Verification of Nonlinear Hearing Aids: Considerations for Sound-Field Thresholds and Real-Ear Measurements

Francis Kuk, Ph.D., CCC-A
Widex Hearing Aid Company, USA

Carl Ludvigsen, M.S.,
Widex ApS, Denmark

Audiologists have used the behavioral index of the sound-field aided threshold as a tool to verify the performance of linear hearing aids for many years. Its use on nonlinear hearing aids requires a different interpretation and extra precaution. In this paper, we will explore the meaning of the aided thresholds, review the variables that may affect its reliability, and compare its use with real-ear insertion gain measurement.

Introduction

A recent survey on hearing aids dispensed in the USA showed that as many as 20% were digital signal processing (DSP) hearing aids (Kirkwood, 2001a). This level of user acceptance is seen not only in the adult population, but also in children at various ages. The trend towards increased acceptance of digital hearing aids will likely continue as more are made aware of the potential benefits that DSP hearing aids provide . Informally, we also observed that more school-age children have replaced their analog hearing aids with digital hearing aids as more parents and audiologists realize the potential benefits of this type of technology.

Unlike analog conventional and programmable hearing aids that are relatively homogeneous in their processing algorithms and features, DSP hearing aids vary greatly in their processing and among the features from one manufacturer to another, and from one model to another model of the same manufacturer. Although almost all digital hearing aids are nonlinear (more gain for soft sounds and less gain for loud sounds), they vary in the complexity of their nonlinear processing. Some use simple, two-channel wide dynamic range compression (WDRC). Others include more complex features like multiple compression channels (as many as 21 channels), advanced loudness mapping algorithms, very low compression thresholds, noise reduction algorithms, feedback cancellation algorithms, adaptive directional microphone systems etc. The immediate question that many audiologists face in this technology maze is whether the verification measures they are employing for conventional hearing aids would be appropriate for DSP hearing aids. This may be especially problematic for those who work in an educational setting, where students may be wearing different makes of hearing aids recommended by their respective dispensing audiologists. It is practically impossible to know the special features of all the hearing aids used by the students so that appropriate verification of their performance can be made. On the other hand, it is important to verify that the students' hearing aids have met at least the most fundamental objective for hearing aids – that audibility for the softest speech sound is ensured.

These considerations led us to the purpose of this paper – to reaffirm the usefulness of a verification tool that provides the most fundamental and yet powerful information on the appropriateness of nonlinear hearing aids, digital or otherwise. The behavioral sound-field aided threshold measurement has been popularized since the 1960s (e.g., Markle & Zaner, 1966) and many clinicians and researchers have reported on its use and reliability (e.g., Hawkins & Schum, 1984; Macrae & Frazier, 1980). Because many of the discussions on this measure had been made in reference to linear hearing aids only, a re-examination of this measure as it applies to nonlinear hearing aids seems warranted. It is possible that the information that one obtains with this measure, the interpretation one can make with the results, and the method of measurement differ for linear and nonlinear hearing aids. Consequently, audiologists need to thoroughly understand the issues involved in this measure. This is especially relevant for audiologists working in an educational setting because Tharpe et al. (Tharpe, Fino-Szumski & Bess, 2001) showed that as many as 80% of audiologists working in such a setting routinely perform sound-field measurements to verify if the hearing aid is providing adequate benefit. In this paper, we will review the meaning and usefulness of the aided threshold (and functional gain) in the verification of nonlinear hearing aids, examine how to minimize its measurement variability, and compare the usefulness of this measure with real-ear (or insertion gain) measurement.

The meaning of the aided threshold (and functional gain)

Threshold is a behavioral response that reflects the lowest sound pressure level at which a child perceives the presence of a sound 50% of the time. In a clinical situation, the unaided threshold represents the lowest dial setting that produces a signal at the eardrum that exceeds the threshold criterion. In an aided condition, this represents the lowest dial setting that produces an input to the hearing aid microphone which, when added to its real ear gain (or in situ gain), reaches an output at the wearer's eardrum that exceeds the threshold criterion of the wearer. The difference in dial settings between the unaided and the aided thresholds is the functional gain provided by the hearing aid. The important point to realize is that the aided threshold is a behavioral response at one perceptual level, i.e. threshold. For a given hearing aid gain (or volume control, VC) setting, there is only one aided threshold and one value of functional gain. Although a higher output is possible from the hearing aid at a higher input, the aided threshold (and functional gain) remains the same. The aided threshold (as well as functional gain) changes only when the gain setting or volume control setting is changed.

A numeric example may clarify the previous discussion. For example, a child has an unaided sound-field (SF) threshold of 60 dB HL at 1000 Hz. If a hearing aid has a real-ear gain (not coupler gain) of 30 dB at 1000 Hz, the child will yield a threshold response (e.g., raise his hand) when the input level is 30 dB HL. This is because the input level of 30 dB HL, when added to the insertion gain of the hearing aid, yields an output of 60 dB HL, just enough to elicit a threshold response from the child. One can also see that the child will not respond at input levels below 30 dB HL (i.e., no aided threshold) because the output of the hearing aid is below the threshold of the child at such inputs (e.g., 20 dB HL input + 30 dB gain = 50 dB HL). On the other hand, the aided threshold will remain at 30 dB HL even though the stimuli can be increased to higher levels, e.g., 40 dB HL. In this case, even though the output of the hearing aid (40 dB HL + 30 dB gain = 70 dB HL) is higher than the unaided threshold of 60 dB HL, the child still exhibits a threshold response.

On the other hand, if one increases the insertion gain of the hearing aid to 40 dB (e.g., by turning up the volume control wheel), an input level of 20 dB HL will elicit a threshold response from the child (i.e., 40 dB gain + 20 dB HL input = 60 dB HL). Thus the aided threshold is lowered as the gain or VC is increased. Conversely, if the insertion gain on the hearing aid is lowered to 20 dB, the aided threshold will increase to 40 dB HL instead.

These examples suggest that the aided threshold represents the lowest dial setting on the audiometer that is required by the child to detect the presence of the input when the hearing aid is worn at a particular gain or volume control (VC) setting. In other words, if the child does not adjust the VC further in his/her daily environments, the aided thresholds will reflect the softest sound that the child hears in such environments. This may not be true in a linear hearing aid because the VC will need to be adjusted in daily situations to accommodate the changes in input levels. However, the nonlinearity in a WDRC hearing aid is designed to overcome variations in input levels. That is why a VC may not be needed in WDRC hearing aids and many are manufactured without such a control. Applying the definition that functional gain is the difference between the aided and the unaided thresholds suggests that functional gain reflects the improvement in auditory sensitivity of the child consequent to wearing the hearing aid.

The above examples show that the magnitude of the aided threshold is inherently tied to the gain setting on the hearing aid. There are two implications to this observation. First, one can achieve a low aided threshold (e.g., 0 dB HL or lower) or high functional gain as long as the hearing aid allows such gain adjustment. In practice, a number of factors such as circuit noise, feedback, and loudness discomfort could set a limit. The second issue relates to the meaningfulness of such an index. If a change of gain on the hearing aid (as mediated through the use of the volume control, VC) changes the aided threshold (or functional gain), the assertion that functional gain represents the insertion gain for conversational speech (Mason & Popelka, 1986), or that the aided threshold represents the softest sound that one hears would not be valid. Such an assertion is only valid on hearing aids where no VC is available or no wearer gain adjustment is allowed.

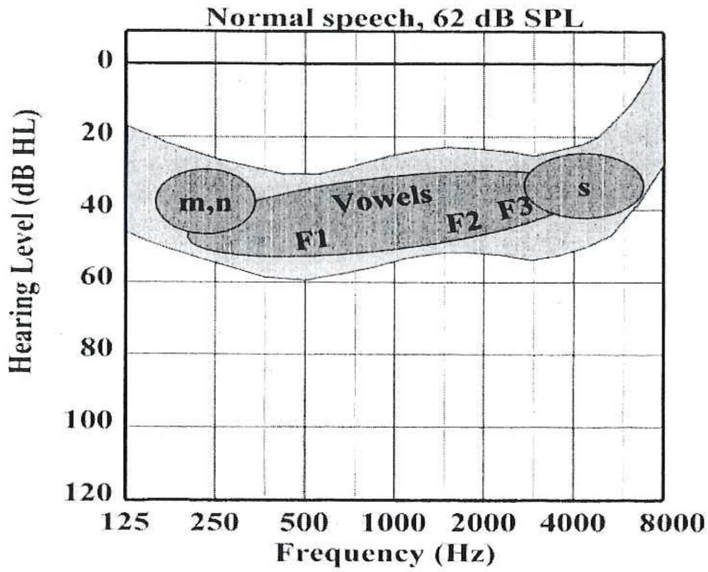
What should be the ideal aided threshold?

Because the aided threshold reflects the auditory sensitivity of the child with the hearing aids, it is logical to ask where the ideal aided threshold should be. A confounding factor in such consideration is the presence of the volume control. This is because the VC allows the child to adjust the output from the hearing aids to be different from the level at which the aided thresholds are determined. This would alter one's interpretations of the aided threshold. Consequently, the following discussion assumes the use of nonlinear hearing aids that do not allow wearer adjustment of the recommended settings or that the range of gain change offered by the VC is known. This is not unreasonable given that many high performance digital hearing aids today do not have a VC. Changes in gain levels on such hearing aids are achieved through automatic gain control (AGC) algorithms.

If hearing aids were to enable the child with a hearing loss to hear the softest sound comparable to a normal hearing person, then they must provide sufficient gain to overcome the threshold displacement resulting from the hearing loss. If one can assume that the performance of a hearing aid is not limited by physical constraints like circuit noise and acoustic feedback, then the theoretically optimal aided threshold should be 0 dB HL since normal threshold is defined at 0 dB HL. Consequently, one must provide 40 dB of functional gain to someone with 40 dB hearing loss and 80 dB to someone with 80 dB hearing loss so that all hearing-impaired people can hear sounds as soft as 0 dB HL regardless of their degree of hearing loss. However, because of microphone noise and the difficulty with feedback from high gain, the practical aided threshold may have to be higher than 0 dB HL.

If one assumes that the primary function of hearing aids is to allow speech comprehension, the ideal aided threshold should be determined using the audibility of the lowest level of normal speech as the criterion. Figure 1 shows the speech spectrum displayed on an audiogram. In this case, the lowest level of normal speech is around 20 dB HL across frequencies. Using this criterion, the optimal aided threshold should be equal to or better than 20 dB HL across frequencies. The ideal aided threshold could be even lower in order to ensure audibility for shorter and/or softer speech stimuli.

Figure 1. Audiogram with the average speech spectrum superimposed. Note the softer part of average speech occurs at around 20 dB HL across frequencies.



Would the type of processing matter?

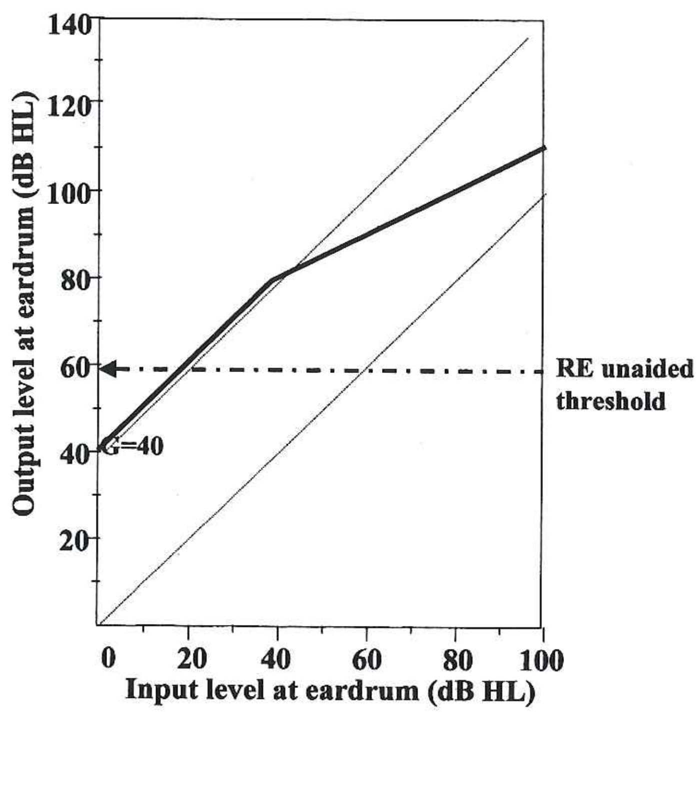
The type of processing could affect the acceptability of the optimal aided threshold. For a linear hearing aid (same gain at all input levels), achieving a low aided threshold would require high gain from the hearing aid for someone with more than a moderate-to-severe degree of hearing loss. While gain for soft sounds may be adequate with a low aided threshold, gain for medium and loud sounds may become excessive unless one can adjust the VC setting. Otherwise, it would cause discomfort and increase the risk of additional hearing loss (Macrae, 1995; Markle & Zaner, 1966) from over-amplification. Because a person with a mild-to-moderate hearing loss would typically require less gain, achieving an aided threshold at 20 dB HL (or lower) may be acceptable only for mild-to-moderate losses with linear hearing aid use. One may have to accept an aided threshold that allows audibility of medium level sounds (i.e., 30-50 dB HL) for people with severe-to-profound hearing losses with linear hearing aids.

As a numeric example, consider a child with a 50 dB hearing loss at 1000 Hz. A functional gain of 30 dB must be provided by the hearing aid in order to achieve an aided threshold at 20 dB HL. Because a linear hearing aid provides the same gain at all input levels (prior to saturation), an output of 90 dB SPL will be measured in the child's ear at an input of 60 dB SPL, and 110 dB SPL at an input of 80 dB SPL. This level of output is high, but may be tolerable to a child.

Consider now a child with 80 dB hearing loss. One would have to apply 60 dB of functional gain in order to obtain an aided threshold at 20 dB HL. At that gain level, the same linear hearing aid will have an output of 120 dB SPL for a conversational input of 60 dB SPL, and as much as 140 dB SPL output when the input is 80 dB SPL! Unfortunately, an input level of 80 dB SPL is not atypical. The sound pressure level of a child speaking at a normal conversational level will measure 80-85 dB SPL at his/her own ear level . Unless some type of output limiting mechanism is available on the hearing aid, and unless the child has the cognitive and physical ability to adjust the VC on the hearing aid, this output level could be excessive and may result in additional hearing loss over time.

An aided threshold at 20 dB HL or lower may not be as problematic for nonlinear hearing aid wearers as it may be for linear hearing aid wearers. This is because a nonlinear hearing aid reduces its gain as input increases beyond the compression threshold (CT). At a high input level, the output of a nonlinear hearing aid that has the same gain as a linear hearing aid at a low or medium input would be lower than the linear hearing aid. Figure 2 shows the output difference between a linear and a nonlinear hearing aid having the same aided threshold at 20 dB HL. In this example, both hearing aids (linear is dotted line, and nonlinear is solid line) provide 40 dB of gain below a 40 dB HL input. This yields an output of 80 dB HL at an input of 40 dB HL. However, at an 80 dB HL input, the linear hearing aid yields an output of 120 dB HL, whereas the nonlinear 100 dB HL. This is a difference of 20 dB! Thus, a low aided threshold should be achieved with a nonlinear hearing aid in order to achieve extra audibility while minimizing the risk of discomfort and potential over-amplification. Obviously, evaluating for potential discomfort is still necessary.

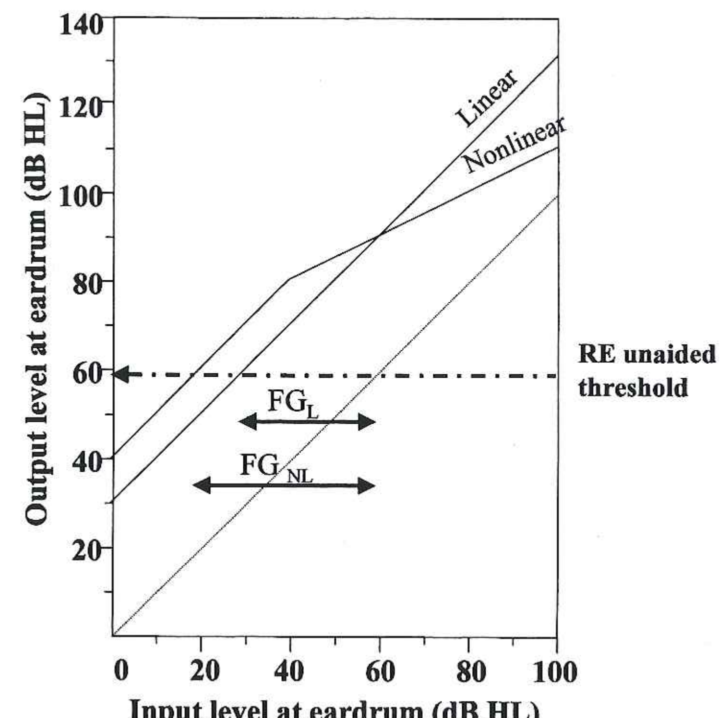
Figure 2. Hypothetical output difference between a linear and a nonlinear hearing aid that have the same aided threshold around 20 dB HL. Note the gain decrease in the nonlinear hearing aid above 40 dB HL.



The fact that the aided threshold in a nonlinear hearing aid reflects the softest sound that the child hears is extremely valuable for verification and counseling purposes. For the purpose of verification, one can compare the value of the aided threshold to a target aided threshold that would suggest optimal audibility. The task during verification is to adjust the settings on the hearing aid to reach that target. Our previous discussion suggests that the optimal aided thresholds across frequencies should be around 10 - 20 dB HL for up to a moderately severe degree of hearing loss. Because of feedback issues, it may be practical to expect a higher level (e.g., 30-40 dB HL) for people with a severe-to-profound loss unless an active feedback cancellation mechanism is available on the same hearing aid to minimize feedback. Also, recent reports on desensitization and dead regions (e.g., Ching et al, 1998; Hogan and Turner, 1998; Moore, 2001) would suggest that it may not be desirable to have a low aided threshold when the hearing loss for certain hearing loss configurations (e.g., steeply sloping losses) exceeds a particular degree (e.g. 60-70 dB HL). Aside from these caveats, it seems reasonable to aim for a low aided threshold from a nonlinear WDRC hearing aid in order to maximize audibility. The specific conditions in which this hypothesis is valid (and invalid) would need to be delineated in future research.

To fit a linear hearing aid optimally, most fitting approaches recommend gain to bring conversational speech to the wearer's most comfortable listening range (Skinner, 1988). Thus it is not surprising to find that functional gain for a linear hearing aid reflects gain for conversational speech. On the other hand, for a nonlinear hearing aid to provide the same insertion gain at the medium input level as a linear hearing aid, it would provide more gain for lower input sounds and less gain for higher input sounds. Figure 3 illustrates the case by comparing the static input-output curves of a linear hearing aid and a nonlinear hearing aid that are matched in output at a medium input level. One can expect a lower aided threshold (and higher functional gain) from the nonlinear hearing aid than the linear hearing aid in this case. Indeed, the nonlinear hearing aid would allow the child to hear sounds as soft as 20 dB HL, whereas the linear hearing aid would allow the child to hear sounds at or greater than 30 dB HL. Clinically, this means that if a child who has been wearing linear hearing aids is fitted with nonlinear hearing aids, one should expect to achieve higher functional gain (or a lower aided threshold) with the nonlinear hearing aids than the linear hearing aids if both hearing aids sound equally loud at a medium conversational level. If both hearing aids yield the same aided threshold, either the nonlinear hearing aids may not be providing adequate gain for medium input sounds, or the linear hearing aids may have been providing too much gain in the past. In both cases, the child may complain that the nonlinear hearing aids are not loud enough even though the audibility for soft sounds is similar for both hearing aids. One may opt to increase the gain setting on the nonlinear hearing aid in this case. On the other hand, if the aided thresholds with the nonlinear hearing aid are better than those with the linear hearing aids and the child still comments of a "weaker" volume, s/he may be reacting to the reduced loudness typical of a nonlinear hearing aid. In such a case, it may be sufficient to counsel the child to adapt to the new loudness percept.

Figure 3. Hypothetical input-output curves of a linear and a nonlinear hearing aid mathcedin output to a medium (60 dB HL) input. Note that the nonlinear hearing aid has a higher output below the 60 dB HL input. This would result in a lower aided threshold and a higher functional gain. FG_L refers to the functional gain obtained with the linear hearing aid, and FG_{NL} refers to the functional gain obtined with the nonlinear hearing aid.



From a counseling standpoint, the fact that the child has aided thresholds around 20 dB HL across frequencies is indicative of his/her ability to hear sounds as soft as this level. Realizing that the average speech spectrum (as displayed on an audiogram format) ranges from 20 dB HL to 50 dB HL (seen in Figure 1), aided thresholds at 20 dB HL suggest that the child has the ability to hear the softest speech sounds with the use of the hearing aids. Aided thresholds that are higher than this would indicate inaudibility until that level at the indicated frequencies. This information made in reference to the speech spectrum is meaningful and easy to explain to parents, teachers or other personnel involved in the care of the child. It may also serve as a basis to validate the child's complaints about inadequate loudness (as described previously). Better yet, this interpretation can be applied to any hearing aids regardless of its technological platforms and processing complexity.

While the value of the aided threshold as a fundamental measure of hearing sensitivity for any hearing aid is indisputable, one must guard against over-extending its applications. As indicated earlier, the aided threshold reveals the softest sound that the child hears when the VC is not adjusted. However, it does not reflect the perception of sounds at levels other than the threshold (which is typically low). Even so, it reflects detection ability only. It does not reflect or predict speech perception skills (or speech recognition scores in quiet or in noise) of the child or if the output of the hearing aid may be uncomfortable. Furthermore, it does not reflect the processing of any special features (e.g., noise reduction) on the hearing aid other than its potential gain for soft sounds at different frequencies in quiet. Thus, only the gain parameters that are related to low input levels (e.g., compression threshold, gain for low input) would have a direct effect on its magnitude. Gain parameters for high input levels will not affect the measured aided threshold. In short, the aided threshold reflects the potential hearing sensitivity of the child unbiased by any special processing on the hearing aid. Nonetheless, the information provided by this index on audibility is critical because every aspect of auditory perception – from a simple detection task to a complex speech understanding task originates from the child's ability to hear the sound first. The aided threshold is the only verification index that provides such information.

Measurement variables

One of the hesitations that may have prevented some audiologists from obtaining aided thresholds routinely is its noted measurement variability (Arlinger & Jerlvall, 1987; Hawkins et al., 1987; Humes & Kirn, 1990). However, measurement variability is inherent in all clinical procedures when human responses are obtained. Rather than abandoning its use and missing the information that it provides, a more appropriate approach is to minimize the source of variability. Walker (1995) provided an excellent tutorial on the factors that affect sound-field threshold measurement. Some of these variables include:

Noise – Ambient room noise and low level noise in the test environments may act as maskers and raise the level of the aided and/or unaided thresholds and alter the magnitude of the functional gain. This would be especially true for frequencies below 500 Hz. Children with normal or a mild hearing loss in low frequency would be affected especially. Consequently, test rooms for conducting sound-field testing must be free from any extraneous noise sources. Furthermore, Macrae and Frazier (1980) pointed out that circuit noise from hearing aids could impose a floor effect on the aided threshold and raise its level (thus decreasing functional gain).

Standing waves – Because most test environments are enclosed, standing waves will likely develop from reflections off the walls of the test booth. To overcome such occurrence, the use of frequency modulated pure tones (or warble tones) as the test stimuli is recommended because they cover a narrow frequency region and are less susceptible to room resonance (Walker, 1995).

Head and body movement – Any movement by the test subjects during sound-field testing might result in a threshold shift since the sound level is likely to differ at different positions due to heterogeneity of the sound field. The effect may be more significant in the higher frequencies because of their shorter wavelengths.

VC adjustment – A source of variability in aided threshold measurement is that subjects were sometimes allowed to adjust the VC on the hearing aid (Hawkins et al., 1987). Subjects should not be allowed to make any VC adjustment on the hearing aid during testing and upon re-test.

Identical test conditions – Aided and unaided threshold measurements must be obtained under identical test conditions in order to compute functional gain. It is inappropriate to compute functional gain by comparing the monaural unaided thresholds obtained under headphones (or inserts) to the aided sound-field thresholds obtained in a binaural manner. The true functional gain provided by a hearing aid is the difference between the monaural unaided threshold and the monaural aided threshold obtained in the same sound-field test environment. Use of earplugs and/or masking under headphones may be necessary to prevent the non-test ear from participation.

Nonlinear hearing aids – Because of the dynamic change in gain characteristics over time, the timing characteristics (attack and release times) of a nonlinear hearing aid could interact with the test stimuli and affect the reliability of the measured aided thresholds. This is especially true in slow-acting WDRC hearing aids and hearing aids with a low CT.

First, because a compression hearing aid decreases its gain as input level increases beyond the CT, the typical bracketing approach in threshold estimation may yield more variable results in testing a nonlinear hearing aid than a linear hearing aid. A bracketing approach involves frequent, large intensity difference (and thus gain difference) between stimulus presentations (e.g., up 10 dB and down 5 dB or 15 dB in each "bracket"). This could result in frequent unpredictable output changes. While test stimuli that are below the CT would not activate the compression circuit to introduce output uncertainty, those that are at or above the CT of the hearing aid may interact with the time constants of the compression hearing aid to yield unpredictable output. An alternative approach is to present test stimuli in a 5-dB, ascending manner once the vicinity of the aided threshold is bracketed. This would minimize unpredictable gain swing and its associated variability in the aided threshold measure. In addition, a stimulus duration of approximately 1-2 seconds should be used for a reliable measure.

The interval between stimulus presentations could also affect the reliability of the aided threshold. Delays between presentations mean that the test stimuli could be presented at different times during the gain recovery phase of the hearing aid. This means that each stimulus could potentially receive different gain (to result in a different output) to yield variable aided thresholds. If possible, one should wait for the period of the release time before the next presentation to minimize variability from presentations. However, if one uses an ascending approach and assumes that the intensity difference between presentations is typically 5 dB, the minimum required interval between presentations can be as brief as 1-2 s even for a slow-acting WDRC hearing aid with a low CT.

In summary, the attack time and the release time on a nonlinear hearing aid may affect the reliability of the obtained aided threshold. A nonlinear hearing aid with a high CT (assuming linear processing below) may yield less variability in the aided threshold than ones with a lower CT. The use of expansion below the CT may add variability further. Typically, an ascending approach with a stimulus duration of 1 to 2 s and an interval between presentations of 1 to 2 s should ensure reliable outcome.

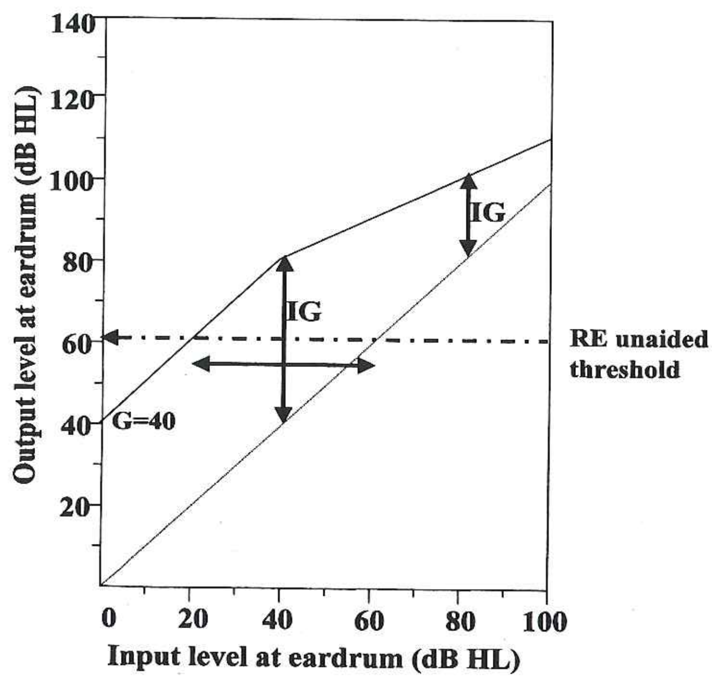
Relationship between functional gain and insertion gain

It is inevitable that some will compare the measure of functional gain (and aided threshold) with real-ear insertion gain measure. After all, this measure originated from the same need as functional gain in that the coupler gain of a hearing aid does not reflect the actual gain available to the hearing aid wearer. By measuring the sound pressure delivered in the child's ear at a particular presentation level without any hearing aid (real-ear unaided response, REUR) and with the child's hearing aid (real-ear aided response, REAR), the audiologists compute the difference between the REUR and the REAR as the real-ear insertion response (REIR) or the real-ear insertion gain (REIG) at the intended input level. Because functional gain measure also accounts for these variables, many (e.g., Hawkins & Schum, 1984; Mason & Popelka 1986) suggested that functional gain should be identical to the measured insertion gain in linear hearing aids. This may be true for a linear hearing aid, but may not be so for a nonlinear hearing aid.

A numeric example may help in its understanding. Consider the same child with the same 60 dB hearing loss at 1000 Hz. Let us also assume that the same linear hearing aid with a 30 dB functional gain is used for the real-ear measurement (although in this case, the actual insertion gain is the focus of measurement). When an input of 50 dB SPL is presented in an unaided condition, a REUR of 52 dB SPL is measured at 1000 Hz (due to the sound-field-to-eardrum transfer at 1000Hz). When the hearing aid is worn, a REAR of 82 dB SPL is measured. This corresponds to a real-ear insertion gain of 30 dB (82 – 52 or 30). This is identical to the functional gain. As the input level increases (e.g., 70 dB), the aided response increases (in this case, 102 dB SPL). Nonetheless, the same insertion gain is obtained (102-72 or 30 dB).

The dissimilarity between functional gain and real-ear insertion gain becomes apparent in nonlinear hearing aids when the input level changes. Figure 4 shows the input-output curve of a nonlinear hearing aid with insertion gain of 40 dB below the compression threshold (at 40 dB HL). For a 60 dB hearing loss for which this input-output (I-O) curve is recommended, the aided threshold is measured at 20 dB HL. This corresponds to 40 dB functional gain. The insertion gain, however, differs from 40 dB depending on the level of the input signal. An input signal of 40 dB HL would yield an aided output at 80 dB HL, or an insertion gain of 40 dB. Indeed, the insertion gain will remain at 40 dB as long as the input level is lower than the compression threshold (in this case, less than 40 dB HL). On the other hand, the aided output is 90 dB HL when the input is 60 dB HL. This translates to an insertion gain of 30 dB instead. Indeed, the higher the input above the CT, or the higher the compression ratio, the lower is the insertion gain, and the greater is the difference between functional gain and insertion gain in a nonlinear hearing aid. Other researchers have reported also discrepancy between measured insertion gain and functional gain in hearing aids (e.g., Schwartz and Larson, 1977; Seewald et al, 1992; Stelmachowicz and Lewis, 1988). .

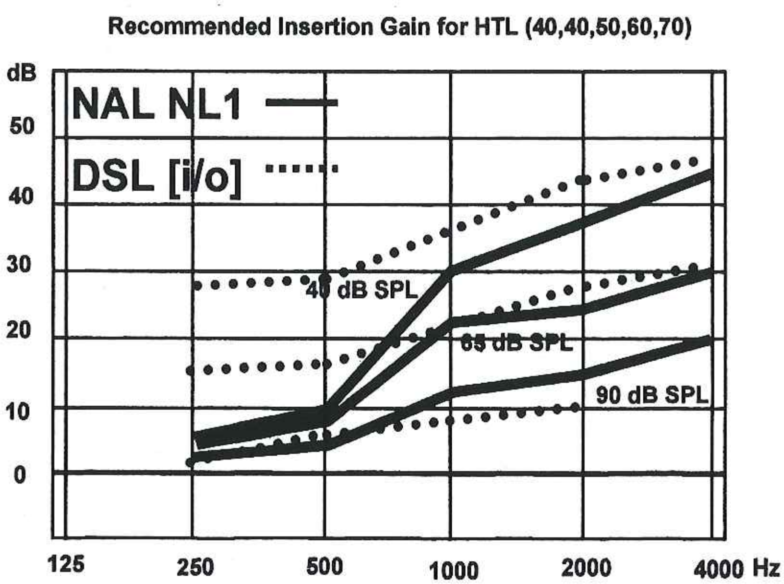
Figure 4. Hypothetical difference between functional gain and insertion gain at a gain setting of 40 dB on a nonlinear hearing aid. The two terms are identical for inputs below 40 dB HL. (compressing threshold, CT). The two are different for inputs above the CT.



Functional gain and insertion gain should not be interpreted the same way. Insertion gain measures reflect the gain (and output) of the hearing aid at a specific input level. This is meaningful information especially for nonlinear hearing aids where gain changes at different input levels. Using this tool, one may examine the output of the hearing aid to a medium level input or one may examine the output at a high input (e.g., OSPL90). Also, the effect of special processing like noise reduction, directional microphone etc on the output may be examined using real-ear measures under proper stimulus and test conditions. Such information cannot be determined with functional gain measures.

However, one should recognize that the ability to obtain different outputs at different input levels by itself may not be sufficient. For example, knowing that the output of a hearing aid is 80 dB SPL when the input is 60 dB SPL does not tell us the appropriateness of this output. If the child has a 90 dB hearing loss, an output of 80 dB SPL will be inaudible; whereas a child with severe hyperacusis will find this 80 dB SPL output intolerably loud. Consequently, one has to have specific guidelines to interpret the appropriateness of the output. Generic prescriptive gain targets (e.g., NAL-NL1,; DSL[i/o],) are formulated to offer guidelines for the "average" individual. One of the difficulties in using these targets is that each specifies a different gain/output target. For example, Figure 5 compares the DSL [i/o] and the NAL-NL1 targets for a moderate hearing loss. The significant difference in the recommended gain at various input levels raises the question of the "right" formula to use. While the debate over the right target will continue, it has been shown repeatedly that as long as audibility for soft sounds and comfort for loud sounds are ensured, there is minimal difference in the real-world efficacy among different targets . Thus, the value of knowing the exact output through real-ear measurement may be limited. Indeed, in the adult literature, it has been demonstrated that real-ear output that met a prescriptive target does not necessarily predict real-world satisfaction of the hearing aids or the wearers' speech recognition scores .

Figure 5. Prescribed gain difference between the NAL-NL1 and DSL [i/o] targets for a moderately sloping hearing loss.



Another difficulty with prescriptive targets is that many have not considered the effects of channel number and time constants on the gain formulation . This has the effect of altering the real-world output of a multichannel nonlinear hearing aid. For example, Kuk and Ludvigsen showed that as the number of channels increases, the real-world output of a nonlinear multi-channel hearing aid to a complex signal increases. Consequently, if one uses one of the generic prescriptive formulae to adjust the output of a multichannel DSP hearing aid, one needs to be cognizant that its real-world performance may deviate significantly from the intended target.

There are also other differences between functional and insertion gain measures. Insertion gain does not reflect subjective perception or audibility. One can determine insertion gain on KEMAR but one cannot measure functional gain or the aided threshold on the same manikin. It is not tied to the wearers' threshold perception and no subjective participation is necessary for its determination. One advantage is that the variability arising from inconsistent threshold criteria would be eliminated. However, variability from measurement error would still be present (e.g., Tecca et al. 1987). Errors from the placement of the loudspeakers, depth of insertion of the probe tube etc. would alter the actual sound pressure level measured in the earcanal. Thus, the two gain terms serve different purposes and should not be used interchangeably.

From a measurement standpoint, sound-field thresholds can only be determined using warble tones or narrow bands of noise in order to provide frequency-specific gain information. Use of broadband stimuli such as speech-shaped noise would only yield threshold information for a narrow frequency range where the sound spectrum matches the threshold. The exact frequency contributing to the threshold response might not be known precisely. This would be true for both linear and nonlinear hearing aids. On the other hand, insertion gain can be measured using any types of acoustic stimuli, e.g., sweep tones, speech-shaped noise, white noise etc. to yield the same results if the hearing aids are linear. However, the same will not be true for multichannel nonlinear hearing aids. Different insertion gain may be reported depending on the nature of the stimulus and the processing of the nonlinear hearing aids (e.g., number of channels, compression ratio, compression threshold etc). The choice of stimulus for insertion gain determination in nonlinear hearing aids must be done judiciously.

Conclusions

The determination of the aided sound-field thresholds allows audiologists to estimate the lowest input level that is audible to the wearer of nonlinear hearing aids. This index provides information on the ability of the hearing aid to meet its fundamental fitting objective, i.e., ensuring audibility of the softest sound. Such information is not provided by other verification measures and is important in the verification and fine-tuning of all hearing aids. After all, no auditory processing may take place without audibility. Thus, one should strive to obtain this index as reliably and as accurately as one can; and interpret its values in a meaningful manner. Verification measures like real-ear measurement may be necessary if one is interested in the exact output of hearing aids. However, one must also be careful in the interpretation of its results to avoid over-interpretation.

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