Real Ear Measurements and Digital Hearing Aids: Realities, Myths, and Measurement Techniques

Dave Fabry, PhD

Introduction

Digital hearing aids have dramatically improved the precision with which hearing aids can be adjusted to compensate for conductive and sensorineural hearing loss. Furthermore, the array of advanced signal processing features available in modern hearing aids has expanded well beyond the realm of "digital screwdrivers". These features include adaptive directionality, noise cancellation, feedback reduction, and automatic program selection, to name but a few. Despite these technological breakthroughs, which should theoretically improve benefit, the proportion of clinicians using objective or subjective measures for verification of hearing aid performance has actually declined in recent years (Kirkwood, 2003). That is, clinicians are relying increasingly on the hearing aid software to automatically make the appropriate adjustments for individual patients.

This may be due, in part, to uncertainty regarding which test methods and/or procedures may be employed with sophisticated digital circuitry. In fact, some of this confusion has led to the establishment of several "urban myths" regarding verification of digital hearing aids that are unsubstantiated by data. The focus of the present article is to dispel a few of these myths, and also to demonstrate why clinical assessment of advanced hearing aid features is an important component of the hearing aid fitting process.

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Figure 1

Difference in measured 2-cc coupler gain and software-simulated 2-cc coupler gain for 28 hearing aids. Negative values indicate that the measured 2-cc coupler gain is less than the software-simulated 2-cc coupler gain. From Hawkins and Coor (2003), Fig 1.
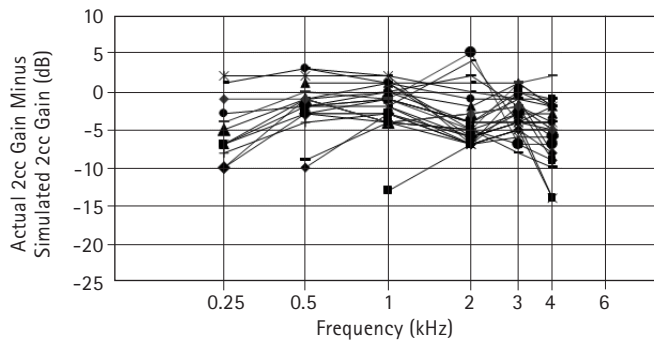

Figure 2

Difference between measured insertion gain and software-simulated insertion gain for 12 patients. Negative values indicate that the measured insertion gain is less than the software-simulated insertion gain. From Hawkins and Coor (2003), Fig 2.
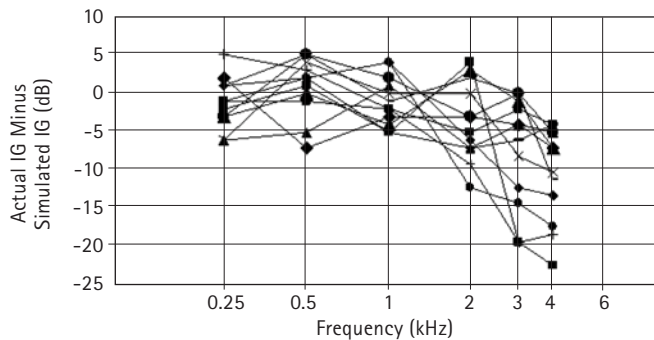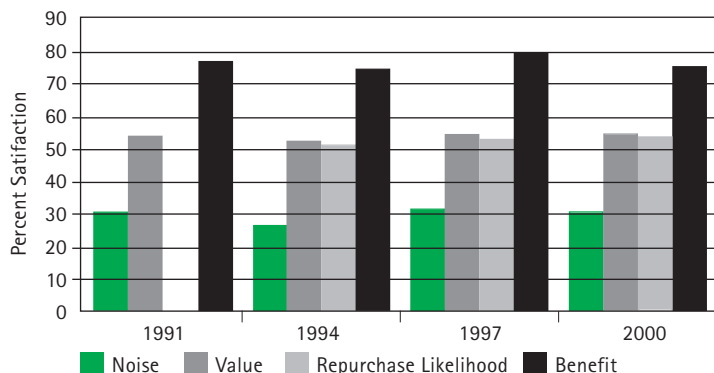

Figure 3

Customer Satisfaction with value, benefit in noise, overall benefit and likelihood of repurchasing current brand of hearing instrument (hearing instruments <3 years of age; source MarkeTrak III (1991) – MarkeTrak VI (2000)).


Digital "Mythology"

The most popular myth related to verification of hearing aid fitting is that the manufacturer's fitting software may be used to accurately predict target gain and output values for individual patients. Although the simulated gain and output settings displayed in fitting software are the product of extensive measurements of hearing aid function, they nonetheless represent data for average ears. A recent study by Hawkins and Cook (2003) indicated that simulated coupler- and insertion-gain values often over-predicted measured values (Figures 1-2).
In some cases, the differences exceeded 15 dB; differences of this magnitude could certainly impact user satisfaction and benefit, and it is perhaps no coincidence that the recent trend towards relying solely on predicted, nominal insertion gain values parallels the plateau in perceived value, benefit, and hearing aid performance in noise during the past decade (Figure 3). It is imperative that the unparalleled flexibility provided by digital hearing instruments be used to optimize hearing aid performance for individual ears, using clinically efficacious methods.

Another popular myth suggests that accurate real-ear measurements are not possible with digital hearing aids. This stems from the popular misconception that non-linear gain and noise reduction circuitry prevent accurate assessment of digital hearing aids. In fact, the basis for this erroneous assumption is grounded in reality: although some digital hearing aids are intelligent enough to determine whether they are at home or at a party, they are not smart enough to figure out when they are in a test box. That is, many digital hearing aids employ a noise cancellation system that interferes with the test signal used by many real-ear measurement systems. In most cases, measured insertion gain underestimates actual use gain, because the noise canceller reduces gain in an attempt

Figure 4

2-cc coupler measurements for a digital hearing aid with noise cancellation activated, using composite noise, for three test conditions: for "light" and "moderate" settings, and when noise cancellation has been tuned off.
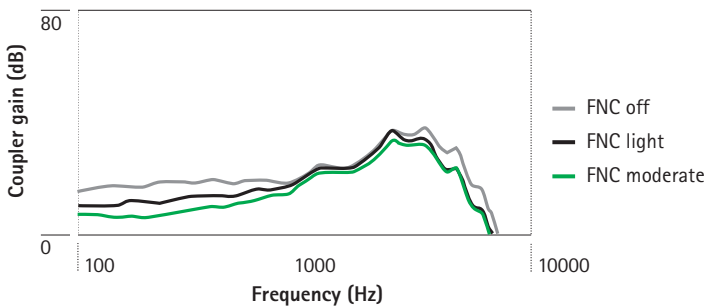
to "eliminate" the noise. This is illustrated in Figure 4, where coupler gain is plotted for three conditions (noise cancellation off, "light", and "moderate"). One solution that some hearing aid manufacturers have employed is the use of a test mode that removes the noise canceller from the circuit for evaluation, but this prevents the circuit from operating as it would under actual use conditions. Another approach used by many clinicians is to quickly switch the test signal on- and off- before the noise canceller acts to reduce gain and output. The developers of real-ear measurement systems have provided an even better answer to the problem: by adapting the spectral and/or temporal properties of the test signal, the noise cancellation system identifies it as speech, rather than noise (e.g. ICRA noise). In addition, several real-ear manufacturers have developed test systems that permit live or recorded speech passages to be used as the test signal, which provides additional test validity. Consequently, insertion gain measured in this fashion provides a precise and repeatable indication of real-world performance with hearing aids that employ noise reduction in combination with non-linear signal processing. The remaining challenges however, include the development of phonetically and phonemically balanced speech materials that are valid and clinically efficient.

Another misnomer relates to the inability of prescriptive target formulae to be used with non-linear hearing aids. Because most digital hearing aids use multiple-band compression that change gain as a function of input, the "old fashioned" target formulae (e.g. NAL-R, Byrne & Dillion, 1986) cannot fully describe their performance because they used a single target gain characteristic. Newer prescriptive formulae (e.g. DSL [i/o], Cornelisse et al, 1995) provide gain formulae that change across input, to optimize intelligibility, loudness, and/or comfort. Some manufacturers have taken the matter into their own hands, by developing proprietary target formulae for their hearing aids. Although this is understandable, given the complex signal processing used in modern hearing aids, it also makes verification of fitting targets a challenge. That is, although most real-ear test equipment includes NAL-NL1 (Byrne et al., 2001) or DSL 4.1a (Seewald et al., 1997) formulae, manufacturer's proprietary target values must be provided to clinicians in order to use them clinically with individual patients.

The bottom line is that clinicians working with the latest technology should upgrade their real-ear test equipment to include "digital" test signals or "real" speech. In essence, use of only the nominal gain values provided by the manufacturer's software is analogous to using horoscopes rather than telescopes to forecast movements of the stars and planets. Reliance on mythology is sometimes effective, but rarely repeatable. Similarly, use of software settings alone will occasionally provide patient satisfaction, but use of the appropriate tools will enable more precise "acoustical matching" of hearing aids to individual ears. These measurements will not guarantee success, but they will ensure an efficient means for separating "facts" from "myths" in the hearing aid fitting process.

Suggested Basic Test protocol

Test Stimuli As stated above, the choice of stimulus is extremely important. Clinicians may wish to consider using actual speech, as it provides the most realistic stimulus for evaluation of advanced signal processing features (e.g. multiple-band compression and noise reduction). That said, keep in mind that "live" speech suffers from the same shortcomings related to variability as monitored-live-voice (MLV) presentation of monosyllabic word lists used for diagnostic audiologic evaluation. For that reason, recorded, calibrated speech materials are the preferred stimulus of choice, but these have not been widely developed at the present time. An additional barrier is provided by the challenge to develop a "universal" speech stimulus that may be used across all countries and languages. Therefore, at the present time, digital speech noise (e.g. ICRA, digital composite noise) provides a reasonable compromise between the face validity of "real" speech and the ease of use and calibration of noise-based stimuli.

Another stimulus-related issue pertains to comparisons between measurements made in the ear canal with those made in the 2-cc coupler. Typically, 2-cc coupler measurements have been made using pure-tone stimuli (e.g. ANSI S3.22-1996, ANSI S3.22-2003), while real-ear measurements have migrated towards broad-band stimuli in recent years. Simply put, a steady-state or pulsed broad-band test stimulus is more typical of the complex input signals that hearing aids process in "real-world" listening environments than are pure-tone stimuli (particularly for hearing aids that use level-dependent gain circuitry). When comparing coupler-gain measurements to real-ear measurements (i.e. real-ear-to-coupler difference measurements) with digital hearing instruments, it is important to use the same stimuli for both measurements. Be aware that pure-tone stimuli and broad-band stimuli differ in terms of their overall

levels and their spectrum levels. By their very nature, broad-band sounds (like speech) are defined in terms of their overall level (RMS), frequency spectrum, and crest factor (Figure 5).

Figure 5

Characterization of broad-band noise test signal, in terms of frequency response, crest factor, and overall level (from ANSI S3.42: 1992).
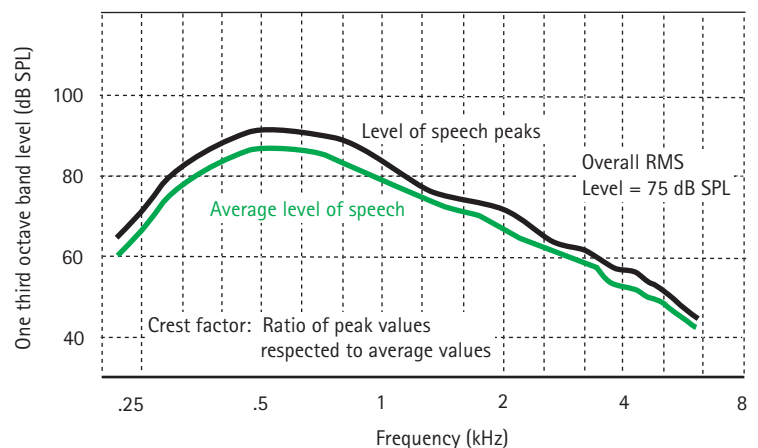
Pure-tone stimuli, on the other hand, are defined in terms of their frequency and intensity, which by definition, defines their spectrum level (the intensity per Hz). That is why it is critical that the same stimulus be used to compare 2-cc coupler to real-ear levels, and also between estimated software values and those measured on individual patients. For digital hearing aids, broad-band stimuli are preferred for gain measurements with multiple-band compression hearing aids, as swept pure-tone stimuli may interact in unusual ways with the transition regions between adjacent frequency bands. Familiarity, however, breeds content, and new ANSI standards continue to be developed for pure-tone stimuli, although standards for broad-band stimuli exist (ANSI S3.42-1992). If "digital" noise or real speech signals are not available, clinicians should either 1) use very brief noise pulses to measure gain and output, or 2) disable noise reduction in software, if possible.

Suggested test procedure

A clinically relevant protocol for initial verification of non-linear hearing aids is governed by the motto that hearing aid gain should be seen AND heard. That is, because real-ear measurements test gain – not hearing – clinicians should ensure that they are not overly reliant on what they see on the computer screen. Behavioral measurements, such as functional or in-situ (through the hearing aid) gain, provide an excellent confirmation that insertion gain is actually translated into useful audible information. Although the focus has been on the development of new prescriptive formulae that employ multiple fitting targets for different input levels, it is not necessary to

"throw the baby out with the bath water". That is, although the traditional prescriptive formulae (NAL-R, Berger, POGO) were developed for linear hearing aids, they still provide very reasonable mid-level targets for non-linear hearing aids. The important point for clinicians is that they must also measure how non-linear hearing aids preserve audibility for low input levels and prevent discomfort for intense sounds. This does not need to be a daunting task. Extensive documentation exists regarding multiple target fitting formulae, and yet many clinicians are not convinced of their "real-world" relevance. What follows is a clinically efficient alternative approach for ensuring that soft sounds are audible, moderate sounds are comfortable, and loud speech and other sounds are not uncomfortably loud.

Conversion of audiometric information to SPL

Ideally, all unaided and aided measures should be expressed using a consistent reference point (dB SPL in the ear canal) and plotted on the same SPLogram (Figure 6). As a result, audiometric thresholds may easily be converted from dB HL to dB SPL in the ear canal, and most real-ear measurement equipment uses conversion values to make this possible. The same conversion applies to loudness discomfort levels, if they were measured clinically. Alternately, loudness discomfort values may be predicted from audiometric threshold data and converted to dB SPL. Keep in mind, however, that unless real-ear measurements are made, these conversions use averaged data.

Figure 6
"SPLogram", as characterized initially by David Pascoe (1975, 1978).
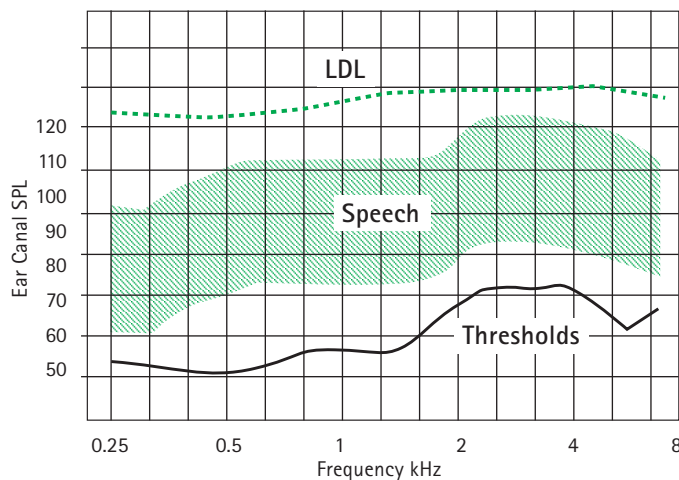
Map residual auditory area on SPLogram.
After thresholds and discomfort thresholds (calculated or measured) have been plotted on the SPLogram, the clinician's favorite prescriptive target may be applied for mid-level targets, and real-ear measurements may be used to match this target at 65 or 70 dB SPL. The recommended stimuli for this measurement are speech weighted noise, "digital" noise (for hearing aids employing noise reduction) or "real" speech (again, for digital instruments with noise reduction). Although the reference for the SPLogram is SPL at the patient's eardrum, this is not practical (or comfortable!) for most patients. Instead, real-ear measurements should be made using a probe-microphone that has been placed within 1.0 cm of the eardrum, which provides measurements of the eardrum SPL within +/- 2 dB for frequencies at 4000 Hz and lower (Figure 7). The two most important factors with real-ear measurements are adequate depth of insertion and consistent placement for unaided and aided measurements. For adults, these objectives may be accomplished by inserting the probe tube microphone 28 mm beyond the tragus; for children, insertion should be 20-25 mm beyond the tragus. Although it is theoretically possible to estimate or measure the SPL through the hearing aid, by comparing hearing aid input with output at the receiver, the impact of standing waves in the ear canal and/or the patient's individual ear canal resonance may result in differences in excess of 10-15 dB from eardrum SPL.

Figure 7

Example of a hearing aid that has been "acoustically matched" to an individual patient's ear, as evidenced by smooth real-ear aided response (REAR), particularly between 2000 and 4000 Hz.
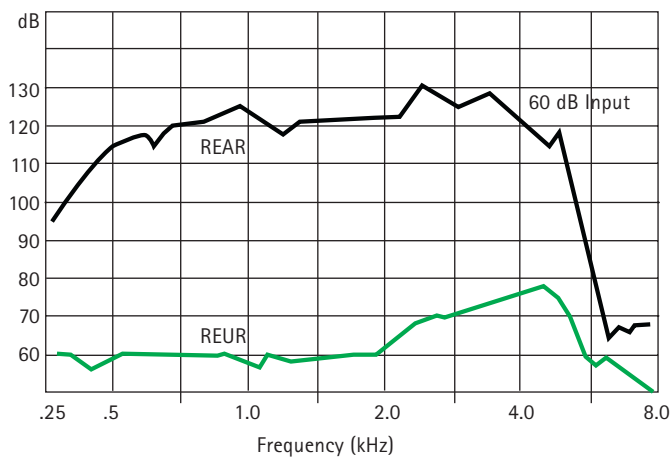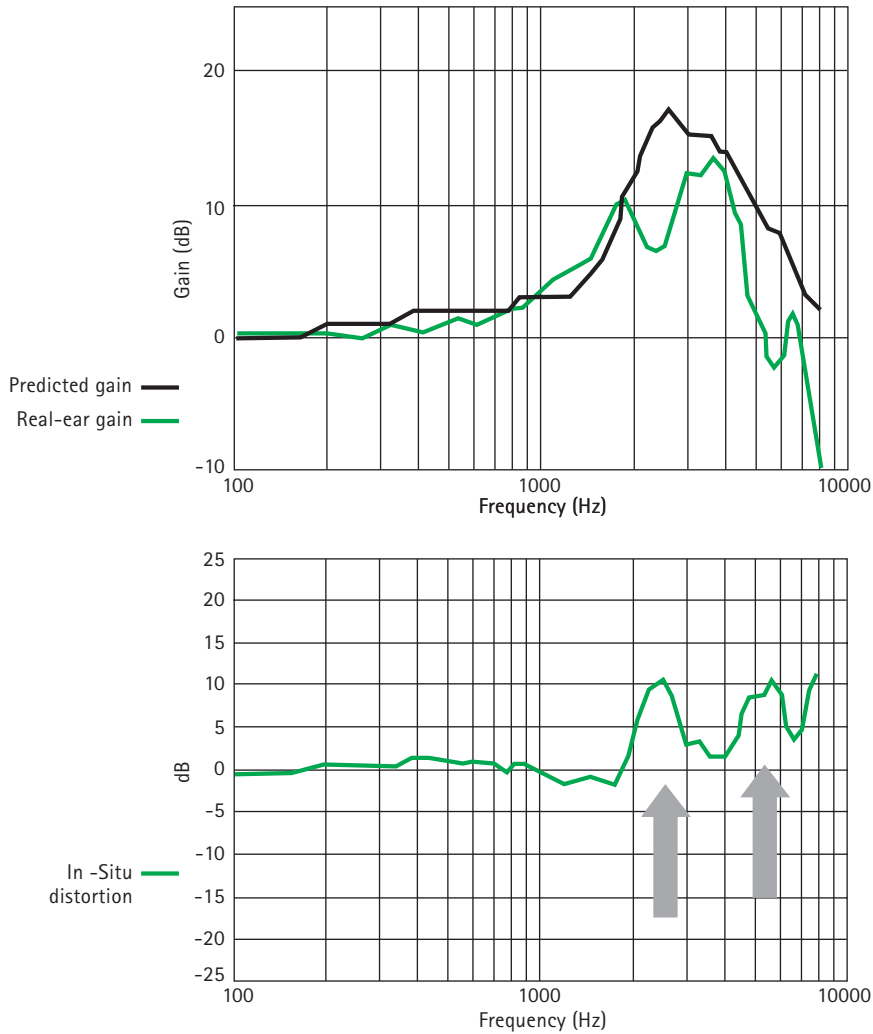
Figure 8

Example of poor "acoustic match" between hearing aid frequency response and patient's ear, as evidenced by numerous peaks in the gain curve.


"Acoustically match" hearing aids to patient's ear

Ideally, the first step for both adults and children is to ensure that the hearing aid assembly (hearing aid shell, venting, earmold, and microphone location) provides a smooth, "acoustic match" to the individual patient's ear canal, concha, and pinna characteristics. The most direct way to accomplish this is to measure the patient's REUR, followed by a real-ear aided response (REAR) that takes into consideration all of these factors for a given input stimulus. The objective is to match prescriptive target, while also providing a smooth REAR, particularly in the frequency region from 2000–4000 Hz (Figure 8). The presence of numerous resonance peaks in the REAR or REIR indicates poor acoustic matching, and will often lead to subjective complaints of poor sound quality.

For some patients, it may be difficult to obtain repeated real-ear measurements; in these cases, acoustical matching may be accomplished via measurement of the coupler response for flat insertion gain (CORFIG), which comprises three primary elements: the real-ear-to-coupler difference (RECD), real-ear unaided gain (REUG), and microphone location effects (MLE) of the selected hearing aid used by the patient. Theoretically, all three measurements must be completed to accurately describe the CORFIG, but research is currently underway to investigate whether the RECD measurement may provide comparable satisfaction and benefit to the complete CORFIG. As a practical measure, RECD is easy to measure clinically on children and adults, and may actually be incorporated into the fitting process for digital hearing aids, without the need for additional verification measurements (Figure 9). Several studies have indicated that results from an "automatic" RECD measurement are comparable to "traditional" measurements of RECD (Munro, 2004). As a result, these studies suggest that RECD is accurate and clinically efficacious, which may help reverse the trend in the US towards clinicians relying exclusively on estimated, average insertion gain settings from hearing aid software (Kirkwood, 2003). As a long as the test method is reliable, acoustic matching by these means will be far superior to using average CORFIG values and may approach results obtained via real-ear measurements for both adults and children. Further research, however, is required before the latter conclusion may be made.

Figure 9

RECD measurement using integrated SPL measurement and coupler comparison to calculate RECD directly.
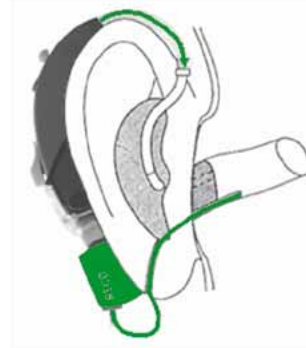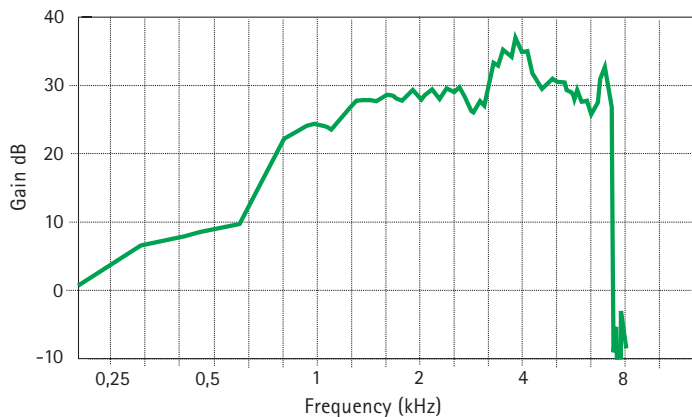

Verification of prescriptive fitting targets for conversational–level speech levels

Regardless of the fitting method selected (e.g. NAL, POGO, BERGER, NAL-NL, DSL), clinicians should begin with a 65-70 dB SPL input stimulus, and match prescriptive targets within +/- 5dB over the range from 250- 4000 Hz. Much debate has been centered on the precision with which target is matched, but the reality is that this process is the beginning – not the end – of the fitting process, and it is better to provide a smooth real-ear aided response (REAR) within 5 dB of targets across a broad range than it is to become obsessively devoted to a "perfect" match. Keep in mind that using pure-tone stimuli in the coupler or real-ear may provide an uneven or jagged response (Figure 10) for some digital hearing aids, but this is stimulus dependent; broad band, "speech like" stimuli will minimize these irregularities. This measurement ensures audibility for conversational level speech.

Figure 10

Pure-tone 2-cc coupler gain measurement of a digital hearing aid programmed to use gain settings.

Verification of prescriptive gain targets for soft speech levels

Subsequent to setting appropriate gain and frequency response for conversational speech, real-ear measurements or in-situ gain may be used to ensure audibility for soft speech and other sounds. This may be accomplished either by visually ensuring that the REAR for a 50 dB SPL broad-band stimulus used with real-ear measurements exceeds thresholds plotted on the SPLogram (Figure 11), or by adjusting maximum gain (e.g. gain 50) until aided threshold for narrow-band stimuli approximates 25-30 dB HL (Figure 12). This assessment "optimal aided threshold" verifies that soft sounds are audible.

Figure 11

Real-ear measurements, using SPLogram format, that indicate: 50 dB SPL composite signal is suprathreshold (above "T" values), 65 dB SPL stimulus matches NAL-R target, and 90 dB brief tone burst stimuli do not exceed patients LDL (UCL on chart, indicated as "U").

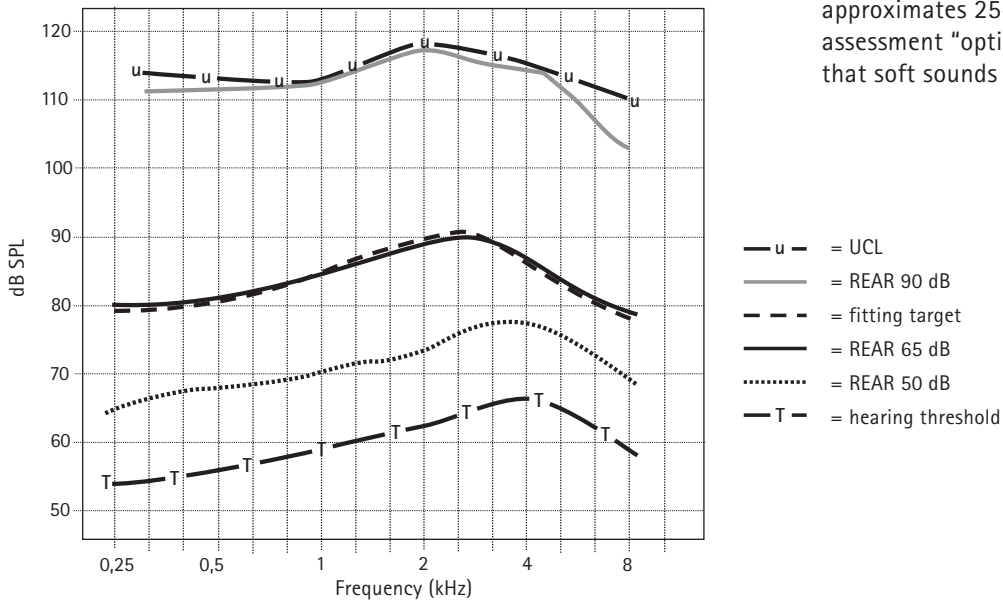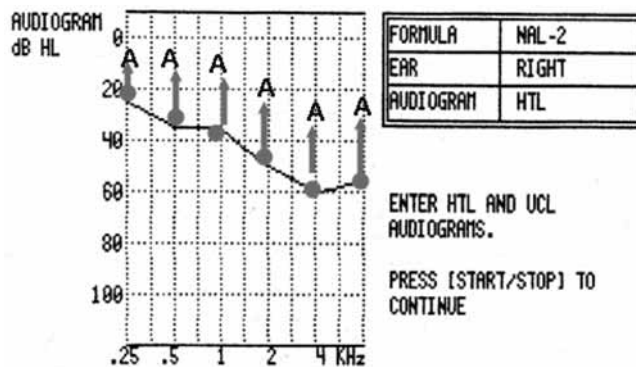|	
—u—	= UCL
	= REAR 90 dB
– – –	= fitting target
——	= REAR 65 dB
·········	= REAR 50 dB
—T—	= hearing threshold

Figure 12

Aided "optimal" thresholds, plotted on the audiogram, indicated that the patient's thresholds in quiet are approximately 25-30 dB HL when evaluated in an audiometric test booth.

Use real-ear measurements to verify hearing aid MPO

The final step in the initial verification process is to make certain that loud sounds do not exceed the patient's loudness discomfort thresholds. This is an extremely important step, particularly for pediatric patients and for those with limited residual dynamic range of hearing. Unfortunately, this step is often omitted (or minimized) from clinical verification procedures, for fear of producing loudness discomfort in the patient. The reality is that when hearing aid MPO is set accurately (neither too high nor low relative to the patient's loudness discomfort level [LDL]), input dynamic range is optimized, particularly for patients with greater degrees of hearing loss. Fortunately, many of the latest hearing aids permit frequency specific MPO adjustments to be made, and this flexibility should be incorporated into the fitting process. Ideally, a number of pure-tone or narrow-band stimuli (equivalent to the number of independently adjustable compression bands) should be presented at 85 dB SPL at 0 degrees azimuth. REAR (in this case, also called real-ear saturation response, or RESR) measurements should approach, but not exceed, measured or predicted LDL for that frequency range (recall that the patient's LDL in HL has already been converted to ear canal SPL). After setting hearing aid MPO independently for all compression bands, a broad-band noise or speech input should be presented at 85 dB SPL, to verify that loudness summation across bands does not produce discomfort. Keep in mind that if spectrum level (level per Hz) is the same for a pure-tone and broad-band noise, the overall level differences may be as much as 15-20 dB. Because most real-ear test systems present data in terms of overall level, LDLs measured with broad band noise will be lower than for pure-tone levels. Although clinicians need to exercise caution when evaluating RESR, it is perhaps the most important stage of clinical verification. If the MPO is set properly, the hearing aids will optimize residual dynamic range, while protecting the user from loudness discomfort in the "real world". Figure 11 shows a completed SPLogram with a good match to low, moderate, and high prescribed target gain values.

## Assessment of advanced signal processing features

The final step in the initial verification protocol is to evaluate whether advanced signal processing features are functioning properly. According to recent surveys, directional microphones now comprise nearly 30% of new hearing aids sold, but very few clinicians use any form of empirical evaluation of their performance. This is not to suggest that practitioners need to purchase an anechoic chamber to make their own polar plot measurements, but rather to provide a few additional measurements that provide information regarding how well the directional microphones are functioning. First, it is useful and easy to determine whether an equalized or non-equalized directional microphone is in use. With the speaker located at 0 degrees azimuth, measure and store an REAR in omni-directional mode for an input of 65 dB SPL. Without changing the volume control, engage the directional microphone

and repeat the measurement. If the REARs of the two measurements overlap, an equalized directional microphone is in use (Figure 13). More commonly, if the directional microphone response indicates a low-frequency roll-off below 1-2 kHz, a non-equalized directional microphone is in use. One of the reasons for the "wow" effects often reported with directional microphones is this low-frequency rolloff, which is independent from the directivity benefits afforded when speech and noise are spatially separate. This low-frequency attenuation is not a problem for most hearing aid users with mild to moderate hearing losses, but it may affect audibility for patients with severe-to-profound hearing loss. As a result, clinicians may want to be sure that speech audibility is not affected if a non-equalized directional microphone is used; if so, they may wish to reprogram the hearing aid using a customized program to compensate for this attenuation. Without verification, however, clinicians will be leaving this issue to chance.

Another measurement that clinicians may wish to make is a real-ear front-to-back ratio (FBR). Using the same 0 degree azimuth for loudspeaker placement, measure REAR, using a digital noise signal or real speech presented at 65 dB SPL. Store this measurement, then turn the patient around (a swivel chair works well for this) until the loudspeaker is now located at 180 degrees and repeat the measu-

Figure 13

Real-ear measurements for equalized and non-equalized directional microphones, compared to omnidirectional microphone, for composite noise measurements at 0 degrees azimuth.

— Omnidirectional microphone
— Equalized directional microphone
— Non-equalized directional microphone
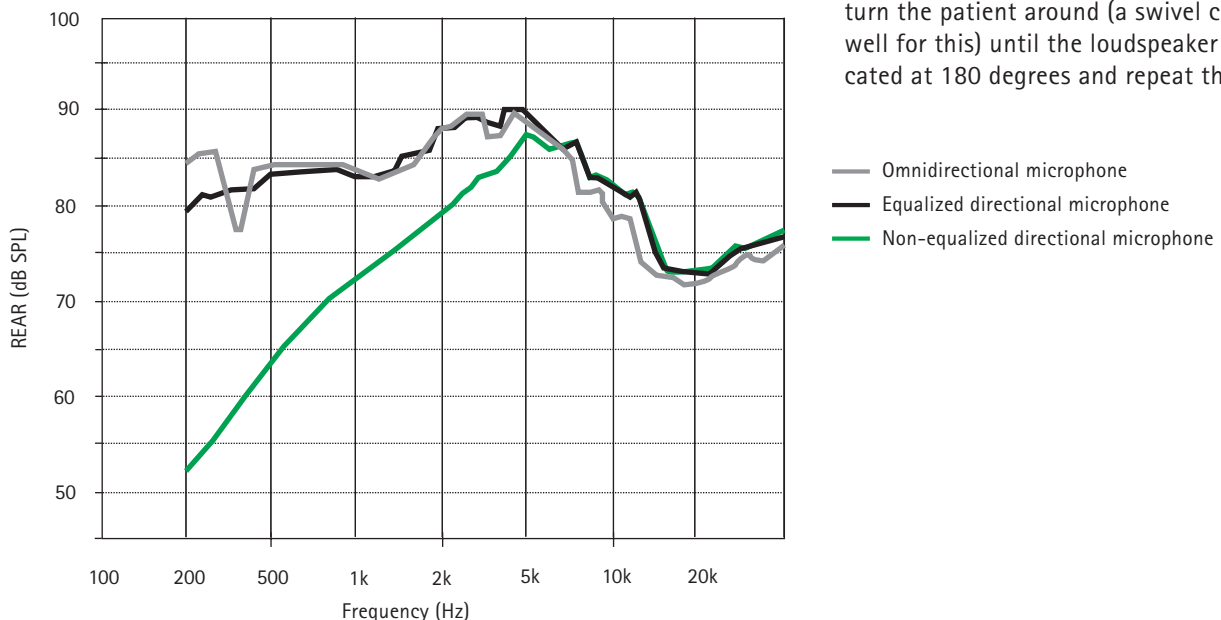
rement. The difference between these two REARs is the real-ear FBR (Figure 14). This simple procedure may be used to demonstrate the function, and also to counsel the patient regarding how to use directional microphones in different listening situations to optimize communication.

In total, these measurements will add only a few minutes to the test protocol, but it may save counseling time, as well as embarrassment if the directional microphones are not working. Speech measures, such as the R-SIN (Cox, Gray & Alexander, 2001), HINT (Nilsson, Soli, & Sullivan, 1994) or QuickSIN (Etymotic, 2001) may also be used for verification, but they are more time-consuming measures. Verification of other attributes, including occlusion and feedback, is possible, but these are often incorporated into the manufacturer's fitting software. Often, these methods employ behavioral responses from the patient, but there have been several attempts to provide an effective technique to measure the occlusion effect with probe-microphone measurements (e.g. Killion, Wilber, & Gudmundsen, 1988).

In summary, real-ear and behavioral aided measurements both provide information related to verification of hearing aid fitting goals. That said, relying exclusively on functional gain or optimal aided threshold as a method of fitting verification does not meet the two main criteria of a good assessment: validity (does it test what it is supposed to test?) and reliability (do you get the same result for repeated measures?). If the goal is to provide pediatric and adult hearing aid users with audible yet comfortable speech for a range of inputs, then optimal aided thresholds provide necessary – but not sufficient—information towards this goal. Although assessment of behavioral thresholds for soft, narrow-band sounds assures audibility, it does not guarantee that speech will be heard at those levels, and is subject to interactions with feedback cancellers, DNR systems, and multiple-band compression. Instead, the use of speech and/or speech-weighted noise in combination with real-ear measurements will provide more accurate information regarding audibility of speech and other sounds over a range of input levels. Furthermore, real-ear measurements may be used to safely and accurately assess comfort and safety of hearing aid maximum output (MPO) in relationship to an individual patient's loudness discomfort level (LDL).
As for reliability, the data stand for themselves: real-ear measurements are much more reliable than either functional gain or optimal aided threshold. Behavioral measures should be used to verify hearing, but real-ear measurements should serve as the primary means of setting hearing aid parameters.

Figure 14

Real-ear aided response measurements for real-ear FBR procedure, measured at 65 dB SPL for 0 degrees (solid) and 180 degrees azimuth.

FBR = SPL (front)
SPL (behind)

—— from front
- - - from behind
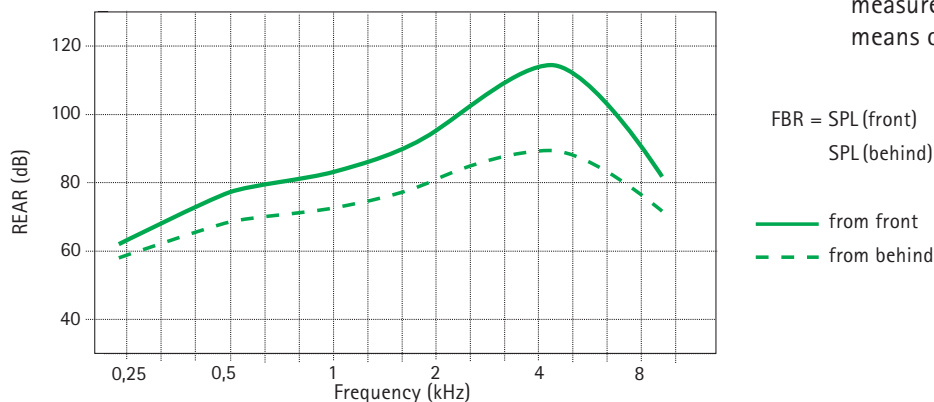
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Dave Fabry, PhD

Biography:

Dave Fabry is Director of Clinical Research for Phonak Hearing Systems in Warrenville, Illinois. Previously, he worked at Mayo Clinic in Rochester, Minnesota, from 1990-2002, and he served as Director of Audiology from 1994-2002. Dave served on the American Academy of Audiology Board from 1997-2003, and was President of the Academy from 2001-2002. He is a past editor of the American Journal of Audiology, and is a member of numerous professional associations. He lives in Rochester, Minnesota with his wife, Elizabeth, and his daughter, Loren.

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