimum, when the phase of R is π and the magnitude of R is 1. *A phase of π for the reflectance defines the standing wave frequency*, the standing wave frequency denoted in the phase plots of Figs. 1(a)–1(d) by " SWF." The input admittance of the ear is dominated by the ear canal, the ear canal being reasonably represented by a uniform cylinder up to 6 kHz (Stinson, 1985), and so the standing wave frequency is defined by the angle of π.

In Fig. 1, $|1+R_m|$ has a spectrum consistent with a standing wave at the microphone in the ear canal. In panels (a)–(d), the characteristic notch in the magnitude spectrum is evident, the notch frequency being associated with the standing wave frequency. The frequency corresponding to the reflectance phase of −0.5 cycles, the standing wave frequency, is not exactly the same as the frequency where $|1+R_m|$ is a minimum due to $|R_m|$ varying with frequency (when $|R_m|$ is independent of frequency, the two are the same). In panels (e)–(h), $|1+R_m|$ decreases as a function of frequency, no notch being evident because the standing wave frequency is above the upper frequency limit (6 kHz). The standing wave frequency may be predicted from the straight line fit to the reflectance phase. The term $(1+R_m)$ is also the denominator term in Eq. (1), the magnitude of which should have the same frequency-dependence as the magnitude of the sound pressure measured at the microphone due to standing waves in the ear canal. The numerator term for admittance magnitude, $|1-R_m|$, has a general tendency to have an opposite frequency-dependence to $|1+R_m|$. For a hard-walled cavity, $|1-R_m|$ will have a maximum value when the phase of R is π and the magnitude of R is 1, the same condition for the minimum in $|1+R_m|$. In the ear, as for the minimum of $|1+R_m|$, the frequency where $|1-R_m|$ is a maximum may not coincide with the standing wave frequency due to $|R_m|$ varying with frequency.

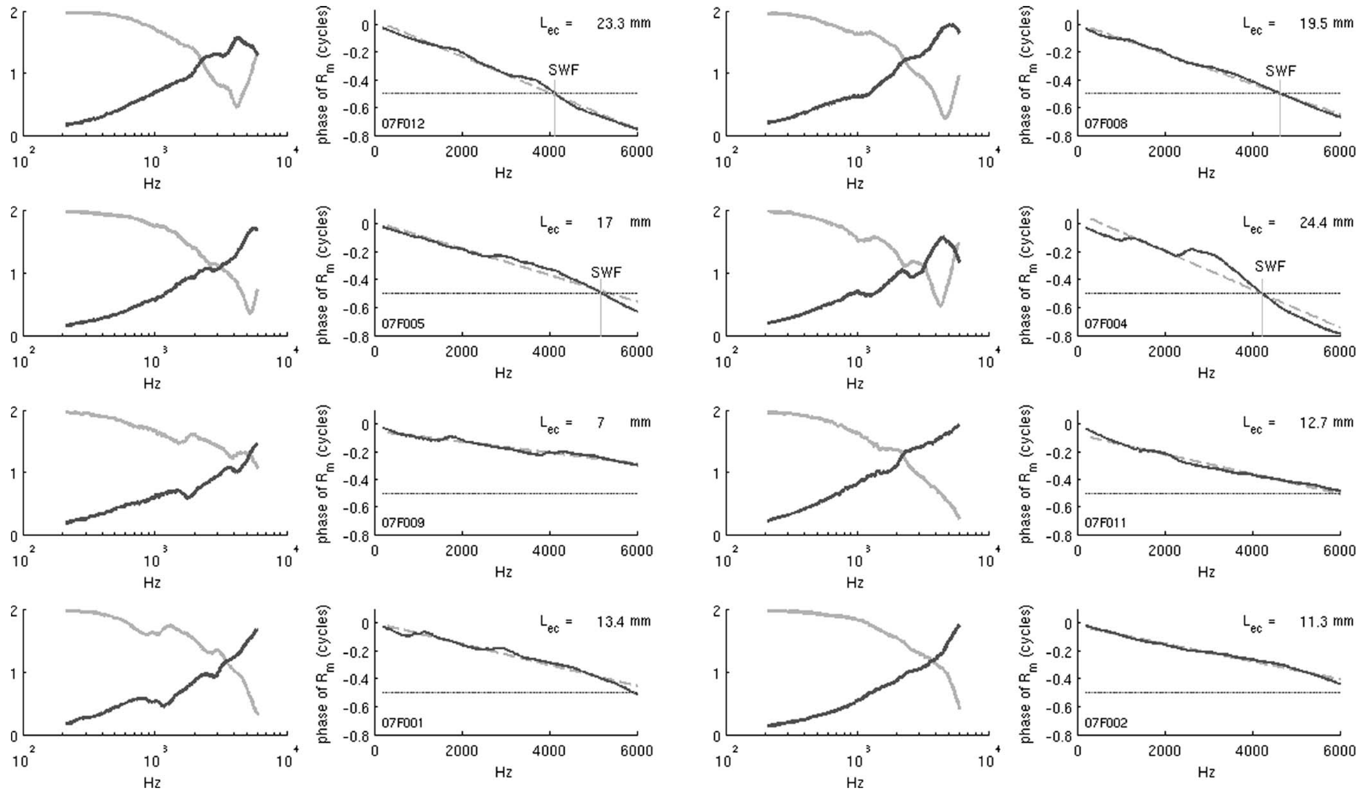

FIG. 1. $|1-R_m|$ and $|1+R_m|$, and the phase of R_m, as a function of frequency for eight subjects. $|1+R_m|$ has a spectrum consistent with a standing wave at the microphone in the ear canal; in four cases, a notch in the magnitude spectrum is evident. The reflectance phase corresponding to the magnitude data identifies the standing wave frequency by the intersection of a straight line at a phase of −0.5 cycles and the reflectance phase. In four cases, no standing wave frequency is identified, the standing wave frequency being above 6 kHz. The phase of the reflectance in each case has been fitted with a straight line. The standing wave frequency may be predicted from this straight line fit to the reflectance phase.

The phase of the reflectance is presumably dominated by ear canal acoustic delay. In Fig. 1, the phase of the reflectance in each case has been fitted with a line of best fit. The slope of this line of best fit provides an estimate of the ear canal acoustic delay from which the length of the ear canal between the eardrum and the microphone can be derived. It is notable that for seven of the eight subjects (excluding 07F004), the phase of the reflectance is well described by the line of best fit. This argues for sound to predominantly be reflected from a single site of reflection (the eardrum) with a pure delay, consistent with planar sound propagation in the ear canal. The contribution of the tympanic membrane/middle ear to the reflectance phase appears to be small, a not surprising finding, particularly for that part of the frequency range above 1 kHz where the input impedance of the middle ear in humans is predominantly resistive.[2] A poor fit of a straight line to the reflectance phase data requires multiple sites of reflection.

From the slope of the linear regression of reflectance phase versus frequency, the distance from the microphone to the eardrum was calculated for each subject, this estimate (in millimeters) given in Fig. 1 in the phase data panels.

B. Quantifying the sound delivered to the ear

Figures 2 and 3 show hearing threshold levels in terms of (i) the SPL measured at the microphone [SPL $=20\log(|P_m^{\mathrm{th}}|/P_{\mathrm{ref}})$ (dB)], where $P_{\mathrm{ref}}=0.000\,02$ Pa, (ii) the incident sound pressure wave [FPL$=20\log(|P_t^{\mathrm{th}}|/P_{\mathrm{ref}})$ (dB)], (iii) the portion of the incident wave transmitted to the middle ear [TPL$=20\log(|P_t^{\mathrm{th}}|/P_{\mathrm{ref}})$ (dB)], and (iv) ETSPL (dB), as a function of frequency for the eight subjects of Fig. 1. Figure 2 shows the four subjects whose ear canal standing wave frequency is below 6 kHz. The SPL at the measurement microphone is larger than the incident SPL below 3 kHz. This is to be expected at low frequencies where, as $|R_m|\rightarrow 1$ and the phase of $R_m\rightarrow 0$, Eq. (5) reduces to $P_i \simeq P_m/2$ and so the SPL measured at the microphone should tend to 6 dB larger than the incident SPL. As frequency increases, any discrepancy between SPL and FPL will depend on the value of $|R_m|$ and the difference in phase between the incident and reflected waves. At 4 kHz, SPL is less than FPL, this being the frequency region where the ear canal standing wave will have the most effect on the measurement of sound pressure at the probe microphone. The standing wave frequency for the four cases shown in Fig. 2, based on the slope of the phase of the reflectance, is 4.1, 4.6, 5.2, and 4.2 kHz, respectively. Note that the magnitude of the difference between the SPL measured at the probe microphone and the SPL in terms of the incident pressure wave near the standing wave frequency is dependent on the reflectance magnitude near this frequency. Hearing thresholds, in terms of that portion of the incident sound pressure wave transmitted to the middle ear, show no particular frequency-dependence across the four subjects; these thresholds better quantify the signal the cochlea is receiving but are qualified by the accuracy of estimating the length of the ear canal. Hearing thresholds in

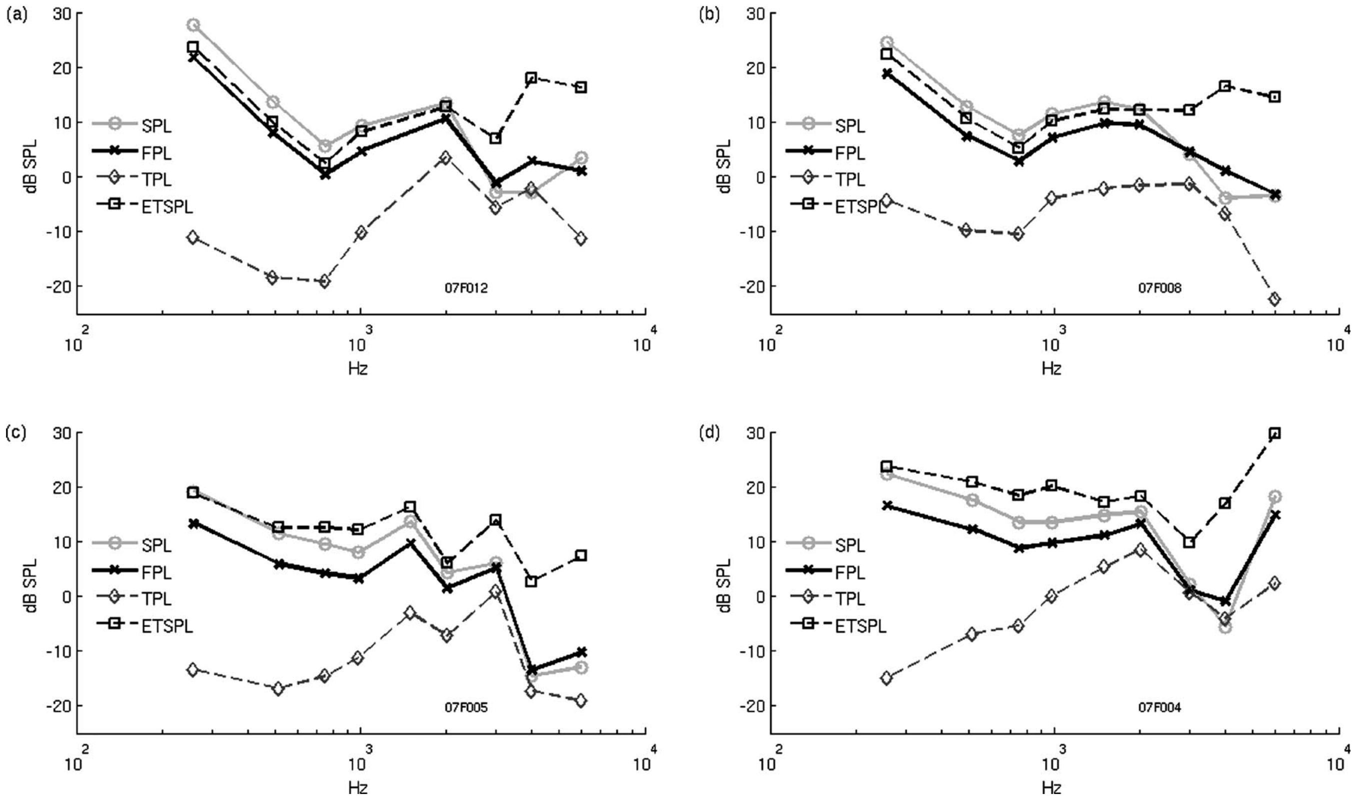

FIG. 2. Hearing threshold levels in terms of (i) the sound pressure measured at the microphone [$SPL=20\log(|P_m^{th}|/P_{ref})$ (dB)], (ii) the incident sound pressure wave [$FPL=20\log(|P_t^{th}|/P_{ref})$ (dB)], (iii) the fraction of the incident wave transmitted to the middle ear [$TPL=20\log(|P_t^{th}|/P_{ref})$ (dB)], and (iv) ETSPL, as a function of frequency for the four subjects whose ear canal standing wave frequency was *below* 6 kHz. $P_{ref}=0.000\,02$ Pa.

terms of ETSPL, based on calibration of the ER10CP ouput in a DB100 coupler measured with a coupler microphone, show reasonable agreement with the SPL configuration up to 2 kHz. At low frequencies, where the ear canal and DB100 effectively reduce acoustically to simple volumes, SPL and ETSPL should be similar in value if the ear canal acoustic volume is similar to the DB100 acoustic volume. Above 2 kHz, ETSPL configuration departs significantly from the configurations for SPL and FPL, although SPL is confounded by standing waves. The discrepancy between ETSPL and FPL demonstrates the coupler-based calibration error in quantifying SPLs at the eardrum.

Figure 3 shows hearing thresholds for the four subjects whose ear canal standing wave frequency at the measurement microphone was above 6 kHz. As for Fig. 2, hearing thresholds for SPL are greater than FPL up to 3 kHz. If the ear canal standing wave frequency is well above 6 kHz, then any observed convergence between the values of SPL and FPL would be mostly attributable to the value of $|R_m| \to 0$. This is the case for panel (a) where SPL remains larger than FPL up to 6 kHz but converges to FPL as frequency increases. Figure 1 for subject 07F009, corresponding to panel (a) of Fig. 3, shows 07F009 to be the ear with the shortest length from eardrum to microphone and so has the highest ear canal standing wave frequency; based on the length estimate of 7 mm, the standing wave frequency is 12.7 kHz. In panels (b)–(d), ears where the eardrum to microphone acoustic length corresponds to standing wave frequencies in the range 6.6–7.9 kHz, FPL exceeds SPL at 6 kHz [and at 4 kHz in panel (b)]. As for Fig. 2, SPL for these subjects is confounded by standing waves at one or more frequencies. Hearing thresholds in terms of that portion of the incident sound pressure wave transmitted to the middle ear (TPL) show no particular frequency-dependence across the four subjects, as observed in Fig. 2. Hearing thresholds in terms of ETSPL, similar to Fig. 2, show divergence in configuration relative to SPL and FPL above 2 kHz, although least so in panel (a) where the standing wave frequency is well above 6 kHz.

It is to be expected that the transmission line properties of the ear canal will preclude either ETSPL or SPL being an accurate estimate of the sound signal at the eardrum. Hearing thresholds expressed in terms of the incident or transmitted sound pressure are not confounded by ear canal standing waves. Either would serve as a suitable replacement for the current practice of hearing thresholds expressed in terms of hearing level or SPL calibrated in a 6cc or 2cc coupler.

IV. DISCUSSION

The acoustic input admittance of the ear can be quantified by using a sound delivery and measurement system with the Thevenin equivalent acoustic parameters, P_s and Z_s, determined using a cavity calibration procedure (e.g., Allen, 1986). The sound signal delivered to the ear can then be quantified in terms of the forward-going sound pressure (or forward-going sound intensity) wave, providing a calibrated signal that is not affected by standing waves in the ear canal (Farmer-Fedor and Rabbitt, 2002). This calibration is valid up to 6 kHz, the plane wave assumption up to 6 kHz being

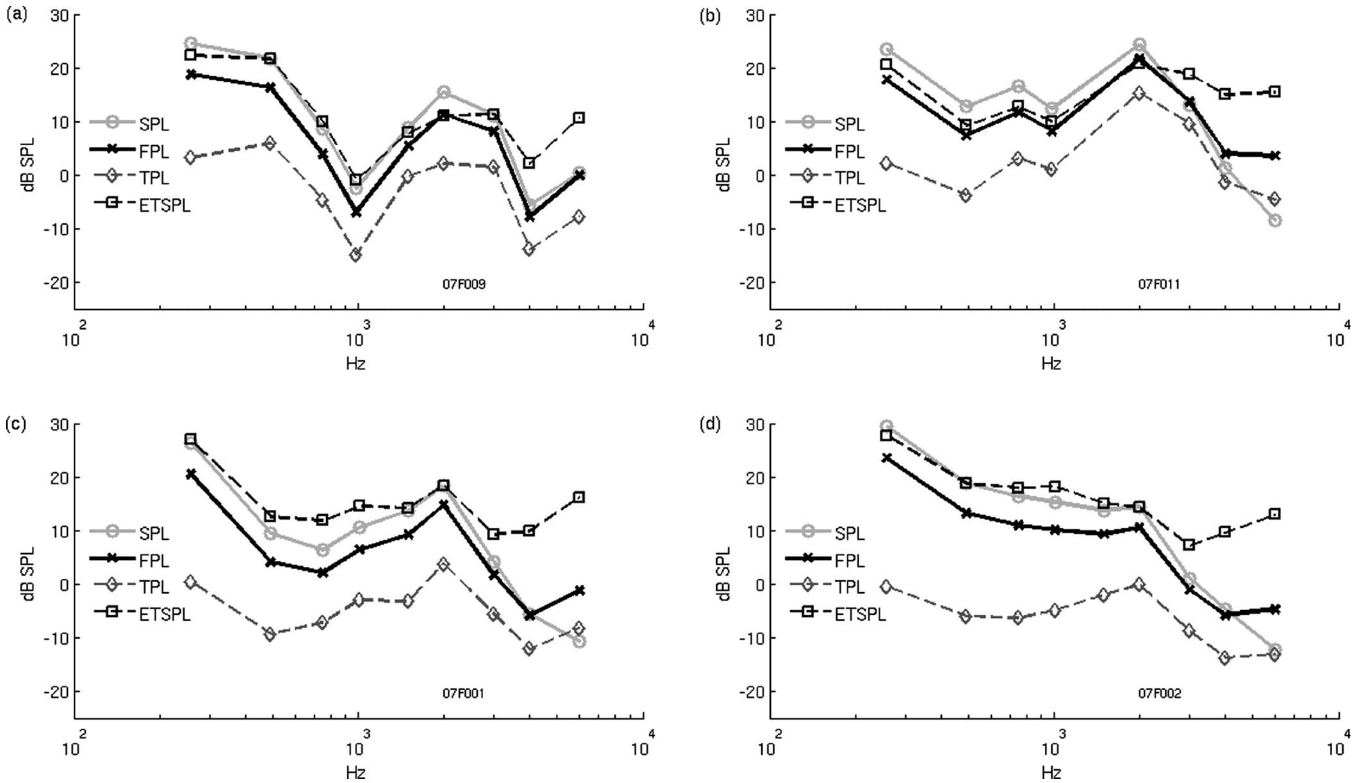

FIG. 3. Hearing threshold levels in terms of (i) the sound pressure measured at the microphone $[\mathrm{SPL}=20\log(|P_m^{\mathrm{th}}|/P_{\mathrm{ref}})\ (\mathrm{dB})]$, (ii) the incident sound pressure wave $[\mathrm{FPL}=20\log(|P_i^{\mathrm{th}}|/P_{\mathrm{ref}})\ (\mathrm{dB})]$, (iii) the fraction of the incident wave transmitted to the middle ear $[\mathrm{TPL}=20\log(|P_t^{\mathrm{th}}|/P_{\mathrm{ref}})\ (\mathrm{dB})]$, and (iv) ETSPL, as a function of frequency for the four subjects whose ear canal standing wave frequency was *above* 6 kHz.

supported by reflectance phase data that are well described by a pure delay and a single site of reflection at the eardrum. Expressing the sound delivered to the ear in terms of $|P_i|$ is preferable to the current convention of not measuring the acoustic signal *in situ* and estimating the sound pressure from measurements in a calibration cavity or artificial ear. Neither the sound pressure measured at the microphone nor sound pressure calibrations in a coupler accurately quantify the sound pressure at the eardrum. The availability of commercial probe assemblies that house both a speaker and a microphone (e.g., ER.10C) and the capacity to quantify the forward-going sound pressure wave argues for pure tone audiometry to be revised so that the acoustic signal is calibrated in each and every ear canal, thereby removing any contamination by standing waves.

The value of R used in this study to calculate hearing thresholds in terms of the incident and transmitted sound pressure was that obtained to a 60 dB (peak) SPL chirp. It is expected that the cochlear input impedance at hearing threshold stimulus levels contributes to R (Allen, 2001), perhaps generating hearing threshold microstructure (Elliot, 1958). At high stimulus levels, cochlear input impedance has been found to be predominantly resistive over much of the frequency range examined in this study (Aibara *et al.*, 2001). Obtaining R to stimulus levels near hearing threshold may furnish a more accurate estimate of the incident sound pressure wave at hearing threshold. However, the increase in signal averaging time to obtain such a value of R (and the use of a pure tone stimulus rather than a chirp to exclude nonlinear stimulus interaction in the cochlea) may render threshold reflectance measurements clinically impractical.

The portion of the incident sound pressure wave transmitted to the middle ear is an attempt to estimate the signal the cochlea receives from the sound pressure incident on the eardrum. The impedance mismatch between the ear canal and the middle ear determines the percentage of the sound incident on the eardrum that is reflected. However, the necessity of calculating the acoustic length from the microphone to the eardrum only applies to calculating hearing thresholds in terms of the transmitted sound pressure. Knowledge of acoustic length is not required for calibration in terms of the forward-going sound pressure wave. The uncertainty in calculating acoustic length argues for expressing hearing thresholds in terms of the incident or forward-going sound pressure as the preferred calibration.

[1]For a plane sound wave, sound intensity and sound pressure are related by the equation $I=P^2/Z_0$ (Lighthill, 1978).

[2]It is likely that the phase oscillates about some mean value over the frequency range that the input impedance is predominantly resistive (O'Connor and Puria, 2008, Parent and Allen, 2007).

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