Use of forward pressure level to minimize the influence of acoustic standing waves during probe-microphone hearing-aid verification

Ryan W. McCreery,[a] Andrea Pittman,[b] James Lewis,[c]
Stephen T. Neely, and Patricia G. Stelmachowicz
Boys Town National Research Hospital, 555 North 30th Street, Omaha, Nebraska 68131

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Probe-microphone measurements are a reliable method of verifying hearing-aid sound pressure level (SPL) in the ear canal for frequencies between 0.25 and 4 kHz. However, standing waves in the ear canal reduce the accuracy of these measurements above 4 kHz. Recent data suggest that speech information at frequencies up to 10 kHz may enhance speech perception, particularly for children. Incident and reflected components of a stimulus in the ear canal can be separated, allowing the use of forward (incident) pressure as a measure of stimulus level. Two experiments were conducted to determine if hearing-aid output in forward pressure provides valid estimates of *in-situ* sound level in the ear canal. In experiment 1, SPL measurements were obtained at the tympanic membrane and the medial end of an earmold in ten adults. While within-subject test-retest reliability was acceptable, measures near the tympanic membrane reduced the influence of standing waves for two of the ten participants. In experiment 2, forward pressure measurements were found to be unaffected by standing waves in the ear canal for frequencies up to 10 kHz. Implications for clinical assessment of amplification are discussed.
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I. INTRODUCTION

Measures of sound level in the ear canal are an integral part of clinical hearing-aid verification. Modern hearing-aid verification systems determine sound pressure level (SPL) in the ear canal using speech or speech-like stimuli in an effort to characterize the audibility of speech as a function of frequency for individual listeners. These measures account for the acoustic variability of individual ear canals. Probe-microphone measures of the hearing-aid response are preferable to behavioral verification methods because they provide better test-retest reliability (Hellstrom and Axelsson, 1993), a continuous representation of the frequency response instead of data only at discrete frequencies, and the ability to test infants or young children who are not able to participate reliably in behavioral assessments (Zemplenyi *et al.*, 1985).

Despite these advantages, interactions between the incident SPL in the ear canal and the acoustic reflections from the tympanic membrane (TM) limit the accuracy of these measurements for frequencies above 4 kHz. Specifically, partial cancellation of the acoustic signal occurs for frequencies with wavelengths less than four times the distance between the termination of the probe microphone and the TM. Such variations in SPL are therefore dependent on the position of the probe microphone relative to the TM (Gilman and Dirks, 1986; Chan and Geisler, 1990). Gilman and Dirks (1984) demonstrated that, when a probe microphone is placed at a fixed insertion depth, marked frequency-dependent differences in SPL will occur due to individual variations in ear-canal length. The purpose of the current study was to examine two different approaches to quantifying *in-situ* probe-microphone measurements at frequencies above 4 kHz.

Because many hearing aids have limited usable gain above 4 kHz (Boothroyd and Medwetsky, 1992), clinical recommendations regarding probe-tube insertion depth have not previously considered the potential influence of pressure minima in the ear canal at higher frequencies. Several investigators have recommended procedures to minimize the influence of standing waves at frequencies that are within the bandwidth of most hearing aids. Burkhard and Sachs (1977) recommended a probe-tube insertion depth of 5 mm past the termination of the earmold (EM). The authors argued that this depth would limit errors within the frequency range for most hearing aids and avoid acoustic irregularities, such as evanescent modes, stemming from the increase in diameter of the sound channel from the EM sound bore to the ear canal. Caldwell *et al.* (2006) measured consonant spectra and speech-weighted noise at 1, 5, and 10 mm past the termination of the EM and found that the 10 mm position provided the highest overall sound level, as well as the highest output at 6.3 and 8 kHz. Despite mean results showing that the sound level in the high frequencies is greatest at the deepest insertion depth, the impact of standing waves on individual probe-microphone measurements cannot be determined from the group data presented by Caldwell *et al.* (2006).

[a]Author to whom correspondence should be addressed. Electronic mail: mccreeryr@boystown.org
[b]Present address: Department of Speech and Hearing Science, P.O. Box 870102, Tempe, Arizona 85287.
[c]Present address: Department of Communication Sciences and Disorders, The University of Iowa, Iowa City, Iowa 52242.

Current clinical recommendations have been developed to minimize cancellation from standing waves for frequencies below 4 kHz. However, investigators have demonstrated that hearing-aid bandwidth can be extended to higher frequencies. Amplification at frequencies as high as 16 kHz was demonstrated more than 25 years ago (Killion and Tillman, 1982). The current ANSI standard (ANSI, ANSI S3.22-2003) for determination of a hearing aid's frequency range (bandwidth) is determined by calculating the average full-on gain of the device at 1000, 1600, and 2500 Hz and drawing a line 20 dB below this average parallel to the abscissa. The lowest and highest intersecting frequencies represent the bandwidth of the device. This procedure tends to over-estimate the bandwidth available for a hearing-impaired listener because of insufficient gain to amplify the relatively low amplitude of speech energy at frequencies above 4 kHz. Moore *et al.* (2008) conducted a study to determine the gain required to make speech audible at frequencies up to 12.5 kHz for listeners with mild to moderate hearing losses. Despite limitations in hearing-aid gain and reduced speech energy at high frequencies, Moore *et al.* (2008) found that speech could be made audible at 10 kHz in approximately 40% of ears in their cohort. Although most current prescriptive formulas for hearing-aid gain do not provide targets for frequencies above 6 kHz, these studies suggest that making speech audible at frequencies up to 10 kHz is possible for some individuals with mild to moderate hearing loss.

Investigators have evaluated the effects of extended bandwidths on ratings of listener preference and sound quality. Ricketts and colleagues (2008) found a preference for the sound quality of a signal with 9 kHz bandwidth over a 5.5 kHz bandwidth for adults with normal hearing and those with hearing loss. Other investigators have suggested that rating of sound quality for music is greatest with an upper frequency of 16 kHz or greater, and that sound quality ratings for speech can suffer when the low-pass frequency cutoff is less than 10 kHz (Moore and Tan, 2003). Children, in particular, may experience additional improvements in speech recognition with bandwidths that exceed those of currently available devices. Stelmachowicz *et al.* (2001) systematically varied the cut-off frequency of a low-pass filter to assess the perception of /s/ for children and adults with normal hearing and hearing loss. Results suggested that for all listeners, perception of /s/ for female and child talkers reached maximum performance only when the upper bandwidth extended to 9 kHz. Further investigation (Moeller *et al.*, 2007) found that children with hearing loss were delayed in their acquisition of fricative sounds, even when amplification was provided at an early age. The authors concluded that these delays may be related to the limited bandwidth of hearing aids relative to the high-frequency gain necessary to achieve audibility for fricative sounds.

Recent advances in hearing-aid technology, which have resulted in devices with broader bandwidth, have created the need for reliable and valid probe-microphone measures at frequencies above 4 kHz. Theoretically, placement of the probe microphone at or in close proximity to the TM minimizes acoustic standing waves for frequencies up to 10 kHz because the distance between the probe microphone and TM would be less than the wavelengths for frequencies greater than 10 kHz (Dirks and Kincaid, 1987; Gilman and Dirks, 1986). However, the frequencies of standing waves in a closed ear canal are not determined solely by the distance between the probe microphone and the TM. Because the TM is not perpendicular at the termination of the ear canal, the distance between the probe microphone and the TM cannot be accurately represented by a single value. Consequently, the acoustic impedance at the eardrum has also been evaluated to account for the influence of acoustic reflections in the ear canal on estimates of *in-situ* sound level (Stinson *et al.*, 1982). Previous studies have also demonstrated large variations in SPL measurements taken at or near the TM at high frequencies (Dreisbach and Siegel, 2001; Khanna and Stinson, 1985). The sound level measured in the ear canal is a combination of the incident or forward sound presented to the ear as well as an outgoing reflected component. Therefore, accurate estimation of sound pressure in the ear canal should take into account impedance characteristics in order to estimate both forward and reflected pressure components.

The problem of standing waves in the ear canal is not unique to probe-microphone measures for hearing-aid verification. Acoustic standing waves have also been shown to affect *in-situ* calibration for evoking and measuring otoacoustic emissions. Variations in the length of the ear canal, depth of probe insertion, and reflectance characteristics of the TM can lead to significant differences in the level of the stimulus used to generate an emission. These challenges have led researchers to propose several approaches to improve the validity of sound level measurements in the ear canal. Farmer-Fedor and Rabbitt (2002) proposed a measure of *in-situ* sound level in a manner that differentiates the incident acoustic intensity in the ear canal from the outgoing reflected acoustic intensity. Their results suggested that such a method is more reliable than SPL and less likely to be affected by variations in the sound level related to reflectance. Scheperle *et al.* (2008) used a measure of forward pressure level (FPL) that also allows for an estimation of the incident acoustic pressure in the ear canal without the influence of reflected components that are responsible for acoustic standing waves. They compared *in-situ* calibration for distortion product otoacoustic emissions using SPL, sound intensity level (SIL) (Neely and Gorga, 1998), and FPL. Their results suggested that for *in-situ* calibration, SIL and FPL resulted in a more stable calibration and constant measure of sound level across frequency than SPL.

More recently, Withnell *et al.* (2009) found that behavioral thresholds expressed in incident pressure were not affected by standing waves when compared to *in-situ* measures of SPL. Because the acoustic impedance of the ear varies significantly across individuals, the amount of reflected sound energy will also lead to variability in the frequency and extent of standing waves, as well as in the sound transmitted to the middle ear. Estimates such as SIL and FPL, which account for the variability associated with incident and reflected components of the signal, seem well-suited to address the problem of quantifying sound level in the ear canal. These measures have not previously been applied to the measures typically used for hearing-aid verification.

The goal of the present study was to evaluate two different approaches to minimize the influence of acoustic standing waves on real-ear probe-microphone measurements for hearing-aid verification. In experiment 1, measurements at four probe-microphone placements in the ear canal were compared to determine the influence of standing waves at frequencies up to 10 kHz for adults with personal EMs. In experiment 2, behavioral threshold measures referenced to transducer voltage (dB re 1 μV), ear-canal SPL (ecSPL), and FPL in the ear canal were obtained in children to determine if a measure of incident pressure can provide a more accurate representation of input to the middle ear than ecSPL for frequencies above 4 kHz. It was hypothesized that the variation across frequency of FPL would be more similar to decibel referenced to transducer voltage than to ecSPL values, because FPL is not affected by acoustic standing waves in the ear canal. For the purposes of this discussion, ecSPL and FPL will be used to describe measurements taken at the medial end of the probe tube in the ear canal, while SPL and dB μV describe measurements referenced to coupler and voltage values, respectively.

II. METHOD: EXPERIMENT 1

A. Participants

Ten adults (five females and five males) age 25–54 years (mean=38.7 years) participated in experiment 1. All participants had normal middle-ear pressure ($\pm$50 daPa) and TM mobility as evidenced by 226 and 1000 Hz tympanograms (Tympstar, Version 2, GSI). None of the participants had a history of ear surgery. Otoscopic evaluation confirmed that none of the participants had excessive cerumen in the ear canal. Participants were selected for this study because they had a personal EM. Each EM included in the study was constructed of acrylic material. An equal number of right and left ears were included in the study. Three participants had EMs with parallel vents, while the remaining seven participants had unvented EMs. EM vents were occluded with adhesive putty to ensure that the acoustic effects of venting would be minimized.

B. Procedure

For each subject, ear-canal measurements were obtained in one ear using an Etymotic Research ER-7C probe-microphone system. Two Etymotic Research ER 7-14C probe-microphone tubes were glued together to maintain a constant 2 mm separation between the two probe-microphone locations. Measurements were taken at four positions in the ear canal: 2 mm past the sound bore of the EM (EM+2 mm), 4 mm past the sound bore of the EM (EM +4 mm), 2 mm distal to the TM (TM−2 mm), and at the TM (TM). A 5-s segment of broadband noise (70 dB SPL) was delivered to the ear canal via an Etymotic Research 2A earphone, which was coupled to the EM. The stimulus presentation and microphone recording were processed by a personal computer using a Digital Audio Laboratories CardDeluxe sound card. The sampling rate was 32 kHz, and the probe-microphone response was low-pass filtered at 10 kHz. All measurements were completed in a sound-treated booth.

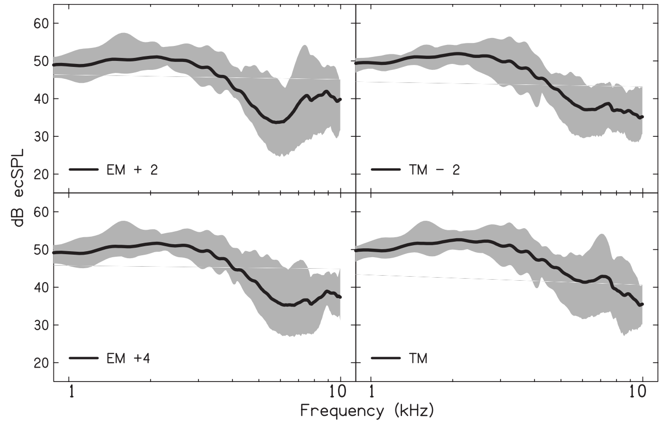

FIG. 1. Each panel displays ecSPL as a function of frequency for each probe-microphone placement. The solid black line is the mean across subjects, and the shaded region represents the range of ecSPL values measured at each frequency.

Results were obtained in a single session approximately 30 min in duration. Subjects were seated in a chair, and the probe-microphone tubes were placed at the eardrum for the first set of measurements. TM placement was verified using a tactile method in which the subject indicated when the probe tube touched the TM. The probe was then withdrawn slightly and secured to the inter-tragal notch using adhesive putty to prevent the probe tube from moving when the EM was placed in the ear. The EM was placed in the ear canal adjacent to the probe tubes for all subjects. In rare cases, acoustic measures revealed that placement of the EM collapsed the probe tube, resulting in a significantly reduced acoustic response. In these instances, the EM was removed and repositioned. Three repetitions of each measurement were made at the TM and at TM−2 without repositioning the probe microphone. The EM was then removed, the probe tubes were placed at EM+2 mm and EM+4 mm positions, and the procedure was repeated. The output files were saved and analyzed using custom software.

III. RESULTS: EXPERIMENT 1

A. Frequency and depth of pressure minima

Figure 1 shows the mean probe-microphone responses across subjects (in spectrum level) as a function of frequency at the four probe placements. The gray shaded area denotes the range of values across subjects. The largest range of responses across subjects occurred at frequencies above 2 kHz. Comparisons across the four probe placements were made for both the frequency of the minimum ecSPL (i.e., notch frequency) and the notch depth, defined as the difference between the maximum ecSPL below the notch and the ecSPL at the notch frequency. These calculations were analyzed to determine if placements at or near the TM resulted in a smaller notch depth or if the notch occurred at higher frequencies than for probe placements near the terminal end of the EM. Means and standard deviations for notch depths and notch frequencies for each placement are reported in Table I.

Analyses of variance (ANOVAs) were used to analyze differences between probe placements with notch frequency and notch depth as within-subjects factors. Notch depth is

TABLE I. Mean and standard deviation notch depth and frequency.

	EM+2	EM+4	TM−2	TM
Notch depth (dB)	23.8 (3.25)	22.04 (3.11)	21.19 (2.56)	19.7 (5.37)
Notch frequency (Hz)	6422 (1431)	7043 (1201)	7832 (1726)	8567 (1470)

plotted as a function of frequency for each probe placement in Fig. 2. Mean notch frequency increased significantly with proximity to the TM [$F_{3,27}=6.06$, $p=0.003$, $\eta^2_p=0.402$]. Post hoc tests using Bonferroni adjustments made for multiple comparisons ($p=0.0125$) revealed a significant difference between the TM and EM+2 conditions only. The depth of the notch did not vary significantly across probe placement [$F_{3,27}=3.01$, $p=0.15$, $\eta^2_p=0.104$]. The smallest individual notch depth at the TM position was greater than 10 dB, suggesting that clinically-significant notches were present in all participants even at the TM placement.

B. Test-retest reliability

Figure 3 displays the mean ecSPL differences across the three trials for each probe-tube placement, as well as a shaded area representing the range of ecSPL differences across trials. Below 4 kHz, the average difference in ecSPL was less than 2 dB for all four positions. Above 4 kHz, the average difference was less than 5 dB for all four positions. Variations greater than 10 dB were observed for each placement, except for the TM position, where the maximum variation across trials was approximately 8 dB. The maximum variation in ecSPL occurred between 4 and 10 kHz, which corresponds with the frequency range where significant pressure minima are most frequently observed.

IV. DISCUSSION: EXPERIMENT 1

Previous clinical recommendations for probe-tube insertion depth are based on minimizing the influence of acoustic pressure minima on *in-situ* probe-microphone measurements by placing the probe microphone in close proximity to the TM. However, the results from experiment 1 suggest that clinically-significant standing waves influence probe-

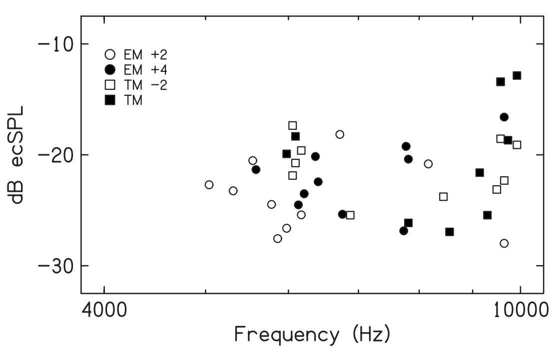

FIG. 2. Notch depth in ecSPL is plotted as a function of frequency for each participant. Probe placements are represented by symbols with circles for EM placements and squares for TM placements. Open symbols correspond with distal probe positions, while solid black symbols correspond with medial positions.

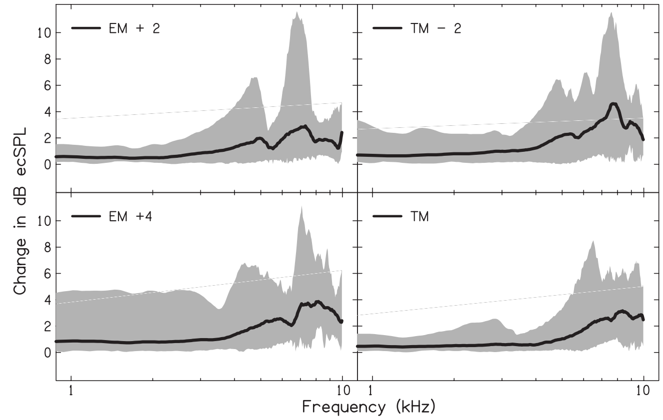

FIG. 3. The solid black line represents the mean differences in ecSPL across three trials as a function of frequency. The shaded area represents the minimum and maximum deviations observed at each frequency across three trials. Each panel represents a different probe placement.

microphone measurements even when the probe is placed in close proximity to the TM. Several participants had pressure minima at frequencies significantly lower than expected based on the distance between the probe microphone and the TM and the frequency corresponding with $\frac{1}{4}$ wavelength. All participants had marked pressure minima below 10 kHz and several participants showed pressure minima at the TM for frequencies within the bandwidth of current hearing aids. A probe-microphone response that decreases at higher frequencies could lead clinicians to make unnecessary adjustments to the response of a hearing aid, which could lead to patient discomfort and/or rejection of amplification.

Significant standing waves measured in close proximity to the TM in all participants raises concerns that the tactile method of probe placement may not have resulted in the desired probe insertion depth or that notches influence measurements near the TM. While some previous studies have revealed that the frequency of the pressure minimum in the ear canal is dependent on the distance between the probe and the TM (Gilman and Dirks, 1986), other studies where *in-situ* measurements of sound level were taken at or near the TM revealed significant pressure minima that would not be predicted based solely on this distance (Dreisbach and Siegel, 2001; Khanna and Stinson, 1985). Given significant individual differences in the impedance and reflectance between individuals at frequencies above 4 kHz (Voss and Allen, 1994; Stinson *et al.*, 1982), the fact that pressure minima cannot be easily predicted based on the measurement distance from the TM is not surprising. Even if probe-microphone placement at the TM provided an estimate of *in-situ* sound level that was not affected by standing waves, the potential for patient discomfort and difficulty in establishing and maintaining this position makes the clinical recommendation impractical.

Data from experiment 1 also suggest that the within-subject test-retest reliability for all probe positions was good. Below 4 kHz, the mean variability was less than 3 dB. Greater variability was present at higher frequencies where changes in probe position resulted in larger deviations across trials. As might be expected given the complex interaction between the incident and reflected acoustic energies for frequencies above 4 kHz, the largest deviations across trials occurred in the frequency region of the notch. The amount of variability at frequencies >4 kHz across trials where attempts were made to maintain the position of the probe at the same location is indicative of the difficulty faced by clinicians in making consistent measurements at higher frequencies. While probe-microphone measurements have acceptable within-subject test-retest reliability, the presence of significant pressure minima for frequencies above 4 kHz suggests that ecSPL may not provide a valid measure of ear-canal sound level for hearing aids with wider bandwidths.

Distance between the probe and TM is one factor that influences the frequency of ecSPL pressure minima in the ear canal, but the oblique orientation of the TM at the end of the ear canal means that the probe distance from the TM is never a single value. Alternatively, the acoustic impedance of the sound source and ear canal may help to more accurately characterize *in-situ* sound levels. Scheperle *et al.* (2008) used a calibration method for otoacoustic emissions that measures the Thevenin-equivalent source impedance and pressure to isolate the incident and reflected components of the ear-canal response. Their results indicated that the ear-canal response expressed in FPL was less likely to contain substantial variations in pressure than measurements expressed in ecSPL. Therefore, experiment 2 was conducted to determine if FPL, which accounts for the acoustic impedance of both the transducer and ear canal, would provide a more accurate estimate of the sound input to the middle ear compared to traditional probe-microphone measures expressed in SPL.

V. METHOD: EXPERIMENT 2

A. Participants

16 normal-hearing children (8 males and 8 females), ages 9–15 years (mean=12.1), participated in the second experiment. This age range was selected to reduce the likelihood that behavioral thresholds were elevated at high frequencies due to noise-induced hearing loss. None of the participants had a history of ear surgery. Otoscopic evaluation was completed to confirm that participants did not have excessive cerumen in the ear canal or tympanostomy tubes.

B. Procedure

Data collection took place in a sound-treated room over two sessions approximately 30–45 min in length. At the first visit, hearing thresholds were measured in the test ear of each subject at 4, 6, 8, 9, and 10 kHz using Sennheiser HD 25-1 earphones. These results were compared to normative data collected from 32 children in a previous study using the same transducer (Stelmachowicz *et al.*, 2007). An automated method of limits with adaptive step-size was used until the standard deviation of the threshold response was less than or equal to 2.5 dB at each frequency. If a participant did not meet the inclusion criteria in one ear, evaluation of the other ear was attempted. Individuals with variations in behavioral thresholds were excluded to limit the presence of notches that were the result of threshold variability instead of standing waves. Four participants were excluded on the basis of

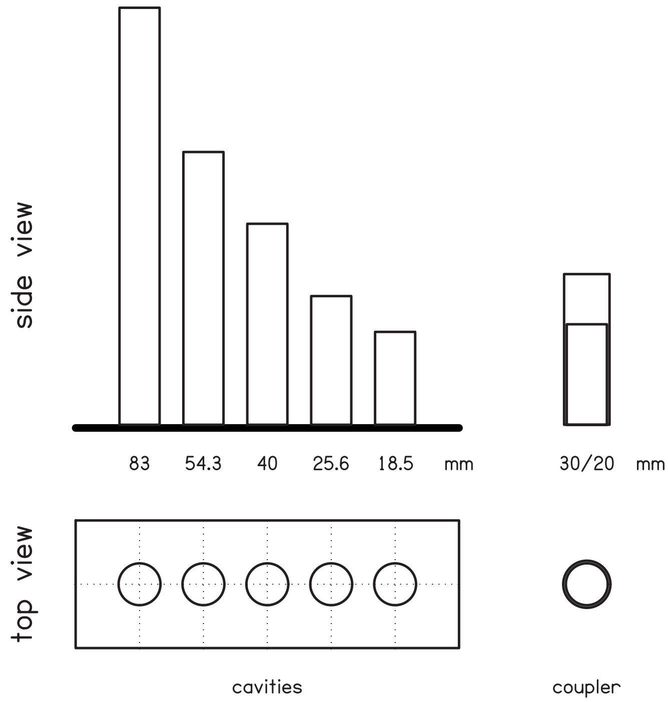

FIG. 4. Schematic of brass cavities and coupler used to obtain source calibrations in experiment 2.

hearing outside the normative range or variations in threshold >15 dB between adjacent frequencies, reducing the number of participants to 12. Following high-frequency audiometric testing, an impression of the ear canal was taken for the test ear of each participant in order to make a custom EM. Ear impressions were sent to a laboratory for fabrication. All EMs were a full-shell style constructed from clear, vinyl material and were tubed with No. 13 standard EM tubing with a 1.9 mm internal diameter and 3.6 mm external diameter. EMs were either unvented due to limitations in ear-canal size or vents were occluded with putty. The tubing of the EM extended to the termination of the sound bore at the canal and extended 2 cm from the lateral surface of the EM to simulate the approximate tubing length needed for a behind-the-ear (BTE) hearing aid. The result was an average transmission-line length of 34.17 mm (range 29–38 mm). Following receipt of the fabricated EM, participants returned for a second session approximately 2 weeks after the first session.

Prior to testing, a calibration of the sound source was conducted with each participant's EM coupled to the transducer to determine the Thevenin-equivalent source impedance and pressure. The source calibration is similar to that used in previous studies to obtain separate estimates of incident and reflected sound levels in the ear canal (Allen, 1986; Keefe *et al.*, 1992; Scheperle *et al.*, 2008). Specifically, five brass tubes with lengths of 83.0, 54.3, 40.0, 25.6, and 18.5 mm were attached to a brass plate at one end. Tube lengths were selected to provide $\frac{1}{2}$ wave resonant peaks at 2, 3, 4, 6, and 8 kHz and allowed for accurate impedance measurements through 10 kHz. A diagram of the brass tubes and 30 mm brass coupler is shown in Fig. 4. The transducer for both sound source calibration and threshold measurements was an

Etymotic Research ER-2A insert earphone. The earphone tube was coupled directly to the EM tubing in the same manner used for a BTE hearing-aid coupling.

To record sound levels in the ear canal, a probe-microphone tube (ER-7C, Etymotic Research) was placed 4 mm past the termination of the EM sound bore either by threading the probe tube through a parallel pressure vent in the EM or by attaching the probe tube to the outside of the canal portion of the EM using 3M Transpore surgical tape. A distance of 4 mm was chosen to approximate clinical probe distances recommended in previous studies. Stimulus presentation and probe-microphone recording were processed by a personal computer using a Digital Audio Laboratories Card-Deluxe sound card. Ear-canal responses were recorded digitally with a sampling rate of 32 kHz and saved for later analysis.

First, the Thevenin-equivalent source characteristics were derived for each participant's EM. Each EM was attached to the brass coupler using adhesive putty. Broadband noise with a flat (voltage) spectrum and 256 ms duration was presented at 61 dB SPL (as measured in a 2 cc coupler) through the EM to each of the five cavities. Using custom software (EMAV, Neely and Liu, 1994), multiple measurements were taken and averaged in each brass tube to reduce the noise level. The obtained source impedance and frequency response for each brass tube were compared to the expected impedance and frequency response based on the length of the tubes, and an error was calculated based on the deviation from the expected response. If significant deviations between the measured and expected responses were obtained, the EM coupling was adjusted and measurements were repeated until the obtained output closely approximated the expected output for all five brass tubes.

Based on the acoustic response as measured at 4 mm past the termination of the sound bore in each of the brass tubes, a Thevenin-equivalent source pressure and impedance was derived for each individual's EM. Next, the same broadband noise was presented through the EM in the participant's ear to determine the pressure and load impedance characteristics of the ear canal at the same probe position. In rare cases, acoustic measures revealed that placement of the EM collapsed the probe tube, resulting in a significantly reduced acoustic response. In these instances, the EM was removed and repositioned. Custom software determined the frequency corresponding to the minimum ecSPL in each participant's ear canal (i.e., the notch frequency). Behavioral thresholds were then measured through the EM at octave frequencies from 0.5 to 8 kHz plus 9 and 10 kHz. To provide additional resolution in the frequency region of the notch, thresholds were also measured at the notch frequency and in $\frac{1}{4}$-octave steps from one octave below the notch to $\frac{1}{2}$ octave above the notch. An automated method of limits with a 5 dB step-size was used until the standard deviation of the threshold response was less than or equal to 2.5 dB at each frequency. Measurements of load impedance were used to derive the three estimates of sound level for each participant's threshold responses. Thresholds were expressed in (a) dB ecSPL as measured at the probe microphone, (b) voltage at the ER-2A transducer (dB re 1 μV), and (c) ear-canal dB FPL measured at the probe microphone, which was derived using load impedance of the ear canal in a manner similar to that used by Scheperle *et al.* (2008). The decibel relative to transducer voltage was used as a reference because auditory thresholds should not be influenced by local pressure minima resulting from interactions between incoming and outgoing sounds in the ear canal.

TABLE II. Notch frequencies and thresholds for each subject in ecSPL, FPL, and transducer voltage (dB re 1 μV).

Subject	Notch (Hz)	ecSPL	FPL	dB μV
1	8623	23.2	27.1	24.1
2	8289	−4.7	−0.5	16.5
3	7436	−9.8	4.4	19.8
4	7429	22.5	38.9	34.7
5	6965	−4.8	7.4	22.1
6	5907	−1.5	9.7	25.1
7	5773	−20	5	24.4
8	5591	−1.5	6.7	19.6
9	5152	7.1	19.4	29.7
10	3958	14.4	18.5	15.7
11	2933	22.5	22	15.6
12	2848	3.4	4.9	11.7
Mean	5908.67	4.23	13.63	21.59
Std	1942.65	13.98	11.61	6.57

VI. RESULTS: EXPERIMENT 2

The frequency of the minimum ecSPL (i.e., the *notch* frequency) and corresponding values for both transducer voltage and FPL are presented in Table II. Comparisons of sound levels at the notch frequency reveal that 10 of 12 subjects had FPL responses greater than ecSPL responses, while 2 subjects had no difference between FPL and ecSPL (S11 and S12). Figure 5 shows the mean threshold data in relative pressure as a function of frequency in ecSPL, FPL, and transducer voltage (dB μV) for all 12 subjects in experiment 2. The profile of threshold values was similar across all three estimates of sound level with two exceptions. First, at frequencies ≤ 2 kHz, the mean ecSPL response is greater than the mean ear-canal FPL response by approximately 6 dB. In the frequency region >2 kHz corresponding with the minimum ecSPL, FPL was more similar to dB μV than ecSPL. A repeated-measures ANOVA for frequencies ≤ 2 kHz indicated that the mean difference between ecSPL, FPL, and voltage was significant $[F_{2,80}=9.90,\ p=0.003,\ \eta^2_p=0.198]$. Post hoc tests using Bonferroni adjustments for multiple comparisons $(p<0.0167)$ indicated that at low frequencies, SPL was significantly higher than FPL, with a mean difference of 5.5 dB. However, the 1.5 dB mean difference between dB voltage and ecSPL for low frequencies was not significant.

An additional repeated-measures ANOVA compared thresholds at the notch frequency. The mean differences between the three estimations of sound level at the notch (ecSPL, FPL, and dB μV) were significant $[F_{2,22}=19.183,\ p<0.001,\ \eta^2_p=0.636]$. Post hoc tests using Bonferroni adjustments for multiple comparisons $(p<0.0167)$ indicated

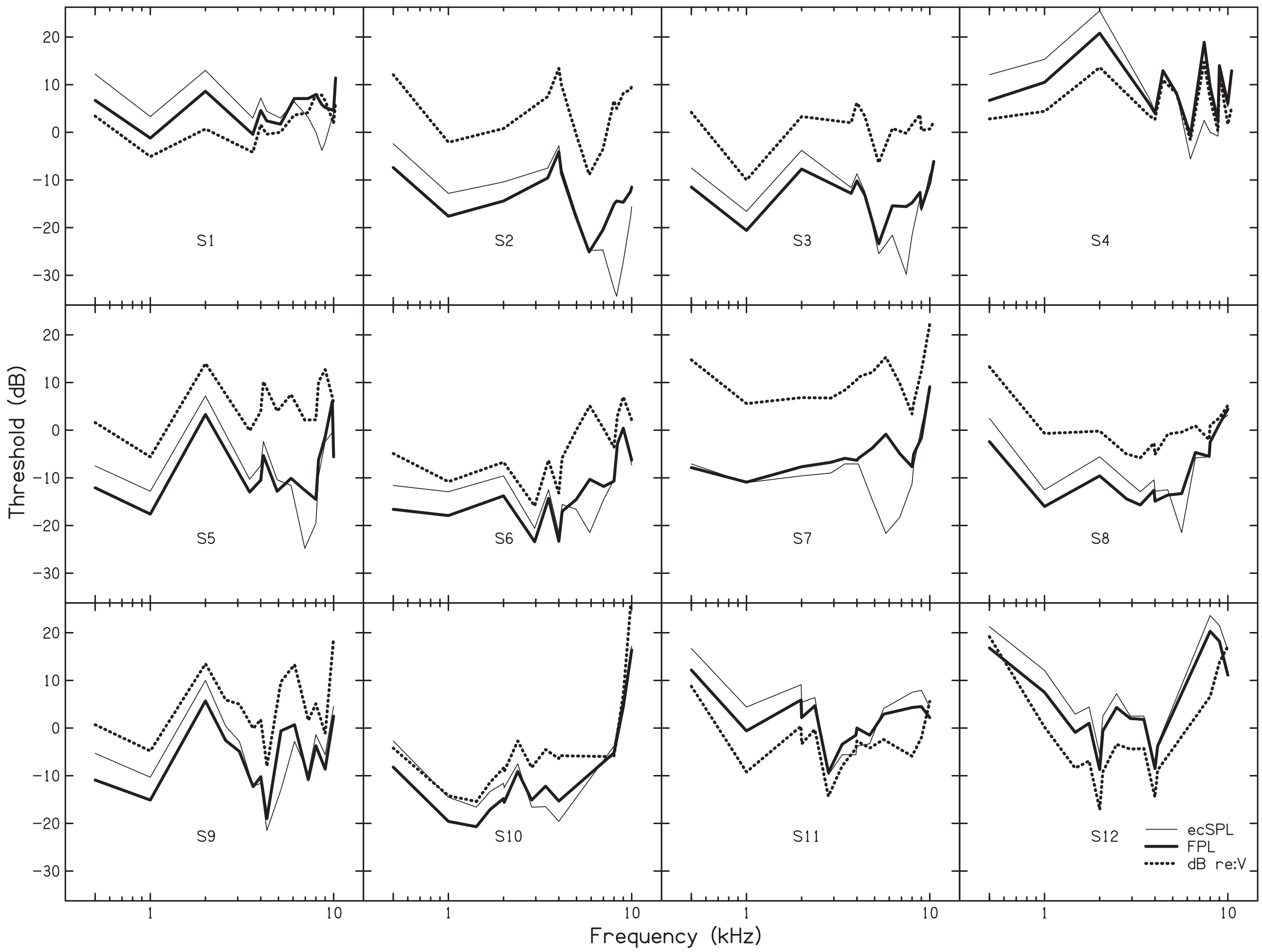

FIG. 5. Behavioral thresholds in relative pressure plotted across frequency for ecSPL (light-shaded line), FPL (dark-shaded line), and dB μV (dotted line). Each panel represents data from an individual subject in descending order by ecSPL notch frequency.

that the differences between all three measures were statistically significant with ecSPL resulting in the lowest estimate of sound level at the notch frequency and FPL resulting in a higher estimate of sound level than ecSPL. The mean difference between ecSPL and dB FPL at the notch frequency was 12.2 dB, indicating that FPL results in higher estimations of ear-canal sound level than ecSPL at the frequency where the influence of standing waves is significant. This finding suggests that thresholds referenced to FPL are less influenced by standing waves at the notch frequency than ecSPL. The variation in FPL near the notch frequency is similar to that of dB μV, which is assumed to be independent of measurement error from acoustic standing waves.

Figure 6 displays the mean threshold data across participants normalized to the notch frequency. To assess the relationship between ear-canal measures of ecSPL, FPL, and dB μV thresholds, a series of linear regression models was calculated and compared using R^2-change tests for nested models. Given that the pattern of the relationships between FPL, ecSPL, and dB μV was different at frequencies below 2000 Hz compared to frequencies above 2000 Hz, the regression models for comparison across the notch frequency were constrained to frequencies greater than 2000 Hz. The full regression model with dB μV as the criterion and both ec-

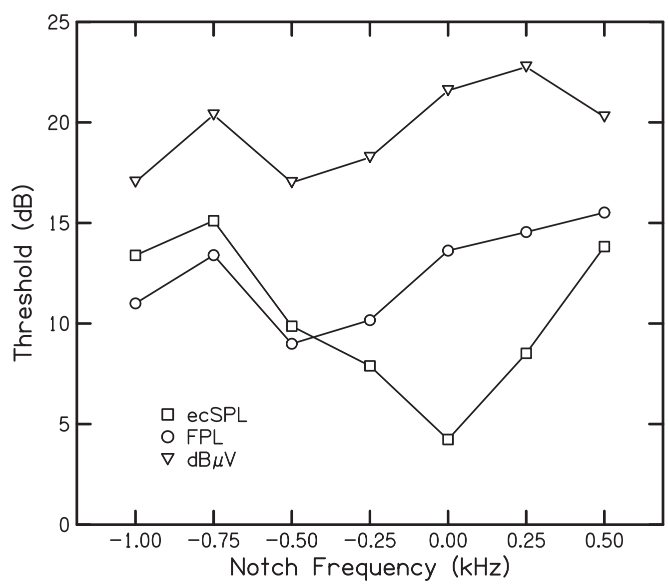

FIG. 6. Mean behavioral thresholds in relative pressure plotted as a function of the range from one octave below the notch frequency to one-half octave above the ecSPL notch frequency. Inverted triangles represent thresholds in dB μV, circles represent thresholds in FPL, and squares represent thresholds in ecSPL.

SPL and FPL as predictors had an $R^2=0.351$, [$F_{2,136}=36.731$, $p<0.001$]. Both ecSPL and FPL had significant regression weights at frequencies above 2000 Hz, but the standardized regression weight (β) for FPL was positive, while β for ecSPL was negative. The pattern of results suggests that as dB μV increases, FPL also increases for frequencies around the notch. Additionally, ecSPL decreases compared to dB μV for the range of frequencies around the notch, consistent with observed patterns of standings waves measured in the ear canal.

Nested comparisons were completed with ecSPL and FPL as individual predictors of dB μV for frequencies above 2000 Hz to determine if either FPL or ecSPL alone was able to predict thresholds in dB μV, as well as the full model that contained both estimates. The linear regression model with ecSPL as the predictor of dB μV threshold revealed an $R^2=0.168$ [$F_{1,137}=27.63$, $p<0.001$]. While ecSPL was found to have a significant relationship with dB μV at frequencies above 2000 Hz, the model with only SPL as the predictor did not predict thresholds in dB μV as well as the full model (R^2-change$=-0.183$, $F_{1,136}=38.31$ $p<0.001$) with significantly less variance accounted for. The linear regression model with FPL as a predictor of dB μV threshold had an $R^2=0.303$, [$F_{1,137}=59.483$, $p<0.001$]. Comparison of the FPL as the sole predictor of dB μV threshold was significantly better than the ecSPL-only model (R^2-change $=0.135$, $F_{1,137}=48.45$ $p<0.001$) and was not significantly different from the model with both ecSPL and FPL as predictors of dB μV threshold (R^2-change$=-0.48$, $F_{1,136}=10.23$ $p=0.15$).

VII. DISCUSSION: EXPERIMENT 2

Results from experiment 1 indicated that probe placement at or near the TM did not reduce the influence of pressure minima on probe-microphone measurements for any of the participants. These findings highlight the difficulty of using distance from the TM as a clinical guideline for probe-microphone measurements. Using a method described by Scheperle *et al.* (2008), FPL for behavioral threshold was calculated, in addition to voltage at the transducer and ecSPL. FPL is an estimate of ear-canal sound level that separates incoming and outgoing pressures in the ear canal when the impedance of the ear canal is known. The primary hypothesis of experiment 2, based on theoretical expectations, was that FPL would be less affected by standing waves than ecSPL, and more similar to the voltage at the transducer in the shape of its frequency response near the notch frequency.

Data from experiment 2 confirmed two distinct patterns of results for the relationship between the three estimates of sound level. At frequencies ≤ 2000 Hz, ear-canal measurements in ecSPL are higher than FPL measurements by approximately 6 dB. This pattern was statistically significant and observed consistently across all 12 subjects. Just as pressure minima at the notch frequency are the result of cancellation between incoming and outgoing sound pressures with opposing phases, the enhancement at frequencies ≤ 2000 Hz is potentially the result of summation of the two components in phase. The enhancement of ecSPL related to the summation of incoming and outgoing acoustic pressures occurs for frequencies ≤ 2000 Hz, which have wavelengths >6.75 in. Given the fact that these wavelengths exceed the length of the ear canal, this increase in ecSPL should be independent of the location at which measurements are taken in the ear canal.

At frequencies >2000 Hz, ear-canal measurements of threshold in ecSPL exhibit minima that are not evident in either FPL or dB μV. Substantial pressure minima in ecSPL, but not in FPL or dB μV, were observed in 10 of the 12 subjects. The two exceptions (S11 and S12) did not have significant differences between ecSPL and FPL, but did have an enhancement of ecSPL at frequencies ≤ 2000 Hz. An enhancement of ecSPL at lower frequencies should also be accompanied by a corresponding pressure minimum in each case. One explanation could be that both subjects had pressure minima at frequencies above 10 kHz, which were not measured in the current study. Both subjects also had minimum ecSPL values at frequencies close to 2 kHz, which was much lower than the other participants. Given that the relationship between each estimate of sound level (ecSPL and FPL) with dB μV is different for frequencies above and below 2 kHz, the close proximity of the ecSPL notch to 2 kHz may have resulted in no difference between the two estimates. Despite the lack of an apparent pressure minimum in each case, FPL and ecSPL were otherwise similar for both participants above 2000 Hz. This observation is an indication that FPL would not likely lead to additional errors in estimation of *in-situ* hearing-aid gain in cases where a SPL pressure minimum is not measured.

VIII. CONCLUSIONS

The purpose of the current experiments was to identify methods of minimizing the influence of acoustic standing waves in the ear canal on *in-situ* probe-microphone measurements, which are used clinically to estimate hearing-aid gain and output. Current methods are based on placing the probe-microphone tube close enough to the TM to limit the frequency of standing waves to frequencies above the bandwidth of commercially-available hearing aids (e.g., Burkhard and Sachs, 1977). Recent research suggests that there may be advantages to designing hearing aids with usable frequencies as high as 10 kHz to improve speech perception (Stelmachowicz *et al.*, 2001) and perception of sound quality (Moore and Tan, 2003). As manufacturers begin to extend the upper frequency limits of hearing aids, current probe-microphone measurement schemes are unlikely to provide valid measures of ear-canal sound levels. The goal of experiment 1 was to determine if probe placements close to the TM could reduce the influence of standing waves to frequencies approaching 10 kHz. Four probe-microphone placements at different positions relative to the EM and TM were used to measure the ear-canal response in ten subjects. Pressure minima ranging from 12 to 26 dB were present at all four probe placements, and even when the probe tube was placed near the TM, clinically-significant pressure minima were observed below 10 kHz. While the use of a tactile method of probe placement at the TM without visual confirmation might not have resulted in insertion depths as close to the TM as desired, our finding of significant variations in ecSPL for measurements taken near the TM is consistent with previous studies (Dreisbach and Siegel, 2001; Khanna and Stinson, 1985). Although additional efforts to place the probe tube closer to the TM might reduce the influence of standing waves, the potential for patient discomfort and difficulty establishing and maintaining such a TM placement would still limit its clinical utility for hearing-aid verification.

The test-retest reliability of probe-microphone measurements at all positions for each subject was acceptable with differences less than 5 dB over most of the frequency range. More substantial errors were observed with the probe placements closest to the TM, where small changes in probe position are likely to have larger effects on the measured response. Larger deviations between repeated-measurements for the same probe position at frequencies above 4 kHz highlight the difficulty in maintaining consistency across measurements, even when care is taken to maintain a constant probe position. In addition to being potentially uncomfortable for patients and not feasible with children, the current data suggest that placement of the probe microphone at or near the TM does not appear to adequately minimize the impact of standing waves on estimates of *in-situ* sound level. While placement close to the TM resulted in a pressure minimum with an average frequency greater than 6 kHz, 18 of the 40 placements revealed significant pressure minima in the ecSPL at frequencies that would be within the bandwidth of most current hearing aids. This finding suggests that acoustic standing waves in the ear canal compromise the validity of probe-microphone measurements even for hearing aids with an upper bandwidth of 6 kHz. Errors in the estimation of ear-canal sound level of this magnitude may lead to inaccurate assignment of gain and output and could pose potential risk to residual hearing sensitivity.

In experiment 2, FPL was used to obtain an estimate of sound level in the ear canal that is independent of outgoing acoustic reflections from the TM. The hypothesis was that FPL would be more similar to decibel based on voltage at the transducer than ecSPL, since the decibel based on the transducer voltage should be theoretically independent of acoustic reflections. For frequencies >2000 Hz, FPL was found to be a better predictor of dB voltage than ecSPL. At frequencies ≤ 2000 Hz, however, ecSPL was found to be approximately 6 dB higher than FPL. The enhancement of ecSPL at frequencies ≤ 2000 Hz is most likely the result of summation between the incoming and reflected sound energy in the ear canal, which is a counterpart to the cancellation at frequencies >2000 Hz, which results in pressure minima in the estimation of the incoming signal. Because summation occurs at lower frequencies with $\frac{1}{4}$ wavelengths greater than the length of the ear canal, the 6 dB enhancement in ecSPL compared to FPL will be constant along the length of the ear canal and at the TM. Therefore, the 6 dB difference between ecSPL and FPL at low frequencies would only need to be taken into account in hearing-aid verification if probe-microphone measurements are taken in FPL and thresholds are referenced to ecSPL. Alternatively, cancellation has the ability to impact clinical decisions about the assignment of amplification. For higher frequencies where cancellation can result in significant pressure minima, the response of the hearing aid will appear to be inadequate. The frequency, gain, and output characteristics of the hearing aid may be adjusted unnecessarily by the clinician to compensate for what appears to be insufficient gain. The use of FPL for hearing-aid measurements could potentially result in smaller errors in estimating *in-situ* hearing-aid gain by taking into account the influence that reflected sound pressure has on *in-situ* measurements of hearing-aid gain and output.

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Allen, J. B. (**1986**). "Measurement of eardrum acoustic impedance," in *Peripheral Auditory Mechanisms*, edited by J. B. Allen, J. L. Hall, A. Hubbard, S. T. Neely, and A. Tubis (Springer-Verlag, New York), pp. 44–51.
ANSI (**2003**). ANSI S3.22-2003 Specification of Hearing Aid Characteristics (American National Standards Institute, New York).
Boothroyd, A., and Medwetsky, L. (**1992**). "Spectral distribution of /s/ and the frequency response of hearing aids," Ear Hear. **13**, 150–157.
Burkhard, M. D., and Sachs, R. M. (**1977**). "Sound pressure in insert earphone couplers and real ears," J. Speech Hear. Res. **20**, 799–807.
Caldwell, M., Souza, P. E., and Tremblay, K. L. (**2006**). "Effect of probe tube insertion depth on spectral measures of speech," Trends Amplif. **10**, 145–154.
Chan, J. C., and Geisler, C. D. (**1990**). "Estimation of eardrum acoustic pressure and of ear canal length from remote points in the canal," J. Acoust. Soc. Am. **87**, 1237–1247.
Dirks, D. D., and Kincaid, G. E. (**1987**). "Basic acoustic considerations of ear canal probe measurements," Ear Hear. **8**, 60S–67S.
Dreisbach, L. E., and Siegel, J. H. (**2001**). "Distortion-product otoacoustic emissions measured at high frequencies in humans," J. Acoust. Soc. Am. **110**, 2456–2469.
Farmer-Fedor, B. L., and Rabbitt, R. D. (**2002**). "Acoustic intensity, impedance and reflection coefficient in the human ear canal," J. Acoust. Soc. Am. **112**, 600–620.
Gilman, S., and Dirks, D. D. (**1984**). "A probe EM system for measuring eardrum SPL under hearing-aid conditions," Scand. Audiol. **13**, 15–22.
Gilman, S., and Dirks, D. D. (**1986**). "Acoustics of ear canal measurement of eardrum SPL in simulators," J. Acoust. Soc. Am. **80**, 783–793.
Hellstrom, P. A., and Axelsson, A. (**1993**). "Miniature microphone probe tube measurements in the external auditory canal," J. Acoust. Soc. Am. **93**, 907–919.
Keefe, D. H., Ling, R., and Bulen, J. C. (**1992**). "Method to measure acoustic impedance and reflection coefficient," J. Acoust. Soc. Am. **91**, 470–485.
Khanna, S. M., and Stinson, M. R. (**1985**). "Specification of the acoustical input to the ear at high frequencies," J. Acoust. Soc. Am. **77**, 577–589.
Killion, M. C., and Tillman, T. W. (**1982**). "Evaluation of high-fidelity hearing aids," J. Speech Hear. Res. **25**, 15–25.
Moeller, M. P., Hoover, B., Putman, C., Arbataitis, K., Bohnenkamp, G., Peterson, B., *et al.* (**2007**). "Vocalizations of infants with hearing loss compared with infants with normal hearing: Part I—Phonetic development," Ear Hear. **28**, 605–627.
Moore, B. C., Stone, M. A., Fullgrabe, C., Glasberg, B. R., and Puria, S. (**2008**). "Spectro-temporal characteristics of speech at high frequencies, and the potential for restoration of audibility to people with mild-to-moderate hearing loss," Ear Hear. **29**, 907–922.
Moore, B. C., and Tan, C. T. (**2003**). "Perceived naturalness of spectrally distorted speech and music," J. Acoust. Soc. Am. **114**, 408–419.
Neely, S. T., and Gorga, M. P. (**1998**). "Comparison between intensity and pressure as measures of sound level in the ear canal," J. Acoust. Soc. Am. **104**, 2925–2934.
Neely, S. T., and Liu, Z. (**1994**). "EMAV: Otoacoustic emission averager,"

Technical Memo No. 17 (Boys Town National Research Hospital, Omaha, NE).

Ricketts, T. A., Dittberner, A. B., and Johnson, E. E. (**2008**). "High-frequency amplification and sound quality in listeners with normal through moderate hearing loss," J. Speech Lang. Hear. Res. **51**, 160–172.

Scheperle, R. A., Neely, S. T., Kopun, J. G., and Gorga, M. P. (**2008**). "Influence of in situ, sound-level calibration on distortion-product otoacoustic emission variability," J. Acoust. Soc. Am. **124**, 288–300.

Stelmachowicz, P. G., Lewis, D. E., Choi, S., and Hoover, B. (**2007**). "Effect of stimulus bandwidth on auditory skills in normal-hearing and hearing-impaired children," Ear Hear. **28**, 483–494.

Stelmachowicz, P. G., Pittman, A. L., Hoover, B. M., and Lewis, D. E. (**2001**). "Effect of stimulus bandwidth on the perception of /s/ in normal- and hearing-impaired children and adults," J. Acoust. Soc. Am. **110**, 2183–2190.

Stinson, M. R., Shaw, E. A., and Lawton, B. W. (**1982**). "Estimation of acoustical energy reflectance at the eardrum from measurements of pressure distribution in the human ear canal," J. Acoust. Soc. Am. **72**, 766–773.

Voss, S. E., and Allen, J. B. (**1994**). "Measurement of acoustic impedance and reflectance in the human ear canal," J. Acoust. Soc. Am. **95**, 372–384.

Withnell, R. H., Jeng, P. S., Waldvogel, K., Morgenstein, K., and Allen, J. B. (**2009**). "An in situ calibration for hearing thresholds," J. Acoust. Soc. Am. **125**, 1605–1611.

Zemplenyi, J., Dirks, D., and Gilman, S. (**1985**). "Probe-determined hearing-aid gain compared to functional and coupler gains," J. Speech Hear. Res. **28**, 394–404.