The clinical utility of expressing hearing thresholds in terms of the forward-going sound pressure wave

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Abstract

Objectives: To assess the clinical utility of quantifying pure tone hearing thresholds in terms of the forward-going sound pressure wave. **Design**: Wideband reflectance and pure tone audiometry was performed on 52 subjects, hearing thresholds quantified in terms of the forward-going sound pressure wave, coupler based calibration, and the sound pressure measured at the microphone. For 20 subjects, the measurements were repeated five times. **Results:** The audiogram configuration differs substantially above 2 kHz for hearing thresholds expressed in terms of the forward-going sound pressure wave versus that obtained from voltage-based values. Repeat testing showed no statistical difference in behavioral thresholds obtained. **Conclusions:** Hearing thresholds expressed in terms of the forward-going sound pressure wave is a more accurate, repeatable means for determining pure tone hearing thresholds, superior to the current coupler-based audiometric technique.

key words: ear, reflectance, hearing thresholds

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Introduction

The antecedents to modern audiometry date back to the latter part of the 19th century (Bunch, 1941), the most significant development prior to 1950 being the incorporation of vacuum tube technology into audiometers to generate frequency-specific stimuli to measure hearing sensitivity (Fletcher & Wegel, 1922). The first commercial audiometer, the Western Electric 1A audiometer, utilized vacuum technology (Fowler & Wegel, 1922).

Western Electric subsequently developed smaller and more affordable commercial audiometers, as did other companies such as Sonotone Corporation (Bunch, 1941)). At issue though was that there was no one calibration standard, exemplified by the comparison of audiometric findings with four different commercial audiometers by Hayden (1938) which demonstrated variable results. This variability in audiometer calibration led to the Council on Physical Therapy of the American Medical Association (AMA) publishing a set of standards that commercial audiometers should comply with in 1939 (A.M.A., 1940). The AMA standard specified that signal output from earphones be calibrated by determining the threshold of hearing of a large group of individuals with the electrical input corresponding to the average hearing threshold constituting the calibrated level. The reference normal hearing threshold was based on the data of the National Health Survey of 1935-1936 using Western Electric type 552 earphones (Beranek, 1993). 'The earphones themselves thus became the standard by which the reference normal threshold was defined' (Beranek, 1993, p. 358). Measurement of the earphone output for the calibrated electrical signal by a microphone coupled to an artificial ear then furnished reference equivalent threshold sound pressure levels (RETSPLs) for the earphone by which other earphones of the same type could then be calibrated using an artificial ear. An artificial ear is a device that couples an earphone to a microphone through a physical volume intended to match the volume enclosed by the earphone on a human ear (Burkhard & Corliss, 1954). Calibration of other types of earphone required a loudness-balancing procedure in which a group of subjects compared the loudness of the calibrated earphone to other earphone types, the electrical voltages that must be applied to the earphone under consideration that correspond to reference normal thresholds constituting the calibration for that earphone (Beranek, 1993).

By the late 1940's, a number of different artificial ears were available (Morrical, Glaser, & Benson, 1949), the American National Standards Institute in 1949 publishing recommended specifications for artificial ear coupler design (ANSI Z24.9-1949). The current ANSI standard (ANSI S3.7-1995(R2003)) for coupler calibration of earphones provides a calibration standard for supra-aural and insert earphones.

Sound pressure levels measured in a coupler provide for a standardization or calibration of earphone output but do not represent sound pressure levels at the human eardrum, particularly above 1.5 to 2 kHz where the geometry of the ear becomes important (Burkhard & Corliss, 1954; Hamershoi, 2006). The ear canal at low frequencies (< 1.5-2 kHz) approximates a simple acoustic volume. At frequencies above 2 kHz, the ear canal acts like a one dimensional transmission line, sound pressure varying with distance from the eardrum due to standing waves (Siegel, 1994; Stinson, Shaw, & Lawton, 1982). 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). At frequencies above 6 kHz, the wavelength of sound is no longer large compared to the transverse dimensions of the ear canal, and so sound propagation is no longer predominantly planar (one dimensional), higher order modes or nonplanar waves becoming significant (Farmer-Fedor & Rabbitt, 2002). This means that sound pressure measurements with a microphone at a single location in the ear canal do not faithfully describe the sound pressure at the eardrum. Evanescent or non-propagating waves form part of the near-field close to the speaker (Brass & Locke, 1997).

In the early 1970s, Zwislocki designed a coupler suitable for calibration of supra-aural, circum-aural and insert earphones (Zwislocki, 1970). It 'was designed to approximate for earphone calibration purposes the dimensional and acoustic characteristics of the average human ear' (Sachs & Burkhard, 1972, p.1). The ear canal was represented by a hard-walled cylinder, terminated by a one half inch condenser microphone to measure sound pressure at the position of the eardrum. The input impedance of the middle ear and cochlea was represented by four branching Helmholtz resonators consisting of a tube extending from the central hard-walled cylinder with a cavity attached to each tube. The Helmholtz resonators depict the middle ear as a bank of parallel simple harmonic oscillators, each tuned to a different frequency, with acoustic damping determining the Q of each oscillator.

Sachs and Burkhard compared sound pressure measured on 11 human ears (converted to eardrum sound pressure), five adult male and six adult female, with that measured in a 2 cc coupler and Zwislocki coupler with insert earphones as the signal source. The response measured in the Zwislocki coupler matched the mean response from the 11 ears within 2 dB up to 7 kHz (Sachs & Burkhard, 1972), with considerable inter-subject variability in the responses from the eleven ears, particularly above 1.5 kHz, illustrating the variability in the acoustic input impedance of the human ear (Voss & Allen, 1994). The response in the 2cc coupler, consistent with it being a simple compliant reactance, followed the contour of the mean response measured in the human ears up to 1 to 1.5 kHz where the human ear canal approximates a simple acoustic reactance, any difference in sound pressure level being due to volume differences. Above 1.5 kHz, the response measured in the 2cc coupler varied widely from that measured in human ears, the transmission line properties of the ear canal not being captured in the 2cc coupler design.

The calibration of supra-aural and insert earphones using the NBS-9A (5.7cc coupler) and 2cc coupler for pure tone audiometry as per the ANSI standard (ANSI S3.7-1995(R2003)) for coupler calibration of earphones ensures that common-type earphones provide the same output in a standardized coupler. But, ear canal acoustics and variation in the acoustic input impedance of the human ear means that calibrated sound pressure levels do not represent the sound pressure at the eardrum. Supra-aural and insert earphones used for audiometry have an acoustic impedance that is similar to the acoustic input impedance of the ear enclosed by the earphones (Voss, Rosowski, Shera, & Peake, 2000) and so sound pressure at the eardrum varies across ears (Burkhard & Corliss, 1954). The clinical impact of the variation in the acoustic input impedance of the human ear on pure tone audiometric thresholds is substantial for ears with normal hearing (Voss & Herman, 2005) and ears with middle ear pathology (Voss, Rosowski, Merchant, et al., 2000).

An alternative to estimating sound pressure levels at the eardrum generated by audiometric earphones using a coupler-based calibration is to measure sound pressures in-situ by placing a microphone in the ear canal. Tympanometry and Otoacoustic Emissions both utilize a microphone housed in a probe assembly with the probe coupled to the ear canal with an eartip to measure sound pressure in the ear canal (placing the microphone some distance from the eardrum), where

$$P_m = P_i + P_r \tag{1}$$

with P_m being the sound pressure at the microphone, P_i the incident or forward-going sound wave and P_r the sound reflected (primarily) from the eardrum (in a normal ear). For Tympanometry sound pressure measurement is restricted to frequencies below 2 kHz where the sound pressure at the measurement microphone is the sum of approximately in-phase incident and reflected waves. For sound waves that sum in-phase at the measurement microphone, the sound pressure measured will be the same as that at the eardrum. For otoacoustic emissions, the frequency range of measurement extends to higher frequencies where the wavelength is not long relative to the length of the ear canal and so the reflected wave undergoes significant phase change relative to the incident wave. A phase difference between the incident and reflected waves results in a sound pressure measurement at the microphone that is not the same as the sound pressure at the eardrum.

For a microphone that is part of a probe assembly coupled to the ear via an eartip (the microphone being some distance from the eardrum), (Siegel, 1994) was the first to point out that significant errors could occur for stimulus levels calibrated in terms of the sound pressure measured at the microphone due to standing waves. In an attempt to address the standing wave issue, Neely and Gorga (1998) examined expressing sound levels in terms of intensity rather than pressure for stimulus calibration for evoking otoacoustic emissions. Farmer-Fedor and Rabbitt (2002) subsequently observed that accurate determination of sound levels at the eardrum based on microphone measurement not at the eardrum requires separating the forward and reverse sound waves that comprise the sound pressure measured at the microphone, either in terms of sound pressure or sound intensity. The incident sound pressure/intensity wave is not affected by standing waves.

The reflectance of the ear expresses the impedance mismatch between the middle ear/cochlea and the characteristic impedance of the ear canal. Unlike tympanometry which is limited to frequencies below 2 kHz by virtue of sound pressure measurements being calibrated against simple acoustic volumes, the reflectance of the ear extends to at least 6 kHz and perhaps higher (Stinson & Daigle, 2005). The human ear processes a wide range of frequencies and so characterizing the function of the outer and middle ear over a wide frequency range is desirable for detecting pathology of the middle ear (Allen, Jeng, & Levitt, 2005; Keefe & Simmons, 2003; Piskorski, Keefe, Simmons, & Gorga, 1999; Shahnaz et al., 2009).

Calibration of sound pressures in the ear canal in terms of the incident sound pressure wave for measurement of behavioral hearing thresholds has been examined by Withnell, Jeng, Waldvogel, Morgenstein, and Allen (2009). Hearing thresholds reported in terms of the sound pressure level of the incident sound pressure wave are not confounded by standing waves and provide a better measure of the sound pressure incident on the eardrum at threshold, in contrast to a coupler-based estimate of signal level.

This study reports pure tone behavioral hearing thresholds obtained from a large cohort of subjects, hearing threshold expressed in terms of the forward-going sound pressure wave, examining the mean and variance in the data and comparing it to the mean and variance of hearing thresholds referenced to the traditional, coupler-based calibration. Expressing hearing thresholds in terms of the forward-going sound pressure wave eliminates the need for coupler calibration, calibration being performed in-situ, with results unaffected by standing waves.

For this method of measurement of hearing threshold to be valid, it must be repeatable. This study also investigates the repeatability of this method by examining behavioral hearing threshold variability over five separate measurement sessions. The clinical utility of expressing behavioral hearing thresholds in terms of the incident sound pressure wave is predicated on the complex reflectance (obtained from measurement of the input impedance of the ear) from repeat measures varying only in terms of phase associated with the position of the microphone in the ear canal.

Further validation of this method is performed by making behavioral threshold measurements with a focus on frequencies around the standing wave frequency, making high density measurements as a function of frequency and comparing behavioral thresholds in sound pressure level with the standing wave obtained from the acoustic input impedance.

Methods

Subjects

Sixty seven adult subjects, of either sex, with no family history of hearing loss and no history of noise exposure comprised the subject group. Fourteen subjects were excluded from the study due to either of i. being older than 34 years of age; ii. a recent history of otitis media; iii. deemed unreliable for behavioral testing; iv. abnormal reflectance curve e.g., acoustic leak.

Three separate experiments were conducted: (1) 52 subjects were seen for wideband power reflectance and pure tone audiometry. (2) 20 subjects were seen for wideband power reflectance and pure tone audiometry, repeated five times (3) 5 subjects were seen for wideband power reflectance and pure tone audiometry with a focus on high density measurements around the standing wave frequency.

This study was completed with the approval of the Indiana University Institutional Review Board.

Data Collection

Wideband Power Reflectance. Ear canal sound pressure was measured on all subjects 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 the ear canal entrance of each ear with the eartip inserted in the ear canal with the goal of the distal end of the eartip being flush with the entrance to the ear canal. An ear canal sound pressure frequency response was obtained from sound pressure measurement in the ear canal to a sweep frequency or chirp stimulus , stimulus level = 60 dB pSPL.

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 (Voss & Allen, 1994). Cavity calibration to obtain Z_s and P_s was performed prior to each day of data acquisition. The input admittance (Y_m) and reflectance (R_m) of the ear at the microphone were calculated from

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

where the s subscript denotes "source", and

$$U_s = \frac{P_s}{Z_s} \tag{3}$$

and

$$R_m = \frac{Y_0 - Y_m}{Y_0 + Y_m} \tag{4}$$

where Y_0 is the characteristic admittance.

Pure tone audiometry. Pure tone stimuli were generated 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. Pure tone audiometry was performed at stimulus frequencies 0.25, 0.5, 1, 2, 3, 4, 6 kHz unless otherwise stated. Hearing threshold was determined using the Hughson-Westlake technique with a 5 dB step-size unless otherwise stated. Stimulus sound pressure (Pm) in the ear canal at the measurement microphone corresponding to behavioral threshold was calculated from the voltage delivered to the earphone and the ear canal sound pressure frequency response to the chirp stimulus, the microphone threshold sound pressure level (MTSPL).

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 one half inch condenser microphone.

To separate the forward-going sound pressure wave from the sound pressure measured at the microphone requires determining the Thevenin equivalent acoustic pressure and acoustic impedance for the source/probe (Allen, 1986; Keefe, Ling, & Bulen, 1992) and then finding the acoustic input impedance of the ear from sound pressure measurement in the ear canal. The forward-going sound pressure wave, P_f is then obtained from

$$P_f = \frac{P_m}{(1 + R)} \tag{5}$$

where P_m is the sound pressure at the microphone and R is the complex reflectance,

$$R = \frac{Z - Z_0}{Z + Z_0} \tag{6}$$

where Z is the acoustic input impedance of the ear and Z_0 is the characteristic impedance. Hearing threshold in dB re the incident sound pressure wave will be designated forward threshold sound pressure level (FTSPL).

Results

Figure 1 shows the mean pure tone hearing thresholds and plus/minus one standard deviation for the 52 subjects, expressed re the incident sound pressure wave, coupler calibration, and in terms of sound pressure level. A number of features stand out in contrasting the three panels of Figure 1 (Withnell et al., 2009, see): (i) The audiograms in the three panels have similar configurations up to 2 kHz. This reflects the ear canal acting as a simple volume up to this frequency and so differences between the ETSPL and MTSPL are due to volume differences between the coupler cavity and the ear. FPSPL will differ from MTSPL by up to 6 dB at low frequencies due to the forward and backward-going sound waves adding in-phase. (ii) Above 2 kHz, ETSPL differs significantly from FTSPL and MTSPL by virtue of the fact that the sound pressure measurement in the coupler is at the 'eardrum' location where the sound adds in-phase at all frequencies, versus sound pressure measurements some distance from the eardrum in the ear. Additionally, the ear canal termination, the middle ear, is not rigid. (iii) The difference between FTSPL and MTSPL above 2 kHz reflects that the former is unaffected by standing waves.

The mean hearing thresholds for ETSPL are similar to RETSPLs quoted for the IEC-711 (ANSI S3.6-2004) within a few dB of each other except at 4 kHz. The DB100 coupler differs from the IEC-711 and so differences are to be expected. The standard deviation was least for the FTSPL, the data not being confounded by standing waves. Sixty eight percent of the hearing thresholds for FTSPL were plus or minus 7 dB of the mean or less.

Figure 2 shows FTSPL, hearing thresholds expressed in terms of the forward-going sound pressure wave, for twenty subjects. Each subject was tested five times. Hearing thresholds were obtained with a 5 dB step-size and so variability over a range of 10 dB would be within test-retest error. The panels in Figure 2, each showing five audiograms on one subject, suggests that, in general, the variability is less than 10 dB, it being true for 84 % of hearing thresholds assessed. A more quantitative evaluation of the repeat measurements of hearing threshold using a one-way ANOVA for each frequency reveals that there is no significant difference between the five measurements of hearing threshold for the 20 subjects at 0.25 kHz, 1 kHz, 2 kHz, and 4 kHz. A significant difference at the 0.05 level for 0.5, 3 and 6 kHz was suggested, necessitating multiple paired sample t-tests with a Bonferroni correction to ascertain which hearing threshold measurements are significantly different. Multiple paired t-tests with a Bonferroni correction showed a significant difference between the third and fourth measurement of hearing thresholds at 0.5 kHz, a significant difference between the first measurement and most of the others at 6 kHz (but not between two through five), and no significant difference at 3 kHz between the five measurements of hearing threshold.

Figure 2 and the statistical analysis of the data in Figure 2 shows the measurement of hearing thresholds in terms of the forward-going sound pressure wave is repeatable and so reliable. This includes the variation in insert placement of the eartip that is to be expected with repeat measurements.

Figure 3 shows $|P_m|$ and $|1 + R|$. These two two terms constitute the numerator and denominator in equation 5, defining the forward-going sound pressure wave. As such, they should be highly correlated. High-density measurements of behavioral threshold around the standing wave frequency is intended to provide validation of the formula of equation 5. Panels A and D show behavioral thresholds to follow the $|1 + R|$ contour reasonably well around the standing wave frequency. Panels B and C suggest a poorer agreement, the behavioral thresholds notch not matching the $|1 + R|$ standing wave notch. Panel E suggests considerable noise associated with the behavioral threshold measurements. It would seem from these results that there is considerable 'noise' associated with the behavioral measurement of hearing threshold, arguing for considerable training of the subjects to be necessary before being able to reliably assess the validity of equation 5 this way. An additional source of variability in behavioral thresholds is threshold microstructure (Elliot, 1958).

Discussion

Clinical pure tone audiometry has well established calibration standards for equipment that guarantees all audiometers generate essentially the same output (see ANSI Standard S3.6 2004). But human ears differ widely in their length, cross-sectional area and shape, producing large variations in ear canal acoustic input impedance (Voss & Allen, 1994, e.g.,). The result is that calibration standards do not provide for constant sound pressure levels at the eardrum for a given voltage delivered to the earphone. Hearing thresholds expressed in dB SPL/HL are a 'best guess'; we do not know the sound pressure level at the eardrum at threshold for each and every person tested.

Sound pressure levels in the ear canal can be quantified by housing a microphone within a probe assembly that also houses the earphone. Sound delivered by the earphone can then be measured by the microphone in the ear canal or in-situ. The microphone, being some distance from the eardrum, measures a sound pressure that is the sum of the forward and backward-going sound pressure waves. Standing waves in the ear canal are a result of the interaction between the forward and backward-going waves. Standing waves at the eardrum add constructively but at a microphone some distance from the eardrum they add destructively. The impact of standing waves on sound pressure measurements in the ear canal can be eliminated by separating the microphone sound pressure into its forward and backward-going sound waves.

This study examined measuring hearing thresholds in terms of the forward-going sound pressure wave. By separating these waves, and examining only the forward-going waves, the effect of standing waves is removed and a valid estimate of the sound pressure at the eardrum is obtained. Figure 1 contrasts hearing thresholds expressed in terms of the forward-going sound pressure wave versus a DB100 coupler-based calibration. Above 2 kHz the difference in hearing thresholds is pronounced. Interestingly, the standard deviation does not differ significantly between the two measures of hearing threshold and the standard deviation is comparable to that reported by Weissler (1968) for hearing thresholds obtained in terms of a coupler-based calibration. This would seem to discount standing waves in the ear canal as a major source of variability at high frequencies.

Farmer-Fedor and Rabbitt (2002) were the first to suggest expressing sound levels in the ear canal in terms of the forward-going sound wave. In 2007, we presented a method for plane waves in the ear canal to pursue this (Hazlewood, Jeng, Withnell, & Allen, 2007) and reported the results of an initial investigation in 2009 (Withnell et al., 2009). Other authors have evaluated in-situ calibration and forward-going sound pressure waves for quantifying stimulus levels for distortion product otoacoustic emissions (Scheperle, Neely, Kopun, & M.P., 2008), probe-microphone hearing aid verification (McCreery, Pittman, Lewis, Neely, & Stelmachowicz, 2009), and pure tone hearing thresholds (Lewis, McCreery, Neely, & Stelmachowicz, 2009). These studies have all shown quantifying the input signal to the ear in terms of the forward-going sound pressure wave to be superior to voltage-based estimates and sound pressure levels measured at the microphone.

As a clinical tool for the measurement of hearing thresholds, FTSPL is repeatable and so reliable. It offers a more accurate means for determining pure tone hearing thresholds, superior to the current coupler-based audiometric technique. Deriving the forward-going sound pressure wave from sound pressure measurements in the ear canal requires the acoustic calibration of the sound source and knowledge of the characteristic impedance of the ear canal. Acoustic calibration of the sound source has been described (Allen, 1986; Keefe et al., 1992) and implemented in commercially available systems, while solutions for the derivation of the acoustic characteristic impedance of the ear canal have recently been proposed (Rasetshwane & Neely, 2011; Rasetshwane, Neely, Allen, & Shera, 2012; Withnell, 2012). Better estimates of the characteristic impedance will refine derivation of the forward-going sound pressure wave (Scheperle, Goodman, & Neely, 2011).

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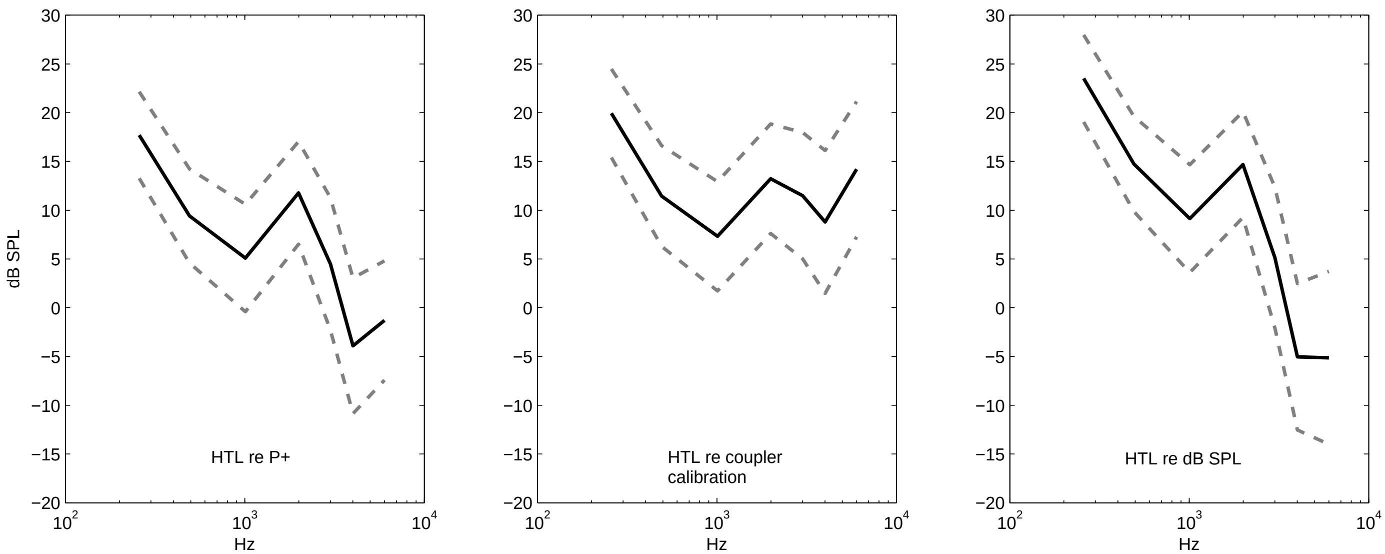

Figure 1. Average pure tone hearing thresholds and plus or minus one standard deviation. n = 52.

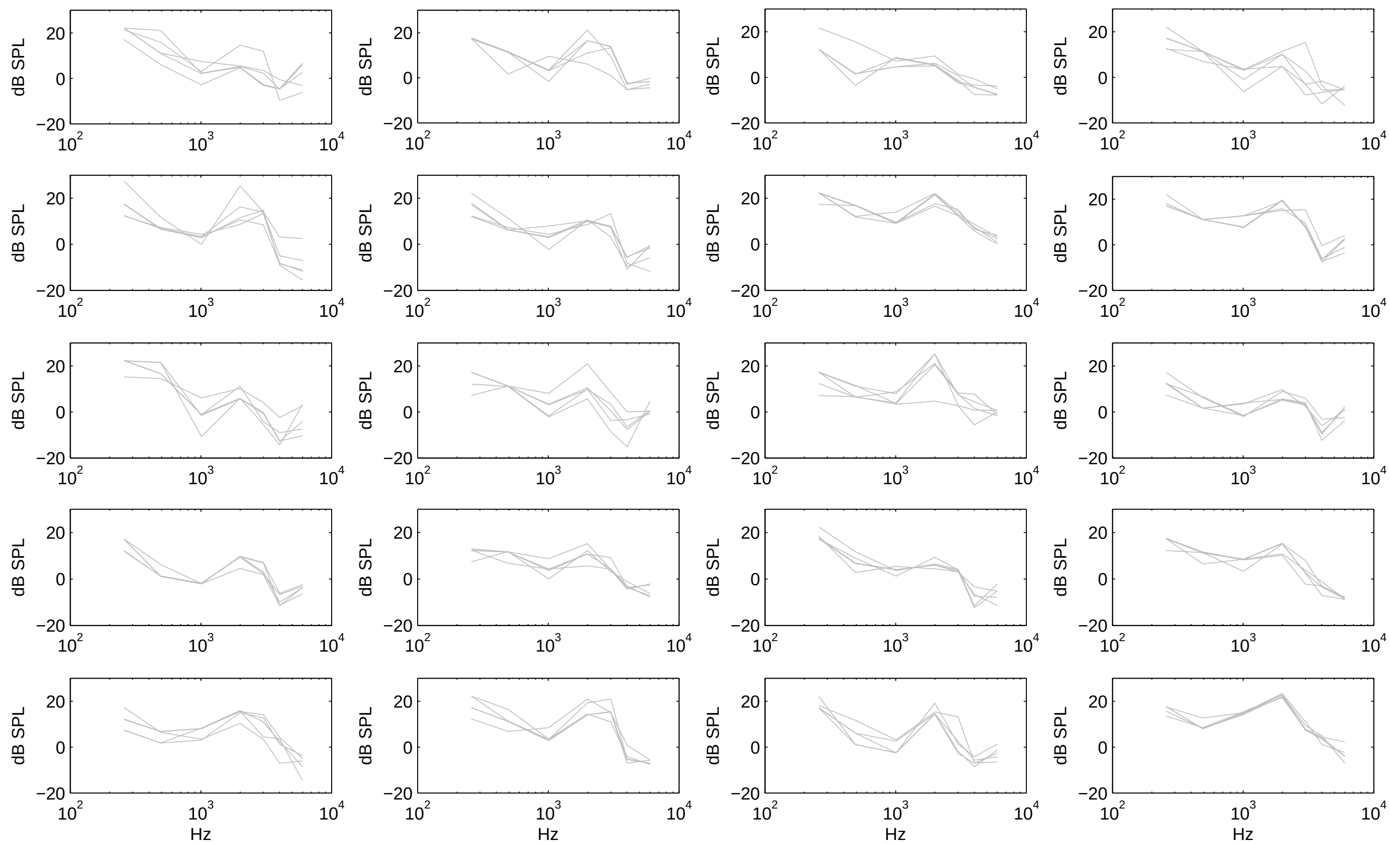

Figure 2. Hearing thresholds expressed in terms of the forward-going sound pressure wave, for twenty subjects. Each subject was tested five times.

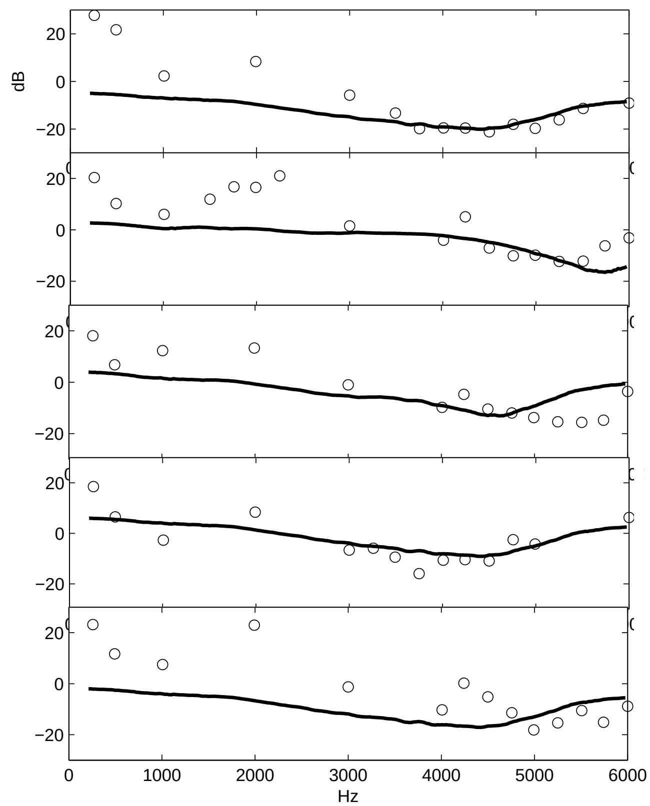

Figure 3. $|P_m|$ and $|1+R|$ on a log scale