Pure-tone signals, pure-tone sweeps, speech, and various composite signals are all being used in the verification of hearing aids. However, each test stimulus can yield significantly different results depending on the types of technologies being tested. Additionally, some test stimuli are not applicable for verification of certain technologies/situations, and there is a need for “apples-to-apples” comparisons among clinicians, hearing scientists, and manufacturers. This article provides a primer and basic background on the concepts and use of pure-tones, speech, and ANSI and ICRA composite signals during hearing aid verification.

Acoustic Stimuli Used in Verification

There are many reasons for verifying the performance of a hearing aid. Some dispensing professionals verify in order to ensure that the chosen device matches the manufacturer’s specifications. This is frequently done using standardized hearing aid couplers in order to ensure the integrity of the hearing aid and its components. Other dispensing professionals verify in order to ensure that the real-ear gain/output of the device matches the desired recommendations of a prescriptive target. This task implicitly assumes that a certain level of real-world satisfaction may be expected when the target output/gain is met.

Pure-tone sweeps (or sinusoids) have been used as the stimulus to verify the performance of hearing aids tested with couplers and worn on real-ears for many years. In recent years, however, more acoustic stimuli have become available as possible alternatives to pure-tones for verifying the performance of hearing aids. While some may expect similar outcomes from the use of different stimuli in verification, the reality is that the choice of a particular stimulus can lead to outcomes that may alter one’s interpretation of the results of verification. This article illustrates some of the effects of using different signals to verify today’s hearing aids.

Pure-tones: Pure-tones are “simple” signals with energy at only one frequency. Pure-tone sweeps (or sinusoids) are pure-tones with continuously changing frequencies presented at the same fixed level over the range of audible frequencies in a fixed amount of time. Pure-tones and pure-tone sweeps are simple because their frequency and intensity levels can be precisely controlled by the test equipment and no special frequency analyzers are needed for their measurement. The signals have been used to estimate the frequency-gain/output characteristics of hearing aids. In addition, they have also been used to determine the input-output characteristics, saturation distortion, attack/release times, as well as the current drain of hearing aids. If the purpose for verification is to determine the integrity of the structural components in a hearing aid, this “simple” stimulus can serve its purpose quite well. Indeed, the most recent ANSI standard (ANSI S3.22-1996) recommends the use of sinusoids in testing hearing aid performance for quality assurance.

On the other hand, since only one frequency is presented at a time, any interaction among frequencies that are likely to occur in nonlinear signal processing cannot be examined using sinusoids. Furthermore, since the validity of verification is limited to the specific stimulus that is used, one cannot generalize the performance of nonlinear hearing aids to complex sounds using results obtained from sinusoids. Thus, pure-tones are not suitable if the purpose of verification is to examine the performance of the
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Choice of Stimuli for Hearing Aid Verification

These are called composite signals—the most appropriate signal for evaluating today's (and future) nonlinear hearing aids. With careful interpretation, one can use the results obtained from composite signals to predict the performance of the hearing aids to real-life complex signals.

There are many different speech spectrum shaped composite signals—each synthesized with different criteria so they are similar, but not identical, to "real" speech. In one commercial speech spectrum shaped composite signal modeled after the ANSI S3.42 recommendation, the composite speech is synthesized to have frequency components from 100 Hz-8000 Hz spaced in 100 Hz intervals. The amplitudes of these individual frequency components are controlled to achieve similar long-term characteristics as speech recorded directly in front of the talker. This has a slope of -6 dB/octave.

Another speech spectrum shaped composite signal is the ICRA signal (International Collegium of Rehabilitative Audiology) that purports to approximate the ideal long-term far-field speech spectrum which has a slope of -9 dB/octave. A reason for using the ICRA signal is that many of the recordings are modulated at rates typical to that of conversational speech. This may be useful when examining nonlinear hearing aids with "noise reduction" algorithms.

Another example is a dynamic stimulus where sweeping sinusoids over the speech range are modulated to resemble the dynamic range of typical speech. This type of composite signal may be useful for analyzing the dynamic response of a hearing aid to speech-like signals.

There are many more speech spectrum shaped composite signals available commercially, each with its own unique design and spectral/temporal characteristics. Because the purpose of this paper is to raise awareness of the potential problems with stimulus differences and not to provide an exhaustive demonstration of the different stimuli, we will focus on the responses of hearing aids to only three signals: 1) the pure-tone sweep (Tone); 2) the continuous speech spectrum shaped composite signal according to ANSI (ANSI), and 3) the modulated speech spectrum shaped composite signal according to ICRA (ICRA).

Figure 1 shows the pure-tone sweep at 50 dB SPL and the spectra of the ANSI and ICRA signals at an overall level of 70 dB SPL that are used in all the following demonstrations. These signals are generated from the Frye 6500 Hearing Aid Test system, and all subsequent output/gain measurements are performed with a default bandwidth of 100 Hz. Clearly, the ICRA signal has more low-frequency energy and less high frequency energy than the ANSI signal.

Response of Linear Hearing Aids to Different Stimuli

At any particular frequency, a linear hearing aid provides the same gain at all input levels for that frequency. A direct consequence is that the measured gain of a

![Figure 1](image1.png)

**Figure 1.** Frequency spectra of the stimuli (measured in a 100 Hz bandwidth) used in the demonstration; pure-tone sweep at 50 dB SPL (green), unmodulated ANSI signal (blue), and modulated ICRA signal (black).

![Figure 2A-B](image2.png)

**Figure 2A-B.** Top (2a): Gain curves of a linear hearing aid (Logo) tested with a pure-tone sweep (at 50 dB SPL), a composite-speech noise (70 dB SPL overall), and the ICRA noise (70 dB SPL). Bottom (2b) : Output curves of the same linear hearing aid tested with the same signals.
linear hearing aid should be identical regardless of the stimuli used for the measurement. Figure 2a shows the frequency-gain curves of a linear hearing aid to the ANSI signal at an overall level of 70 dB SPL, the ICRA signal at the same overall level, and a pure-tone sweep at 50 dB SPL. Note that the frequency-gain curves for all three signals are identical despite the differences in their input spectra.

In other words, gain on a linear hearing aid is insensitive to stimulus level difference (since gain is independent of input level). A corollary is that any stimulus may be used to measure gain of a linear hearing aid. This unique property of linear hearing aids may have helped the use of linear prescriptive formulae where most specify target gain based on the degree of hearing loss.

While gain of a linear hearing aid is insensitive to stimulus level difference, output from a linear hearing aid is sensitive to level differences among stimuli. This is logical since output is the sum of the input level and gain. Figure 2b shows the output of the linear hearing aid to all three stimuli.

It is fair to say that the linear hearing aid yields different output for the different stimuli despite the same gain setting.

The use of stimuli with different spectral characteristics could be problematic during verification when one tries to match the output of the hearing aids (linear and nonlinear) to a prescriptive target. This is because the recommended output level at a certain frequency offered by a prescriptive formula is not only determined by the degree of hearing loss, but also on the intended input signal and on the bandwidth of the measuring device (or hearing aid) within which the level is specified.

Ideally, one should use the same stimulus during verification as that used in the target formulation in order to legitimize the match. The use of a different signal would violate this “apples-to-apples” assumption and make the appropriateness of the matched output questionable. This is demonstrated in Figure 2b where the same hearing aid setting yields different frequency gain curves depending on the stimuli used.

(Authors’ note: To be most accurate, fitting targets recommending a desired output (viz. not gain) should also recommend the bandwidth of the measurement system (eg, 1 Hz, 100 Hz, 1/3 octave, etc) that one can use to verify the output of the hearing aids. This is because different measuring bandwidths would yield different frequency-gain curves with a complex signal).

This discrepancy had not raised concern until recently because many linear and nonlinear prescriptive formulae specify gain targets rather than output targets; sinusoids rather than complex signals are used as the stimuli for verification in these cases. On the other hand, if we consider that human ears respond to output rather than gain, it is logical to include prescriptive targets based on the output of hearing aids. Furthermore, a standardized signal (eg, ANSI or ICRA signal) and a standardized measurement bandwidth (eg, 1/3 octave) must be specified to minimize variability in outcome. This will become more important when nonlinear hearing aids are concerned.

Thankfully, some of these concerns are beginning to be widely addressed today.

Nonlinear Hearing Aid Responses

Effect of multi-channel nonlinearity: The advent of nonlinear hearing aids brings new considerations to the choice of acoustic stimulus for verification. A nonlinear hearing aid provides different gain at different input levels. Typically, more gain is provided to low-input levels, and less gain is provided as the input level increases (except when the hearing aid is in nonlinear expansion).

The effect of nonlinear gain is especially seen in a multi-channel device which has independent gain control at each frequency channel. It suggests the possibility that more variation in the frequency-gain (and frequency-output) curves may be possible when different stimuli are used to test the hearing aid. This is shown in Figure 3a and 3b where a 15-channel nonlinear hearing aid is tested with the same stimuli used previously (a pure-tone sweep, the ANSI signal, and the ICRA signal). Figure 3a shows that more high-frequency gain is reported with the ICRA signal than the ANSI signal, possibly because the ICRA signal has less high frequency energy than the ANSI signal (thus more gain because of the lower input). This is in contrast to the linear hearing aid where all three stimuli show the same frequency-gain curve. As expected, the frequency-output curves (Figure 3b) show significant output differences among the three stimuli used to determine the output responses.

Effect of special signal processing algorithms: Many nonlinear digital hearing aids also employ special “noise reduction” algorithms that seek to identify the nature of the incoming signals to make additional gain adjustment beyond that provided by compression. For example, the Senso Diva hearing aid analyzes the level distribution of the input signal to make such a determination in each of its 15 frequency channels. Analysis that shows a bi- or multi-modal level distribution would suggest “speech” in that channel. A uni-modal distribution, which indicates that the signal intensity is relatively constant over time, would suggest a “noise” (or

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FIGURE 3A-B. Top (3a): Frequency gain curves of a 15-channel hearing aid (Diva) tested with a pure-tone sweep (at 50 dB SPL spectral level), a speech-composite noise (70 dB SPL overall), and the ICRA noise (70 dB SPL). Bottom (3b): Frequency-output curves of the same hearing aid tested under identical conditions. The noise reduction algorithm on the aid has been deactivated.
noise reduction algorithm on the aid is active and the hearing aid tested under identical conditions. The Bottom (4b): Frequency-output curves of the same dB SPL overall), and the ICRA noise (70 dB SPL). 50 dB SPL spectral level), a speech-composite noise (70 dB SPL) channel hearing aid tested with a pure-tone sweep (at 10 seconds for the algorithm to confirm the identity of the incoming signal. In general, the more channels there are and the higher compression ratio (CR) in each channel, the more noticeable this summation effect. This is seen in Figure 5a where four hearing aids—the 15-channel Senso Diva, the 3-channel Senso C+, the 2-channel Bravo, and the single-channel Logo—are matched in output to a NAL-R target using a pure-tone sweep presented at 60 dB SPL. Figure 5b shows the output of the same four hearing aids to the ANSI signal that has an overall level of 60 dB SPL (noise reduction deactivated). The output curve of the 15-channel Senso Diva is almost 10 dB higher than that of the single-channel Logo, and over 5 dB higher than those of the 2-3 channel aids. This means that even though all four hearing aids meet the gain target for pure-tones, they yield different output levels (and possibly loudness) when presented with complex signals.

Problems with Target Matching Careful choice of stimulus is also important if one tries to match the gain/output of hearing aids to generic prescriptive gain target. Kuk & Ludvigsen indicated that, because of channel summation, the real-life output (ie, relative to complex signals) of multi-channel hearing aids matched to a prescriptive target using pure-tone sweeps will be higher than the real-life output of single channel hearing aids matched to the same target.

In general, the more channels there are and the higher compression ratio (CR) in each channel, the more noticeable this summation effect. This is seen in Figure 5a where four hearing aids—the 15-channel Senso Diva, the 3-channel Senso C+, the 2-channel Bravo, and the single-channel Logo—are matched in output to a NAL-R target using a pure-tone sweep presented at 60 dB SPL. Figure 5b shows the output of the same four hearing aids to the ANSI signal that has an overall level of 60 dB SPL (noise reduction deactivated). The output curve of the 15-channel Senso Diva is almost 10 dB higher than that of the single-channel Logo, and over 5 dB higher than those of the 2-3 channel aids. This means that even though all four hearing aids meet the gain target for pure-tones, they yield different output levels (and possibly loudness) when presented with complex signals.

Other Practical Implications If different stimuli (and durations of the same stimulus) result in different outputs from a nonlinear hearing aid, one cannot but ask which is the “optimal” stimulus for use to verify the output of hearing aids for the purpose of target matching. Using the optimal stimulus is necessary so that one will not reject or “mis-adjust” an otherwise satisfactory hearing aid on the grounds that it does not match target specifications. A corollary is that one may match a gain target in the clinic only to find the hearing aids to be too loud (or too soft) in real-life.

Another practical problem may be the choice of different stimuli across clinics. A clinical site that uses a particular stimulus may find a chosen hearing aid to match a specific target whereas another site, using a different stimulus, may find the hearing aid to miss the same gain target. The second site may either return or “mis-adjust” the hearing aid. This is a possibility especially when there are so many hearing aid (and/or real-ear measurement) systems—each with a different measurement bandwidth and each having many different...
stimuli and targets for the clinicians to choose. Such variability would be confusing for anyone, and it may even compromise the fitting.

So, What Should We Do?

It is obvious that the choice of stimulus has a significant impact on results when performing hearing aid verification. To decrease variability and to increase accuracy, it is important that we use the same "optimal" stimulus for verification. This means that one should use the same stimulus on which the prescriptive target is based, and measure the output with a measurement system that uses the same bandwidth as recommended by the prescriptive formula. This would ensure that the hearing aid is optimized for the specific listening condition used during verification. Hopefully, the results would generalize to other real-life situations. The following "lessons learned" should be considered when choosing a stimulus if the aforementioned conditions cannot be realized.

Judicious use of pure-tone signals:

While pure-tone signals have been used for many years for purposes such as quality assurance, evaluating attack/release times, and determining input-output curves, they do not reveal the processing of the hearing aid to complex signals. Furthermore, it may introduce artifacts (eg, "blooming" in the low-frequency) and compromise the goodness of target matching to generic formulae. On the other hand, if the hearing aid user complains that high-frequency sounds are too loud, using a high frequency sinusoid may be better than using a speech-shaped composite signal to verify the output of the hearing aid. But, in general, caution should be taken when using pure-tones if the goal is to estimate the performance of the hearing aid to real-life sounds.

Use of composite/complex signals:

Because composite signals allow one to examine the interaction among frequencies (including intermodulation distortion, channel summation, etc) and they are the sounds that one hears in everyday environments, they should be used for testing today's nonlinear hearing aids. Use of this type of signal should be acceptable for linear hearing aids as well.

On the other hand, one should not use complex signals such as real-speech target matching (real-ear or coupler) because of the difficulty in controlling the stimulus characteristics. For target-matching purposes, use a well-defined composite signal recommended by the specific prescriptive target.

Similarly, it would be extremely helpful if the developers of any fitting formula (generic or device-specific) specify the characteristics of the speech spectrum they used to formulate their targets, as well as the measurement bandwidth (1 Hz, 1/3 octave, etc) that one should use to measure the output of hearing aids. Manufacturers of hearing aid test systems (including real-ear) should also indicate how the output of hearing aids is analyzed. This would minimize variations in the measured output and increase the accuracy of hearing aid fitting.

Agree on a composite signal:

As illustrated before, there are many types of composite signals, each with its unique temporal-spectral characteristics. For example, the ANSI speech signal has more high-frequency energy than the ICRA signal; conversely, the ICRA signal has more low-frequency energy than the ANSI signal. Different stimuli result in different frequency-gain/output curves. This can hinder communication and lead to misinterpretation across clinics if each uses a different stimulus. It is important for the profession to agree on a standard signal for the purpose of verification and target matching. Currently, the ANSI S3.32 (1992) composite speech spectrum shaped signal has been the recommended standard. Other stimulus such as the ICRA signal which is shaped according to the idealized speech spectrum specified in ANSI S3.5 (1997) has been proposed as an alternative.

At the very least, clinicians need to specify the stimulus that they used for verification and check with the manufacturer of the device on the appropriateness of their chosen stimulus. If the stimulus may not be the most appropriate and if the clinician is not in a position to acquire new test stimuli, then he/she may at least be warned of the discrepancy and take this into consideration when interpreting the test results.

Examine both the output and the gain:

As illustrated before, examining the gain of a hearing aid could be deceiving. Because the ear hears the output of a hearing aid and not its gain, it is important to consider the output of the hearing aid during verification as well. For example, it would be more meaningful to examine if the output—measured in a similar manner as the ear would analyze sounds (eg, critical band or 1/3 octave)—is within the wearer's residual dynamic range and not whether it meets a particular gain target.

Know your hearing aids and the purpose of verification:

As indicated earlier, some digital nonlinear hearing aids have noise reduction algorithm that may give a faulty impression of their performance when tested using conventional stimuli like pure-tones or unmodulated ANSI signal. Especially with these hearing aids, one should fully understand the purpose for verification. If the purpose is to examine the frequency response of the hearing aid in quiet, one may deactivate the noise reduction algorithm during verification, present the stimulus for only a brief period, or use a stimulus that will not activate the noise reduction algorithm. If one is interested in examining the effect of the noise reduction algorithm, then the algorithm should remain activated, and a non-modulated composite noise should be presented.
for at least 10 s-20 s before one examines the noise-reduction effect.

The availability of different acoustic stimuli is driven by the types of signal processing available in today’s hearing aids. By its very nature, advances in technology always lead progress in standardization. Thus, the observations we have made may only be valid today and be outdated tomorrow. In this ever-changing industry, one thing that is certain is that the next generation of DSP hearing aids will challenge the appropriateness of the stimuli we have just discussed.

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References