There are several components that typically are included in a comprehensive protocol for the selection and fitting of hearing aids. Areas such as assessment, selection, fitting, verification, orientation, counseling, and real-world validation often are included. Each step plays an important role in assuring a successful audiologic treatment and rehabilitation process. As mentioned, one of these steps is verification – a common component of most protocols in medicine and technology and defined as “substantiating or determining the truth or accuracy.” The verification of hearing aid gain and output is the focus of this review article.

Establishing a Fitting Goal

To state the obvious, in order to verify something such as hearing aid performance, one must have a predetermined set of data, target data, or at the least, general fitting goals. Historically, verification methods have paralleled the fitting goals, with the goal of maximizing speech intelligibility being a primary one. It is not surprising, therefore, that there is a long history of using speech audiometry to assess hearing aid performance.

Speech Audiometry

From the 1950s through the 1970s, the primary verification tool was aided monosyllabic word recognition; a procedure used by 80–90% of audiologists (Burney 1972; Smaldino and Hoene 1981). This verification process, sometimes referred to as the comparative or “Carhart fitting approach,” usually consisted of aided sound field speech testing of two or three different hearing aids (pre-selected to have similar gain and output characteristics). Following testing, the patient was then fitted with the aid(s), which provided the highest percent correct performance. Many questioned, however, if this approach really was verification of the fitting, because of the variability associated with the test procedure (Studebaker 1982; Mueller and Grimes 1983; Schwartz and Walden 1983). For example, Mueller and Grimes, using the NU#6 in multitalker babble (S/N = 0 dB) speech material, conducted three test sessions with the same hearing aid and found mean test-retest differences of 6.7% (72 comparisons; with a mean standard deviation of 6.9%). Figure 1 illustrates...
the cumulative probability based on these 72 test-retest differences, clearly showing that large differences in test scores between hearing aids is necessary. Because audiologists usually compared similar hearing aids, the true differences between hearing aids typically were smaller than the measurement error, rendering this verification procedure meaningless.

In the late 1970s, around the same time that the reliability of speech testing as a verification procedure was being questioned, prescriptive fitting approaches were gaining favor. The notion of “selective amplification” was of course not a new concept, as it had been commonly used in the 1930s–1940s (e.g., Knudsen and Jones 1936; Watson and Knudsen 1940; Lybarger 1944). This resurgence of the use of a prescriptive fitting for adults was primarily because of the work of David Pascoe at CID (1975), Ken Berger’s prescriptive method (Berger 1976), bolstered by his many workshops, and the emergence of a new prescriptive method from Denis Byrne of Australia (Byrne and Tonisson 1976). These prescriptive methods provided fitting targets expressed in dB gain or output values – something that indeed could be verified.

Prescriptive Fitting Approaches

Most would agree that a “good” hearing aid fitting must be some combination of optimizing audibility, optimizing intelligibility, matching preferred loudness levels, avoiding loudness discomfort, and providing good sound quality. These goals have led to the development of several prescriptive methods over the past several years. Some have been named after people (e.g., Bragg, Libby, Lybarger, Shapiro, Victoreen) and others named after places (e.g., Cambridge, Central Institute for the Deaf [CID], Memphis State University [MSU], National Acoustic Laboratories [NAL]). The name of a few described what the method does (e.g., Desired Sensation Level [DSL], Prescription Of Gain and Output [POGO], Visual Input/Output Locator Algorithm [VIOLA]) and one method simply is named after a figure (FIG6). The evolution and development considerations related to many of these methods have been reviewed by Humes (1996) and Byrne (1996). Today, in the U.S., two methods have emerged as the defaults for clinical use: the NAL, with the current version being the NL2, (see Chapter 10 in this proceeding) and the DSL, with the current version being the 5.0 (Scollie et al. 2005; see Chapter 9 in this proceeding).

In general, prescriptive methods are based on the assumption that using the patient’s pure-tone thresholds, and perhaps some supra-threshold measures, hearing aid gain and/or output can be prescribed that will result in the best fitting for the average patient when listening to average input signals, in average listening situations, for the average desired attributes (e.g., intelligibility, sound quality, listening comfort, etc.). There are many variables to consider, and no one procedure is going to “get it perfect” for all patients all the time. But we need a starting point, and research studies have shown that chances for success will be maximized when we use validated procedures such as the NAL and DSL (Mueller 2005a; Scollie et al. 2005; Byrne, Dillon, Ching, Katsch and Keidser et al. 2001).

Verification of Prescriptive Targets

Clearly, we have at least two fitting procedures that provide us with validated prescriptive targets. The clinician’s task, therefore, is then to verify that these targets have been met at the time of the fitting of the hearing aids. The only reasonable method to accomplish this is to use probe-microphone measures (see pro and con discussion on the topic by Dillon and Keidser 2003). For reasons not fully understood, however, most audiologists choose not to use this method. Surveys conducted over the past several years suggest that for adult hearing aid fittings, the routine use of probe-microphone measures for verification in the U.S. is probably no higher than 30–35% (Mueller 2003, 2005b), and there are some data to suggest that the actual use rate is even lower than typical survey results suggests (Mueller 1998). Given this low acceptance rate, it’s obvious that audiologists must be using alternative methods to assure that ear canal SPL is appropriate for a given patient. Let’s review these alternative methods, and discuss why they are not appropriate.

Manufacturers Automated Prescription

It is common for manufactures to have a “preferred fitting algorithm” for their products, and implement this in the fitting software. This feature facilitates the fitting process, as with a single mouse click, numerous parameters are automatically adjusted across many channels. It’s a handy feature, which streamlines the adjustment process. The potential downside, however, is that the resulting gain and output settings may be quite dissimilar.
to the values prescribed by validated prescriptive methods. Byrne (1996) recognized this problem several years ago, and expressed his concerns when he stated: “Scientifically, the concern is that amplification may become prescribed by a wide variety of proprietary formulae of which few, if any, are validated by published research. A possible philosophical problem is that control of the fitting process is taken away from the fitter, who is responsible for the care of the client” (page 378).

Research has shown that the manufacturers’ preferred fitting can be quite different than that of validated methods. Keidser, Brew and Peck (2003), for example, in a study examining the recommended algorithms of five different major manufacturers, showed that it is common for prescribed gain to differ by 10 dB or more from the NAL-NL1 targets in the high frequencies for average-level input signals. Bentler (2004) examined the default algorithm of the premier product from six leading hearing aid manufacturers using a real-speech input (long-term 65 dB SPL input). In general, all algorithms prescribed gain below the NAL-NL1 target levels. Of particular concern was that for key frequencies such as 2000 Hz, the difference in prescribed gain was as much as 15–20 dB.

Simulated Real-Ear Gain

Another popular alternative to the real ear verification of hearing aid performance is to use the “simulated gain” that most manufacturers provide in the fitting software. In some cases, this simulated gain is paired with the targets of a validated prescriptive method, and it’s possible to adjust the frequency response to obtain a match. It is tempting to assume that what is happening on the fitting screen is what is happening in the real ear. Unfortunately, this probably is not true. While we would not expect a perfect match on an individual basis due to individual ear differences (referred to as CORFIGs; Coupler Response for Flat Insertion Gain), Hawkins and Cook (2003) reported that there also is not a match when average data are used. Their research showed a distinct pattern of deviation; an average real-ear reduction of high frequency gain of 5–10 dB. That is, even if an audiologist selected a validated procedure, such as the NAL-NL1, and the computer display showed an excellent match to target, it is possible if not probable that the real-ear gain above 2000 Hz will fall well below this prescriptive target. Similar findings were obtained by Aarts and Caffee (2005), although their work was limited to a single manufacturer.

2-cc Coupler Measures

A final alternative method would be to fit the hearing aid to a prescriptive target using 2-cc coupler measures. While this approach is probably better than the methods we have discussed so far, even when coupler values are matched to target (e.g., measured and adjusted by the audiologist), it is difficult to predict real-ear gain from coupler gain on an individual basis. This is due to the inter-patient variability of the real-ear unaided gain (REUG) (i.e., a standard deviation of 4-5 dB in the higher frequencies; Valente, Valente and Goebel 1991), the real-ear coupler difference (RECD) (i.e., a standard deviation of 4.5 dB in the higher frequencies; Saunders and Morgan 2003) and the microphone location effects (MLE) (i.e., a standard deviation of 2 dB in the higher frequencies; Fikret-Pasa and Revit 1992). This variability, of course, could be reduced by measuring the patient’s RECD, as is often done with infants and children. But if one were to go to this trouble with adult patients, then why not simply conduct probe-mic verification?

Probe-Microphone Measures

The equipment for conducting probe-microphone measures, as we know it today, has been commercially available since the early 1980s. Other than the DSL, commonly used prescriptive fitting methods in the 1980s had gain-based targets, and audiologists were using “functional gain” measures to verify if appropriate gain was present. Probe-microphone testing was a welcome addition, as it could be used as a replacement for functional gain, and brought with it a greater degree of flexibility, accuracy and reliability. Because “gain” values were what was needed to verify the prescriptive methods that were then used, REIG (then called REIR) calculations were the clinic verification standard.

With the 1990s came the increased use of wide dynamic range compression (WDRC) hearing instruments. It became apparent that verification needed to include more than just one input intensity to determine if these instruments were functioning as desired. The concept of a “family” of verification REIGs was introduced.

Routine Verification Protocol

Today, whether the fitting targets are REIG or REAR based, it is recommended to use at least three input intensities when hearing aid verification is conducted. This relates to the fitting concept of making soft sounds
audible, average sounds comfortable and loud sounds loud but not too loud. The exact input-signal values used from clinic to clinic depends somewhat on what type of input signal is used, what a given clinician believes is the “right” values for soft, average and loud, and from a convenience standpoint, what targets are automatically displayed on the probe-mic fitting screen. The three most commonly used input levels are 50, 65 and 80 dB SPL. If gain and compression is set to match these targets, it is probable that there will be a close match for other input levels too. In most cases, the fitting targets are displayed on the screen of the probe-mic equipment. If not, then the targets need to be derived separately (e.g., using the NAL-NL1 or DSL stand alone software), and some back-and-forth visual checking is necessary.

The type of input signal used also is an important consideration. At one time it was common to use swept tones for probe-mic assessment, but today, because nearly all products use multiple channels, compression, digital noise reduction (DNR), and other overlapping algorithms, broad-band input signals are preferred. Many of these signals have been designed to have modulations similar to speech, so that testing can be conducted with DNR active. Of course, another option, available on most equipment, is to use real speech as the input signal (more on that later).

Scollie and Seewald (2002), Henning and Bentler (2005) and others have shown that because of the interactions with compression and multiple channels, hearing aid output can vary significantly based on the input signal used. This of course would then influence the verification process and tweaking of the hearing aid at the time of the fitting. In addition to the spectral content, the shape of the available broad-band signals also may be quite different, which again easily could influence the match to target. Moreover, some systems allow for REIG calculations to be conducted using a different input signal for the REAR than was used for the REUR measurement, certainly a potential source for REIG error. The bottom line is that care should be taken to use an input signal that is broad-band, but that also is most similar to the signal that was used to develop the prescriptive method that is being verified. Shaped real-speech would seem to be the most logical choice.

**Verification of Maximum Output**

In addition to the verification of fitting targets for the inputs of 50 to 80 dB SPL, it also is possible to use probe-mic measures to determine if the hearing aid’s maximum output is set correctly. This procedure is referred to as the real ear saturation response (RESR) and usually is conducted using a swept tone at an input level of 85 or 90 dB SPL. Targets for this verification are derived from measuring the patient’s loudness discomfort level at two or three key frequencies (using a pure-tone or narrow-band input signal), and then converting these dB HL values to ear-canal SPL. There is evidence that this verification procedure is related to real world satisfaction with amplification (Mueller and Bentler 2005). Because of factors such as loudness summation, binaural summation and channel summation, however, this probe-mic verification should be supplemented with aided behavioral judgments of loudness discomfort (see Mueller and Hornsby 2002, for review).

**Verification of Special Features**

It also is important to include the testing of special features as part of the probe-mic verification protocol. Many products are “sold” based on the special features that they possess, and it is necessary to determine if they are working appropriately. DNR algorithms and directional microphone technology are two such features.

It is well documented that DNR algorithms vary significantly from manufacturer to manufacturer, and sometimes even among products within a manufacturer (Bentler and Choiu 2006). The magnitude of the DNR for different frequency regions also can be altered by hearing aid gain, venting, and other factors that are unique to a given patient.

It is possible that some DNR algorithms alter gain for speech (compared to the DNR-Off condition), so a starting point in the verification process may be to conduct testing using a real speech (or speech-modulated noise) signal with DNR “On” versus “Off.” To observe the magnitude of the DNR, and obtain an understanding of the on-set and off-set times, a broad-band input signal then should be chosen that does not have speech-like modulations (i.e., a signal that clearly will be judged as noise by the signal classification system of the hearing aid). Other factors to consider when conducting this testing:

- Some DNR algorithms have a long on-set time, so the input signal should be left on for an extended period (e.g., 30 seconds or so).
- Some DNR algorithms are level-dependant, so different input levels should be used (e.g., 60, 70 and 80 dB) to see magnitude effects.
• With some hearing aids, WDRC, depending on the setting of the compression ratio, will reduce the effects of the DNR, so testing should be conducted with compression adjusted to patient-use settings.

Another feature to assess during the verification process is directional technology. A simple way to do this is to derive a real-ear front-to-back difference in output for inputs from a 0 degree versus 180 degree azimuth, a standard clinical procedure which has been used for several years (Mueller and Bryant 1991). Step-by-step protocols for conducting this measure are available (Mueller 2001a; Ricketts 2001).

There are several reasons why verification of directional technology is important. First, some products simply may not be functioning properly. While this is not common with new instruments, it does happen. On follow-up verification testing, reduced directivity becomes more common, because of dirt and debris in one or both of the microphone port openings. Directionality also can be affected by venting and hearing aid gain. Additionally, directionality will be affected by the positioning of the instrument on the real ear, which changes port alignment and the external timing (Ricketts 2000) – all reasons to conduct real-ear verification. It is important to remember that this testing is conducted as a quality control measure – to determine if the directional technology is functioning properly. It is not intended as a method to compare one product to another (Mueller 2001b), or to predict speech-in-noise intelligibility benefit (Dhar, Humes, Calandruccio, Barlow and Hipsking 2004).

In the past few years it has become common for hearing aids to have automatic switching to and from the directional mode, and to have adaptive polar patterns. If front-to-back probe-mic testing is conducted in the automatic/adaptive setting, a few procedural issues need to be considered:

• The input signal must be loud enough to “trigger” the switch to directional. This varies from model to model, but usually is around 55–60 dB SPL (the trigger intensity can be quickly determined by conducting repeated runs from the back, starting ~45 dB SPL and going up in 5 dB steps).

• For many products, the shift to directional only occurs if the input is classified as noise. It is recommended, therefore, to use an input signal that the hearing aid believes is “noise-like.”

A final point to consider is that WDRC will reduce the front-to-back difference – the larger the compression ratio, the more the reduction. Importantly, this is an artifact of the measurement procedure (front and back signals not presented simultaneously). Directivity in the real world (diffuse noise setting) is better represented with testing in the linear mode. For more information concerning why this is true, see Mueller (2001c) and Ricketts (2000). If the reason for conducting the testing, however, is to simply determine that the hearing aid is working properly, then it is not critical to know the “true” front-to-back difference.

Verification Using Output Versus Gain

As mentioned earlier, historically, the prescriptive targets verified with probe-mic measures used gain values, and hence, the REIG calculations typically have been used for verification. In addition to history, there are several reasons why audiologists prefer to think in terms of “gain” when they fit hearing aids. Hearing aids and fitting ranges are typically classified by gain. Fitting software usually uses gain simulations. Ever since Lybarger’s 1⁄2 gain rule, Libby’s 1⁄3 gain rule, and all rules in between, it has been pretty easy to quickly estimate how much gain a person needs simply based on their pure-tone thresholds. Not so easy to quickly estimate desired ear-canal SPL. Moreover, for most, the “goodness” of a fit is judged much more easily by observing a REIG curve rather than an REAR curve. Finally, the standard audiogram and our casual counseling techniques encourage the use of gain values: “We can move your hearing loss from down there to up here, and all will be better.”

One measurement advantage of the REIG, mentioned earlier, is that the input signal used is not as critical; assuming that the same signal is used for the REUR and the REAR measurements. That is not to say that “input signal does not matter”, as there still are interactions with WDRC affecting the REAR, but not the REUR, which could result in a smaller or larger REIG for one input signal versus another.
Another factor when using REIG verification, which can be judged as a positive or a negative depending on who you ask, is the use of individual REUG values. Anyone who has conducted probe-mic measures knows that an unusually shaped REUG can cause a “nice looking REAR” to become a rather “ugly REIG”! The question then becomes, do you adjust the instrument, altering what was a good REAR, so that an acceptable REIG can be obtained? If your answer is “yes” then you may not like using the REAR for verification, as this approach more or less assumes that the REUR is average.

In recent years, there has been a general trend to move away from the REIG, and shift to using the REAR values for verification purposes. This is not a new concept, as it has long been the standard for verification of the DSL, and was recommended as a verification method for adults back in the early 1990s (e.g., Mueller 1992; Valente, Potts, Valente, Vass and Goebel 1994). Why this change is happening now, rather than 15 years ago is not clear, but is probably related to some combination of the teachings of the DSL group, the availability of REAR targets with NAL-NL1, and the displays provided by the leading manufacturers of probe-mic equipment.

Verifying hearing aid performance in an “ear-canal SPL world” has several advantages, in addition to the fact that “soft” finally falls below “loud.” Most of what we know about hearing aids, speech spectrums, listening conditions and the acoustics of the world is in SPL, not HL. We discussed earlier that a common fitting goal is to package the average speech signal within the patient’s residual dynamic range. Success or failure regarding this task is observed most effectively using the REAR rather than the REIG, both from an audibility and a maximum output standpoint.

When using the REAR rather than the REIG for verification of prescriptive targets, more attention does need to be given in selecting the input signal. Most recently, this concern relates to the use of real speech inputs.

**Real Speech Inputs**

Given that the primary reason individuals purchase hearing aids is to hear and understand speech, speech signals always have been the logical input signal to use for REAR verification. With the equipment of the 1980s, it often was necessary to physically disable the loudspeaker to “trick” the system into measuring aided speech signals – since the system itself had no speech signal, the speech input usually came from a hand-held cassette recorder. Today, much has changed as nearly all manufacturers have a convenient method and meaningful display for using real-speech inputs with their equipment, often with several different speech samples available. In general, this feature often referred to as “speech mapping” has generated considerable enthusiasm among both clinicians and researchers (Moore 2006). In fact, the move toward using real speech is yet another reason why the REAR is replacing the REIG as a verification tool.

Aside from the face validity of the procedure, there are many reasons why real speech is the preferred input. Many features and algorithms of today’s hearing aids react differently to a speech signal than they would for a broad-band noise signal, and most definitely, swept tones. Speech inputs provide a real-ear real-world view of the functioning of such features as overlapping channels, compression, expansion, DNR, and adaptive feedback reduction. In many cases, this expanded analysis alters the hearing aid adjustments that are needed when verification is conducted. An example of this is shown in figure 2, which displays two REARs using a real speech input for the same hearing aid. For the two measures, all settings were the same except the compression time constants (“short” versus “long”; ratio ~2:1). Note that the amplitude of the two amplified speech signals is different, and that the average level (dark line near center of spectrum) is about 5–8 dB higher for the one setting (long time constants) than for the other. Although targets are not displayed, it is clear that the compression...
time constants selected could influence gain adjustments needed to match target.

When using real speech it is recommended to conduct a family of curves using different input levels, as typically has been done over the years with other input signals. The advantage now is that the effects of compression and possibly expansion become more obvious. As both Harvey Dillon and Susan Scollie stated in a recent review article (Mueller 2005b), real speech easily can be used to verify both the NAL and DSL fitting algorithm. The caveat is that the speech signal used must be consistent with the signals used to establish the prescriptive targets (and of course, the task is facilitated if the targets are displayed on the fitting screen).

While it is difficult to think of a “downside” of using real speech, there is one concern regarding the fit to target. As mentioned earlier, it is common to convert the patient’s HL threshold to earcanal SPL and display this on the fitting screen when speech mapping is conducted. Some people use this as “the fitting target.” That is, the goal is to place the lower boundary of average speech, or the mid-point of soft speech at or above this line. In a sense, this then becomes a new, non-validated prescriptive fitting approach, which could differ significantly from the NAL and DSL, especially in the high frequencies when the patient has a severe downward sloping hearing loss. An example is shown in figure 3, which displays the amplified speech signal (65 dB input) superimposed on the NAL-NL1 targets (the upward sloping line immediately below the speech display is the patient’s thresholds).

In this example, the “above the threshold fitting method” would suggest that gain in the 4000 Hz region be increased a little; however, note that the average of the speech (dark line near middle) already is above the NAL target for this frequency. This patient only had a 60 dB loss at 4000 Hz – the difference between methods will become greater as the loss becomes worse. While on the surface, audibility in the high frequencies seems like a good thing, most research suggests that this approach would result in too much high frequency gain (note: the rules for children are different). The message is, it is probably best to stick with the validated methods, even when speech mapping is conducted.

Live Speech Inputs

A popular variation of using real speech signals is to use live speech. That is, the examiner, a family member, or other interested party reads a passage or simply talks to the patient. While this may prove to be interesting, and maybe even entertaining for the patient, this approach does not have the preciseness required to verify established fitting targets. For example, shown in figure 4 are the live voice amplified speech signals from a male and female for the same hearing aid settings. Observe that the output is fairly similar up to 1500 Hz, but in the 3000 Hz...
region, there is as much as a 15 dB difference. Obviously, this hearing aid would be programmed quite differently depending on whose live voice was used. Which one is right? And are either one “average?” Would the voice even be the same on retest? There certainly are advantages of using live voice for demonstrations and counseling, but it seems prudent to use a consistent calibrated real speech signal when target verification is conducted.

Open Canal Fittings

In recent years, open canal (OC) fittings have become increasingly popular. This style is characterized by a small BTE housing, a thin tube or wire leading to the ear canal, and a loose fitting in the ear canal that leaves the canal mostly open – that is, the REOR is similar to the REUR (Mueller 2006). Because the ear canal remains open, there has been considerable discussion in several different areas regarding the use of probe-mic measures with this type of fitting. Fortunately, as described by Mueller and Ricketts (2006), most of the “issues” are really “non-issues”. I will briefly mention the key considerations that have been raised related to probe-mic measures.

Prescriptive Targets

As described earlier, the first step in the verification process is to establish fitting targets or goals. It sometimes has been mentioned that commonly used prescriptive fitting targets (e.g., NAL-NL1, DSL 5.0) cannot be used with OC fittings. This belief may have been encouraged by the following statement from the gain verification section of the recent American Academy of Audiology Hearing Aid Fitting Guidelines:

“...some prescriptive formulas for open fittings may be inappropriate as there is no need to correct for the insertion loss created by including an earmold or hearing aid shell in the fitting process.”

This is a curious statement, as neither the REIG nor the REAR fitting targets correct for insertion loss. At this point, there would seem to be little reason to abandon the validated fitting methods that have proved to be successful over many years, simply because the ear canal is open. As expressed by Dillon (2006), there is no reason why changing the size of the venting should change the ear canal SPL that is optimal for a given individual.

If you consider the popular alternative to using a prescriptive fitting approach – using the manufacturers’ default fitting – there is even more compelling reasons to continue to use the validated methods. Figure 5, taken from Bentler, Wu and Jeon (2006) show the resulting REIG for four different OC products based on the manufacturers’ recommended fitting. The target shown is the NAL-NL1 (65 dB SPL input) for an individual with a high frequency hearing loss (50 dB at 2000 Hz, 60 dB at 3000–6000 Hz). Observe that although there are some differences among manufacturers, the average “recommended” gain falls 10–15 dB below NAL-NL1 targets in the 2000–4000 range.

![Figure 5. Sample REIGs obtained from four different open canal fittings. NAL-NL1 fitting target shown for reference (From Bentler, Wu and Jeon 2006).](image)

REIG versus REAR

As discussed earlier, there has been a general shift to using the REAR rather than the REIG for verification in recent years. Beyond this trend, there has been some discussion that the REIG is not a valid measure for OC fittings. This could be because intuitively, it doesn’t seem quite right to subtract the REUG from the REAG for the REIG calculation, when in fact the patient retains the majority of this open ear advantage (the REUG). As described by Mueller and Ricketts (2006), however, the math does work, and OC verification can be conducted equally well using prescriptive targets for either the REIG or the REAR – a fit-to-target via REIG usually will be a fit-to-target with REAR and vice versa. It is worth mentioning that some OC products have little gain above 2000 Hz, which is much more obvious when displayed in the REIG mode rather than the REAR mode, which may have influenced some of the REAR proponents.
Summing and Cancellation

A third issue relating to OC fittings and probe-mic verification is the summing and cancellation effects that might occur. With the OC fitting, sound enters the ear canal through the direct pathway, which is then mixed with the amplified sound. When these two inputs (from the same source) are fairly equal in amplitude, summation and cancellation can occur across frequencies, depending on the phase relationship at the tympanic membrane. When probe-mic measures are conducted, these bumps and notches may be observed, depending on the sampling rate and smoothing of the system used. If they are smoothed, it may simply appear as slightly more gain where summation is present. As reviewed by Mueller and Ricketts (2006), these effects are not of major concern, as they only occur when the output of the direct path and the hearing aid path are similar. When hearing aid gain is present, the effects will be minimal, as the hearing aid output will be significantly higher than the direct signal. This is illustrated in figure 6, which is data collected at the University of Iowa by Yu-Hsiang Wu. Observe that the only area where summing occurs is in the 1000 Hz region, and even there it is minimal. For the region where target gain is critical for an OC fitting, 1500 to 6000 Hz there is no summation, as the amplified signal is significantly greater than the direct signal.

Equalization of the Input Signal

This final issue, equalization (calibration) of the input signal, is indeed something that needs to be considered when probe-mic assessment of OC fittings is conducted. When probe-mic testing employs an active reference microphone in the vicinity of the ear being tested, as it usually does, it is termed the “modified pressure method.” As defined in ANSI Standard S3.46–1997 (American National Standards Institute 1997), a variation of the modified pressure method, termed the “modified pressure method with concurrent equalization,” measures and adjusts the sound field in real time as the probe-mic measurement is being made. Another calibration option is to base the sound field equalization on data obtained from a prior measurement. The ANSI standard refers to this as the “modified pressure method with stored equalization.” Importantly, the stored equalization method is not the same as the “substitution method,” in which the field is equalized with the patient absent.

With OC fittings, there is a considerable “outflow” of amplified sound from the open ear canal, which is received by the reference microphone. It is possible that when concurrent equalization is used, the intensity of the signal leaking out of the ear exceeds the signal delivered from the loudspeaker. This will cause a decrease in the input signal, which in turn will reduce the measured ear canal SPL (see Mueller and Ricketts 2006, for review). The magnitude of the “mistake” will depend on factors such as the gain of the instrument, the feedback algorithm of the instrument, and the location of the reference microphone. For a typical fitting with gain of 25 dB in the high frequencies, the mistake appears to be around 5 dB (e.g., measured output [gain] is 5 dB less than true output).

The solution is to use the stored equalization method; an option available on all commonly used probe-mic equipment. An example is shown in figure 7, derived from Yu-Hsiang Wu. Observe that the only area where summing occurs is in the 1000 Hz region, and even there it is minimal. For the region where target gain is critical for an OC fitting, 1500 to 6000 Hz there is no summation, as the amplified signal is significantly greater than the direct signal.
real speech signal presented at 65 dB SPL. The spectrum with the higher amplitude was measured using the stored equalization. Observe that if the typical concurrent equalization had been used, gain would have been underestimated by 5–7 dB in the 2000 Hz region.

In Closing

It was back in 1990 when Mike Valente asked me to write an article on probe microphone measures for a special issue of Seminars in Hearing that he was editing. At that time, probe-mic measures had been around for six years or so, and it seemed that just about everything that needed saying already had been said. To find a niche area, I titled my article “Some commonly overlooked uses of probe-microphone measures.” I would have never guessed then, that in 2007 I would be writing that the overlooked use is routine verification! But it certainly is.

For some reason, when it comes to probe-mic verification of hearing aid performance, there seems to be a “can’t do” philosophy. In the early 1990s, we were hearing that “you can’t do probe mic with WDRC,” and then at the turn of the century we were hearing “you can’t do probe mic with digital,” and now only last year we were hearing, “you can’t do probe mic with OC fittings.” All these notions are false, of course, and in fact, the more sophisticated or different the fitting, the greater the necessity for a good verification tool.

This brief report has reviewed several ways in which probe-mic measurers can be used in the verification process, and addressed a few new procedural issues to consider. Sure, the use of real speech has gotten a few people excited, and the evaluation of OC instruments and digital algorithms is a little different and interesting. But the underlying concept has not changed much since we all test drove the new Rastronics CCI-10 at the 1984 ASHA convention. We have substantial research that tells us what earcanal SPL provides a given patient the best chance of deriving benefit from and satisfaction with hearing aids. It is our job to assure that these earcanal numbers are correct, and without probe-microphone verification, it becomes a very haphazard process.

References


