Considerations in Modern Multichannel Nonlinear Hearing Aids

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Introduction

The field of hearing aids has witnessed a significant evolution in the last decade. Prior to the 1980s, over 80% of hearing aids manufactured and dispensed were linear hearing aids with peak clipping as the only option for output limiting. Many of the nonlinear hearing aids were linear hearing aids with output compression limiting. The situation changed so that in the mid-1990s almost half of the hearing aids sold were nonlinear hearing aids (Mueller and Killion, 1996). Most of the nonlinear hearing aids were single channel linear hearing aids with compression limiting, or single channel wide dynamic range compression (WDRC) hearing aids. The sale of nonlinear hearing aids steadily increased in the last few years, with an estimate that over 75% of hearing aids sold incorporated some types of nonlinear signal processing. Of these hearing aids, three-quarters were either programmable or true digital signal processing hearing aids (Strom, 2001). Many of these hearing aids are multichannel devices utilizing complex processing algorithms that extend the benefits of conventional nonlinear hearing aids. Such rapid change in technology has led to an almost inevitable consequence—our verification protocol has failed to keep pace with the advances such that many can neither be easily evaluated nor fairly verified without special considerations. The interpretation that one makes with the results of verification may also be unclear. Failure to properly verify a nonlinear hearing aid could compromise its potential benefit. Thus, this chapter focuses on the changes in current nonlinear hearing aids, explains how they may extend the benefits provided by conventional devices, and discusses how our current practices on verification may need to be updated to reflect the changes in technology. Although the concept and characterization of compression are reviewed, readers are referred to summary articles for a detailed discussion on the rationale, classification, fitting, and efficacy of conventional nonlinear hearing aids (Braida et al, 1979; Walker and Dillon, 1982; Kuk, 1996, 2000; Dillon, 1996.

Characterization of Conventional Compression Hearing Aids

One way to study the operation of a hearing aid is to examine its static I-O curve. This curve can be determined with a steady-state sinusoid (ANSI S3.22–1996) or with a composite noise signal (ANSI S3.42–1992). Because of its simplicity, most reports use a
2000-Hz sinusoid as the stimulus to illustrate the static and dynamic properties of a compression hearing aid. Figure 5–1a shows an I-O curve of a simple compression hearing aid to illustrate the terminology.

### Static Characteristics

**gain (g)**

Gain of a hearing aid is calculated as the difference between the output level and the input level prior to saturation (or the output level beyond which distortion products like harmonic and intermodulation distortion product are present). For a compression hearing aid, a fixed gain value may be difficult to establish because its gain decreases with increasing input level. To meet quality assurance purposes, most manufacturers report the maximum gain of the compression hearing aid. This is usually the gain at the low compression threshold. An I-O curve (ANSI, 1996), or a series of output frequency gain curves at different input noise levels (ANSI, 1992) will be needed to determine gain at various inputs. Figure 5–1a shows an I-O curve and Figure 5–1b shows an input-gain (I-G) curve. These figures show that gain is 30 dB (90 dB - 60 dB = 30 dB) below an input of 60 dB SPL. However, gain becomes 25 dB at an input of 70 dB SPL and 20 dB at an input of 80 dB SPL. It is easy to see that compression is similar to gain reduction.

**compression threshold (ct)**

CT represents the lowest input sound pressure level (SPL) at which the hearing aid reduces its gain. This is frequently described as the knee point on the I-O curve. For conventional compression hearing aids, linear processing is typically employed below the CT. In conventional input compression hearing aids where the volume control of the hearing aid is placed after the compressor (Kuk, 1996), there is usually one CT that typically ranges from 40 to 60 dB SPL. In Figure 5–1a,b the CT is at 60 dB SPL. As indicated earlier, the CT represents the point of maximum gain on an input compression hearing aid.

**compression ratio (cr)**

CR represents the ratio of the change in input levels to the corresponding change in output levels when the SPL of the input is within the compression range (i.e., CR = change in input/change in output). The nominal CR is calculated when the output reaches its steady-state value. The CR can also be calculated as the inverse of the slope of the I-O curve within the compression range. In Figure 5–1a, the CR is calculated to be 2:1 \(\frac{(90 - 60)}{(105 - 90)} = 2\),
whereas the slope of the I-O curve is 0.5 \([\frac{105 - 90}{90 - 60} = 0.5]\).

**Compression range**

Compression range represents the range of input at which the hearing aid is in compression. In this case, it is 30 dB (60 to 90 dB SPL).

**Dynamic Characteristics**

**attack and release times**

Compression circuits operate through the use of feedback loops. The time taken for the compression hearing aid to change from linear gain to within 3 dB of the compressed steady state (i.e., reduced gain) is called the attack time. The time taken for the compression hearing aid to return from the compressed state to within 4 dB of the linear steady state (i.e., full gain) is called the release time (ANSI, 1996).

The ANSI S3.22–1996 standard (ANSI, 1996) specifies that attack and release times be measured with a sinusoid (typically 2000 Hz) that abruptly changes from 55 to 90 dB SPL. The “duration of each of the levels shall be sufficiently long so as to not influence the measured attack and release times” (ANSI, 1996, p. 11). The use of two signals with different steady-state duration is required to test compression hearing aids with adaptive and/or multiple release times. The assumption for using the 55 to 90 dB SPL signal level is that 90 dB SPL is typically above the compression threshold and 55 dB SPL is typically below the compression threshold. The assumption for the low input level is not true for many digital nonlinear hearing aids.

**Differentiating among Conventional Hearing Aid Circuits**

Compression hearing aids are distinguished from each other by differences in their static and dynamic characteristics. As a prelude to the discussion on how technology has improved nonlinear hearing aids, it is important to review how these compression parameters are chosen to achieve various functions. Dillon (1996) provides an excellent tutorial on such classification as well as the efficacy for each type of compression circuit.

**Linear—No Compression-Preserving Input Intensity Difference**

Linear processing refers to an amplification scheme in which the same amount of gain is applied to the range of inputs of interest. Figure 5–1b shows that the range of input from 0 to 60 dB SPL receives linear processing because an input of 40 dB SPL receives the same amount of gain (30 dB) as an input of 60 dB SPL. A corollary is that for a given change in input range (e.g., 0 to 60 dB, or the 60-dB range), the same change in output range occurs (e.g., 30 dB to 90 dB, or 60-dB range for a 30-dB gain). Because the same intensity range is seen in the input and the output, a major advantage of linear processing is its preservation of the intensity difference (or the waveform envelope) of the input signal. Such difference can be an important speech cue, especially for patients with more than a moderate degree of hearing loss (Boothroyd et al, 1988; Dreschler, 1989; Van Tasell, 1993).

Gain on most linear hearing aids is adjusted for a medium conversational (around 65 dB SPL) input (Skinner, 1988). High-input-level sounds (>70 dB SPL), when added to the fixed gain of a linear hearing aid, may sound uncomfortably loud. Although one can limit the output of the hearing aid to below the wearer’s uncomfortable listening level (UCL) using a peak clipping circuit, this method creates high distortion, which may affect sound quality and speech intelligibility. On the other hand, low-input-level sounds (<50 dB SPL) might not be audible. Thus, a volume control (VC) has been made available to allow gain adjustment in different environments.

**Compression Limiting—High CT (>70 dB SPL), High CR (>8:1), Short Attack (<10 ms)—Maintaining Comfort While Minimizing Distortion**

One use of compression in hearing aids is to limit the maximum output through the use
of compression limiting (CL) or output compression limiting (OCL). Instead of providing the same fixed gain at all input levels, a linear hearing aid with CL provides the same fixed gain up to a relatively high input level (usually above an input of 70 dB SPL, or an input that results in an output level that is just below the wearer’s UCL). Thereafter, drastic gain reduction occurs (high compression ratio, usually greater than 8:1) so that high input sounds will neither become too loud nor saturate the hearing aid to result in distortion. It should be noted that CL is an approach to limit the output of hearing aids. As such, it has a very short attack time (less than 10 msec) and a moderately long release time (around 100 to 200 msec). Typically, it is used in linear hearing aids and is placed at the output stage of the hearing aids. Thus, it has also been called an automatic gain control output (AGC-O) circuit. Recently, such a circuit has also been placed at the input stage of hearing aids to minimize distortion at the input, as well as at the output stage of nonlinear hearing aids using WDRC. Thus, the term AGC-O is less meaningful to characterize a compression-limiting circuit than it once was.

Hearing aids with compression limiting have been used commercially since the 1950s (Davis et al, 1947; Hudgins et al, 1948; Lynn and Carhart, 1963). The general research suggests that compression limiting is preferable over peak clipping circuits (e.g., Dillon, 1988; Dawson et al, 1991; Hawkins and Naidoo, 1993). This is probably because of its ability to reduce distortion while maintaining the temporal and spectral integrity of the signal most of the time (because of linear processing in the hearing aid). Dillon (1988) suggested that patients with a mild-to-moderate degree of hearing loss would most likely enjoy such benefits. Additionally, patients with a severe-to-profound hearing loss who have sufficient residual frequency selectivity may also benefit from reduced distortion as long as sufficient output is available from the hearing aid. Killion and Fikret-Pasa (1993) also recommended compression limiting for their type III patients who have a severe-to-profound sensorineural hearing loss.

Wide Dynamic Range Compression (WDRC)—Low CT (<60 dB SPL), Low CR (<4:1), Short Attack (<10 msec) and Release (<50 msec)—Ensuring Audibility and Comfort Without VC Adjustment

Another difficulty with linear amplification is that soft sounds may not be audible all the time unless the wearer actively adjusts the VC on the hearing aids. This could happen with a change in vocal effort of the speaker (Pearsons et al, 1977), or when the speaker-listener distance increases. Although an adult listener may be able to adjust the VC on the linear hearing aid to compensate for the change in input levels, it is unreasonable to expect a very young child to be able to do the same. Unfortunately, it is more likely to find substantial changes in the input levels of a child’s environment than of an adult’s (Stelmachowicz et al, 1993).

To ensure audibility for soft sounds while maintaining intelligibility and comfort for medium level and loud sounds, a hearing aid must provide sufficient gain so that soft sounds, when amplified, exceed the threshold of the wearer. The same amount of gain must be reduced for louder sounds to minimize discomfort at the higher input levels. Thus higher gain (than linear) should be provided for soft sounds, and gain reduction (or compression) should occur above the level of soft sounds so that audibility, intelligibility, and comfort can be ensured at all input levels. Nonlinear hearing aids with such characteristics are often called wide dynamic range compression (WDRC), full dynamic range compression (FDRC), or enhanced dynamic range compression (EDRC) hearing aids. Typically, these devices provide more gain than linear hearing aids for soft sounds. In addition, the compression threshold occurs at a relatively low input level (typically below 60 dB SPL) and at a low to modest rate (i.e., low compression ratio below 3–4:1). Because gain reduction occurs at a low input level, the risk of satura-
tion distortion at high input/output is also minimized. One should note, however, that saturation distortion can still occur at very high inputs (>85 dB SPL). Thus, some high-end programmable and digital WDRC hearing aids also incorporate CL at its output stage to further minimize distortion.

The choice of attack time and release time on a WDRC hearing aid is critical. If one designs a WDRC hearing aid to model after the physiology of the cochlea, the attack time and release time should be short because physiologic data of the outer hair cell (OHC) functions and basilar membrane motion suggest a very fast, active compressive mechanism in the healthy cochlea (Brownell et al, 1985; Ruggero and Rich, 1991). Furthermore, studies comparing the growth of loudness perception of short versus long sounds also suggest a relatively fast compressive mechanism (Buus, 1999).

A relatively short attack time and short release time (or fast-acting compression) are also logical if the purpose of the WDRC is to provide automatic gain adjustment to follow closely the intensity variation within the speech signal. For example, a compression hearing aid designed to follow the intensity variation at a phoneme level should use release time that is less than 80 msec (Plomp, 1983). Finally, a short attack time and a short release time are necessary to achieve the stated static compression ratio on the input-output curves (Kuk, 1998b). To provide the extra audibility for soft speech sounds, these hearing aids typically use a short attack time (less than 10 msec) and a short release time (less than 100 msec).

Research data on the efficacy of WDRC have been mixed. Braida et al (1979), based on a review of research data from 1952 to 1976, concluded that there was no conclusive evidence to suggest this type of compression is beneficial for patients with sensorineural hearing loss. Walker and Dillon (1982), based on a review of the literature up to that time, reported that there may be some “tentative support” for WDRC. Hickson (1994), based on a review of later research findings, reported more positively on the efficacy of such a circuit design. Dillon (1996) surveyed the literature up to that time and concluded that WDRC improved speech recognition in quiet over a linear hearing aid at a low input level and when subjects were not allowed to adjust the VC on the hearing aids.

The difference in performance between WDRC and linear amplification may be explained from the standpoint of audibility. Figure 5–2 compares the amount of gain prescribed for an impaired ear between a WDRC hearing aid (dotted line) and a linear hearing aid (solid line). To restore “normal” loudness, the WDRC hearing aid supplies more gain at lower input levels and less gain at higher input levels when gain at a conversational level (65 dB SPL) is similar for both hearing aids. If no volume adjustment were allowed, the higher gain from the WDRC hearing aid at a low input level would ensure better audibility, and thus higher speech recognition scores. Because less gain is available to a WDRC hearing aid than to a linear hearing aid at high input levels, a wearer of such device may find loud sounds more bearable with WDRC. This may ac-

![Figure 5–2](image_url). Hypothetical input/output curves comparing the difference in output between a linear hearing aid (solid line) and a wide dynamic range compression (WDRC) hearing aid (dotted line) when both are adjusted to have the same output to a conversational input (65 dB SPL).
count for the increased comfort associated with the use of WDRC hearing aids in daily environments (e.g., Moore et al, 1992).

**Automatic Volume Control (AVC)—**

Moderate CT (60–75 dB SPL), Low-Moderate CR (4–6:1), Long Attack (>20 msec) and Release (>1 s)—

**Preserving the Input Range Within the Wearer’s Residual Dynamic Range**

One negative consequence of providing more gain to low input sounds and less gain to high input sounds at a fast rate is a reduction of the intensity contrasts inherent in the speech signal. This effect is termed *smearing*. Its effect can be examined in both the temporal domain and the spectral domain. This could affect intelligibility, especially of intelligibility of speech in noise. It would be desirable to receive the audibility and comfort benefits offered by a WDRC hearing aid without altering the intensity relationship of the input signals. This can be achieved through the use of a longer attack time and a longer release time.

**Effective Compression Ratio**

The static or nominal compression ratio that is determined by measuring the steady-state response of the compression circuit to sinusoids (i.e., nominal CR) is only achievable when very short attack times (less than 5 msec) and release times (less than 30 msec) are used. The effective (or real-life) compression ratio is affected by factors that include the precompressor gain on the hearing aid, the peak-to-valley ratio of the input signals, the overall input level, the time constants of the compression circuit, and the intervals between different acoustic peaks (Blesser, 1969; Stone and Moore, 1992). Because daily acoustic signals (including speech) fluctuate greatly in intensity levels, peak-to-valley ratios, and interstimulus intervals, such factors can interact with the time constants of the compression hearing aid to yield a lower effective compression ratio than the nominal value. Kuk (1996, 1998a) provides a quantitative illustration of how the release time changes the nominal compression ratio. The longer the release time (for the same fixed intersyllabic interval), the more likely the effective compression ratio regresses toward one (1:1, or linear).

**Subjective Preference for Longer Release Time**

The “linearization” of compression hearing aids with a long release time raises potential conflicts with the rationale for WDRC. However, subjective data on the effect of release times seem to support their use. Neuman et al (1995) showed that hearing-impaired subjects preferred a longer release time (1000 msec) in some noise backgrounds in a single channel compression system. Hansen (2001) also reported that both normal hearing and hearing-impaired subjects preferred a longer release time (4000 vs. 40 msec) on a 1/3-octave band multichannel compression hearing aid. Verschuure et al (1996) also showed similar findings. If the supposition that a longer release time could result in a lower effective compression ratio were true, these studies suggest that hearing-impaired people prefer a lower compression ratio than a higher one.

It is of interest to note that Neuman et al (1994) reported that hearing-impaired subjects preferred compression hearing aids with a lower compression ratio (1.3) than a higher one (>3).

**Need for the Nominal or Static Compression Ratio**

The concept of the effective compression ratio raises the question of whether the amount of compression is adequate for the wearer if a longer release time were used. After all, the static CR is estimated to “compress” the range of acoustic stimuli into the residual dynamic range (DR) of the hearing-impaired patient. For example, if one assumes that the range of input is 100 dB and the hearing-impaired patient has a dynamic range of 50 dB at all frequencies, a CR of at least 2:1 is necessary. A compression ratio
smaller than 2:1 would mean some of the input would be beyond the wearer’s DR. A smaller CR is only acceptable (in theory) if the range of sounds in the environment is equal to or smaller than the residual DR of the hearing-impaired wearer.

Although environmental sounds can range from a very low level (lower than 0 dB SPL) to a very high level (over 100 dB SPL), the range of intensity in any single listening environment is smaller than the assumed 100 dB SPL. Indeed, Pearsons et al (1977) measured the range of intensity variations in many listening environments (inside a home, in a department store, etc.) and for the same persons speaking at different vocal efforts. They showed that the intensity range was typically 30 to 40 dB in almost all of the measured conditions. This suggests that rather than assuming a 100-dB range of input levels, a more typical range should be 30 dB in any listening environments. Other researchers also advocated similar ranges (e.g., Boothroyd et al, 1988; Skinner, 1988).

The residual DR of the hearing impaired patient is dependent on the wearer’s degree of hearing loss and the individual’s tolerance limit, i.e., UCL. Although individual differences exist, Pascoe (1988) showed that on average, UCL increased as the hearing loss increased. Although the residual DR decreased with increasing hearing losses, the residual DR was typically greater than 30 dB for hearing losses smaller than 80 to 90 dB HL. Figure 5–3 shows the relationship between thresholds, most comfortable listening levels (MCLs), and UCLs.

These two pieces of evidence suggest that for the average patient with less than a 80 to 90 dB hearing loss, wide dynamic range compression, while capable of achieving extra audibility for soft sounds and comfort for loud sounds, is not necessary to transpose the range of environmental sounds into the patient’s residual dynamic range. This is because the residual DR is typically larger than the range of acoustic inputs in a single listening environment. Although AGC is necessary to achieve audibility of the softer sounds (e.g., less than 50 dB SPL) and comfort for the louder sounds (e.g., higher than 75 to 80 dB SPL) without VC adjustment, it may not be necessary to use fast attack and release times in order to preserve the nominal or static compression ratio. Indeed, it may not even be subjectively desirable (Neuman et al, 1994, 1995). Once the overall gain level of the hearing aids has been automatically adjusted based on the input level, it can actually remain in a “linear processing” mode so that the intensity range of sounds from the specific listening environment is maintained. Kuk (1998b) provides a detailed explanation for the rationale of using long attack and release times in WDRC hearing aids.

Compression hearing aids with a long attack time and a long release time have been called automatic volume control (AVC) hearing aids in the past. These hearing aids have a moderately high CT (around 60 to 70 dB SPL) and a moderate compression ratio (less than 5:1). Even though the release times are classified as long, many are around 1 second in duration. Recently, WDRC hearing aids
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with extra long attack and release times have also become available. To distinguish from the traditional AVC hearing aids, which typically use an intermediate compression threshold, these hearing aids are called slow-acting wide (or enhanced) dynamic range compression (SA-WDRC) hearing aids to reflect the lower CT in these hearing aids.

Relatively few studies were conducted to examine the efficacy of AVC hearing aids. Those that did, supported the use of AVC, especially at a high noise input (King and Martin, 1984; Neuman et al, 1994). Because fast-acting multichannel compression systems introduce temporal and spectral smearing (Boothroyd et al, 1988; Van Tasell, 1993) and patients with a hearing loss need consistent audibility and as many speech cues as possible over a wide range of input without adjusting the VC, Byrne (1996) suggested the use of slow-acting compression instead of fast-acting WDRC to minimize any potential smearing while allowing for automatic gain adjustment.

Results of Recent Advances in Nonlinear Technology

More Complex Input-Output Curves

The use of digital signal processing (DSP) in hearing aids allows the implementation of various forms of I-O functions in current nonlinear hearing aids. Although many DSP hearing aids still use I-O curves that suggest basic functions, many use advanced algorithms that result in complex I-O curves. They have several functional units and may include many that were described in the previous sections. One example of a complex I-O curve is discussed below. There may be other complex I-O curves as well.

Figure 5–4 shows the I-O curve at the 2000-Hz channel of a 15-channel DSP hearing aid. Instead of having the linear-compression-peak clipping segments that were described in Figure 5–1a, this I-O curve has five segments, each with a different compression ratio and each serving a different function. From an input level of 0 to 2.5 dB SPL, the output of the hearing aid increases from 42.5 to 47.5 dB SPL. This represents a gain change from 42.5 dB at an input of 0 dB SPL to a gain of 45 dB at an input of 2.5 dB SPL. A region of the I-O curve that shows an increase in gain as input level increases is termed an expansion region. In this case, the region from 0 to 2.5 dB SPL illustrates expansion. If one uses Walker and Dillon's (1982) approach to calculate the compression ratio in this region (change in input range divided by the change in output range), one obtains a compression ratio of 2.5/5 or 1.2. In other words, a compression ratio less than 1 describes an...
expansion circuit. On the I-O curve, an expansion circuit also has a slope (change in output divided by change in input) that is greater than 1. Although an expansion circuit can be achieved using analog means, currently only DSP hearing aids utilize such circuitry.

As one moves from 2.5 to 7.5 dB SPL, one notes that the output changes from 47.5 to 52.5 dB SPL. The sameness in input range (5 dB) and output range (5 dB) defines linear processing. This region is included to provide a smooth gain transition as input increases from 2.5 to 7.5 dB SPL.

As the input increases from 7.5 to 52 dB SPL (average conversational level at 2000 Hz), the output increases from 52 to 70 dB SPL. Because the change in input range (44.5 dB SPL) is greater than the change in output range (18 dB SPL), compression is indicated in this input region. In this case, the compression ratio is 2.5:1. This is called low-level compression (LLC) to designate the fact that the CT is significantly lower than conventional CT (at around 40 dB SPL or higher), and the fact that the same CR is applied only up to the conversational level.

As one increases the input from 52 to 90 dB SPL, output increases from 70 to 100 dB SPL. This translates to a compression ratio of 1.3:1. This is called high-level compression (HLC) to distinguish from compression limiting, which typically starts at a high input level above 60 to 70 dB SPL. In this case, HLC starts at the conversational level and continues until a high input level is reached. The compression ratio is closer to 1 and is smaller than the compression ratio for the LLC. This is to approximate the freedom from recruitment seen in normal loudness growth functions at high input levels (Hellman and Meiselman, 1990) and to preserve the intensity contrasts in the high input signals.

The output above an input of 90 dB SPL does not change significantly as the input increases. Indeed, the output increases to 101 dB as the input increased above 100 dB SPL. This corresponds to a CR of 10:1. This is the compression limiting (CL) circuit and it serves the purpose of minimizing saturation distortion while maintaining comfort.

The complex I-O curve in Figure 5–4 reveals several advances in nonlinear technology over the simpler designs. First, a compression threshold as low as 7.5 dB SPL is possible. This is in contrast to the typical 40 dB SPL seen in conventional designs. Second, an expansion region is available below the compression threshold. Typically in conventional design, a linear circuit is used instead. Third, the range of compression (from CT to compression limiting) is significantly increased. In this case, it is 82.5 dB SPL (from 7.5 to 90 dB SPL) instead of the 40 (from 50 to 90 dB SPL) to 50 (from 40 to 90 dB SPL) dB SPL range seen in the conventional compression hearing aids. Fourth, more than one compression ratio, instead of one fixed CR, is seen in the compression range.

The advantages of having more complex I-O characteristics can be evidenced in at least three areas, as described in the following subsections.

**Audibility with minimal increase in circuit noise—Expansion**

The rationale for expansion (or gain increases as input increases) is to minimize the amount of amplification for sounds below the compression threshold. Typically, compression hearing aids utilize linear processing below the compression threshold. Inputs below the CT will be amplified maximally and may become audible to wearers who have normal hearing or a mild hearing loss.

Figure 5–5 illustrates the effect of an expansion circuit on the audibility of circuit noise. In most hearing aids, the audible circuit noise originates from the microphone that is used. Because the microphone is placed at the front end of the amplifier circuit, any noise that it generates will be amplified by the gain of the hearing aid. If the microphone has a noise level of 20 dB SPL, a hearing aid with 40 dB gain at and below the CT will amplify the noise level to 60 dB SPL. On the other hand, if a 1.2 expansion circuit
is used below the CT, a gain of 20 dB will be at the 20 dB SPL input instead. The amplified microphone noise will be 40 dB SPL instead. This represents a 20-dB reduction of microphone noise level. Depending on the hearing acuity of the hearing aid wearer, this noise level may not be audible.

Because an expansion circuit allows the audibility of soft sounds without the associated circuit noise that may be common with linear processing, an associated advantage of expansion circuit is to allow hearing aid manufacturers to lower the CT further with minimal artifacts from circuit noise. As microphone noise level decreases in the future, one may be able to achieve a lower CT with or without an expansion circuit.

**Figure 5–5.** Two input-output curves that differ in the type of processing below the compression threshold. The linear circuit is expected to yield an output of 60 dB SPL for a microphone noise level of 20 dB SPL. In contrast, an output of 40 dB SPL is expected of the expansion circuit.

Nonlinear WDRC hearing aids are fine-tuned differently than linear hearing aids. Because linear hearing aids provide the same gain at all input levels, adjustment of the gain parameter would affect sounds at all input levels. In contrast, nonlinear compression hearing aids offer gain adjustment in at least three input levels separately: soft, normal (or conversational), and loud. The presence of these parameters allows fine-tuning at specific input levels and minimizes the occurrence of side effects. Kuk (2001a) provides a summary of examples of how these parameters may affect subjective preference and how subjective comments can be related to electroacoustic parameters on a three-channel compression hearing aid.

For example, the I-O curve shown in Figure 5–6 illustrates the specificity of adjustment offered by current multichannel compression hearing aids. Figure 5–6a shows the effect of adjusting the gain parameter for low-level inputs. As the gain parameter is increased (dotted line), the effect is seen at the very low input level. The effect decreases as input increases toward the conversational level. This suggests that adjustment of this parameter will affect only soft sounds, and will not affect conversational and loud sounds. Figure 5–6b shows the effect of adjusting the gain parameter for loud sounds. Its effect is primarily restricted to input levels above the conversational level. Its effect increases as input level increases. Figure 5–6c shows the effect of adjusting the gain parameter for normal conversational input. In contrast to the parameter for soft and loud sounds, the effect is different depending on the extent of adjustment. As the gain parameter is increased or decreased slightly, its effect is concentrated mostly around the conversational input levels. Gain at the higher and lower input levels remained the same or decreased (dashed line). As the parameter is adjusted further, however, gain across the whole input range is decreased (dotted line) or increased (not shown). Such diverse adjustment and increased specificity represents the complex processing used in some of the current multichannel compression hearing aids.

**Figure 5–6.** Specificity of adjustment offered by current multichannel compression hearing aids.
Figure 5–6. Effect of adjusting various gain parameters on the output. (a) Effect of gain parameter for soft sounds. (b) Effect of gain parameter for loud sounds. (c) Effect of gain parameter for conversational sounds.
that a lower CT can be achieved with minimal artifacts like whistling as a result of feedback. Kuk (1999) describes the rationale for current DSP hearing aids and how they can achieve a low CT despite potential artifacts.

There are several advantages to a low CT. One immediate advantage is an increase in gain (and thus output) at inputs below the compression threshold over the same circuit with a higher CT. Indeed, Kuk (1998a) showed that a lower CT is associated with higher gain for inputs below the CT. This would mean a lower (or better) aided threshold in sound-field measurement (Kuk et al, 2001) as well as improved speech recognition scores for low (50 dB SPL) input presentation (Gabbard and O’Grady, 2000; Kuk et al, 2001).

Lee et al (1998) also demonstrated that a lower CT may overcome some of the limitations of hearing aids with directional microphones. These investigators studied the limitation in audibility of directional microphones to sounds presented from the back and the speech in noise improvement offered by them. A total of 30 subjects with a moderate degree of sensorineural hearing loss participated. In the first study, subjects were presented with soft speech (50 dB SPL) in quiet at 180 degrees. They wore a digital hearing aid with a fixed directional microphone and a programmable hearing aid with switchable directional microphones—omnidirectional and directional modes of operation. When the programmable hearing aid was worn in the directional mode, a mean score of 13% correct was achieved. When it was worn in the omnidirectional mode, a mean score of 37% correct was achieved. A mean difference of 24% was seen in the speech score. In contrast, a mean score of 44% was achieved with the digital hearing aid in the fixed directional mode under the same test condition. The investigators attributed the difference in speech scores to the compression threshold used in the hearing aids (50 to 55 dB SPL in the programmable hearing aid versus 20 dB HL in the digital hearing aid).

Some may question that the enhanced audibility of speech presented from the back could reduce the effectiveness of the directional microphone. In a second part of the study, the effectiveness of the directional microphones was compared. In this case, speech at 65 dB SPL was presented in the front and a party noise was presented in the back at a signal-to-noise ratio (SNR) of −5 dB. Rather than scoring lower with the digital directional hearing aid, subjects obtained similar scores between the digital fixed directional hearing aid (with a low CT at 20 dB HL) and the programmable hearing aid in its directional mode (53% for the digital and 45% for the programmable). This confirmed that the digital directional hearing aid was at least similar to, if not better than, the programmable hearing aid in its speech-in-noise improvement while minimally affecting audibility of soft sounds presented from the back. The difference in the presentation levels (50 dB SPL for soft speech and 70 dB SPL for noise) and the spectra of the stimuli may account for such observations.

**How low should the CT be?**

If CT affects the audibility of soft sounds, a logical question to ask is how low should the CT be to ensure best audibility. In theory, if the purpose of a WDRC hearing aid is to enable patients with a hearing loss to hear like a normal hearing person at 0 dB HL, then the ideal CT should be at 0 dB HL. The degree of hearing loss or the configuration of the hearing loss should not be factors to consider. This is also the position argued by Dillon (1996) and Cornelisse et al (1995). There are, however, practical factors to suggest otherwise.

**Microphone Noise Floor**

Because the microphone is the first active component in a hearing aid, any noise from it will be amplified by the gain on the hearing aid to become audible. This suggests that patients with a mild degree of hearing loss may hear the thermal noise from the microphone and react negatively. Thus, a low CT is acceptable only if the microphone noise will not be amplified above the wearer’s thresholds.
There are two potential solutions to this problem. First, if the noise floor of the microphone is reduced, then its amplified level may be below the wearer’s threshold and remains inaudible. The typical microphones today have an overall noise level of around 20 to 25 dB SPL, with the spectral level of around 10 to 15 dB SPL in the mid-frequencies. This places a lower limit for the CT of around 15 to 20 dB SPL in the mid-frequencies. Recent advances in microphone technology have reduced such noise level to an overall level of around 10 to 15 dB SPL and the spectral level under 5 dB SPL in the mid-frequencies (higher in the lows). This could potentially allow for a lower CT in modern nonlinear compression hearing aids.

Another approach to achieve a lower CT with minimal circuit noise is the use of an expansion circuit. This has the effect of providing less gain to microphone noise to result in a lower, inaudible output. Obviously, the need for an expansion circuit is more critical when the amplifier is used with a microphone that has a high noise floor. The need for such circuit may lessen with the newer, quieter microphones. The limitation of the microphone noise floor may no longer be an issue that prohibits the use of a low CT.

Despite the practical ability to achieve a CT below 10 dB SPL, one has to question the practical value of having such a low CT. Clearly, the lower the CT, the lower the input level one can hear. If the purpose of a nonlinear hearing aid is to allow the perception of normal speech, then the required CT would not need to be lower than 20 dB HL across frequencies. This is because the softest portion of normal speech is around 20 dB HL across frequencies when such data are displayed on an audiogram format (Byrne et al, 1994). However, if the goal is to hear the softest portion of soft speech, a CT that is significantly lower than 20 dB HL should be desirable.

Although the actual value for a CT has not been agreed upon, the general consensus is that compression threshold should be as low as possible (Sweeton, 1998). In choosing a compression hearing aid, it may be worthwhile to select one with the lowest CT and make adjustment to it if necessary according to the subjective comments of the patients after they have worn the hearing aid for some time (Kuk, 2001a).

Not everyone favors a low CT

Although a low CT has the advantage of providing increased audibility, the extra audibility may not always be desirable. Barker and Dillon (1999) studied subject preference for a single channel, 2:1, fast-acting WDRC hearing aid that allowed variation of the CT. Subjects wore the hearing aid at a CT of 40 dB SPL and a CT of 65 dB SPL for a total period of 1 month before indicating their preference. Linear processing was used below the CT. The results of the study showed that the majority of the subjects preferred the hearing aid set at the higher CT (65 dB SPL) than at the lower CT (40 dB SPL). Distraction from the ambient noise associated with the lower CT was reported as the major complaint. It is conceivable that the instructions given to the subjects, the type of hearing aid used (single channel, fast-acting versus slow-acting, multichannel), and the type of processing below the CT (linear versus expansion) contributed to the outcome of the study. Nonetheless, it is clear that, although a low CT can be achieved, how it is implemented on a compression hearing aid can directly influence its acceptance. It is not sufficient to conclude on the merit of a feature based on its theoretical considerations; it should be considered in the context of the whole hearing aid design.

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1This is expressed in dB HL and not dB SPL. dB HL and dB SPL are linked by the minimal audible field (MAF) or minimal audible pressure (MAP) functions. In essence, 0 dB HL is the dB SPL needed to reach a threshold sensation. It is also expressed as the audiometric zero. Because the sound pressure level needed to reach threshold varies with frequencies, expressing the needed CT in dB HL has the convenience of the same value across frequencies and not discrete value for each frequency.
More Specificity in Frequency Processing with More Processing Channels

Another major change in modern nonlinear hearing aids is the availability of more independent compression channels. Today's multichannel nonlinear hearing aids vary from as few as two channels to as many as 20 channels. This has the advantage of increasing the specificity of processing in the frequency domain.

Differences between Bands and Channels

A band in a hearing aid refers to a frequency region where gain adjustment is made. A close analogy of the “band” concept is the equalizer where one can make gain (or sensitivity) adjustment in many frequency regions (or bands). The nature of the signal processing (e.g., compression ratio) of all the bands, however, remains the same.

A channel in a hearing aid refers to a frequency region where the same signal processing takes place. The relationship between “bands” and “channels” is illustrated in Figure 5–7, which shows a hypothetical hearing aid that has 10 frequency regions where gain adjustment can be made (i.e., bands) and three broader frequency regions where different ratios of compression can be applied (i.e., channels). In this case, bands 1 to 4 contribute to the “low” channel, bands 5 to 7 to the “mid” channel, and bands 8 to 10 to the “high” channel. Although gain in each of the four bands in the “low” channel may be quite different, the change in gain (or compression ratio) of all four bands is identical because they are in the same channel. This being the case, one may conclude that the number of channels in a hearing aid is equal to or smaller than the number of bands in it.

Achieving desired output/gain with atypical hearing loss configurations

Because gain provided by a processing channel is the same for all inclusive frequencies, the need for more channels increases as the hearing loss configuration becomes more irregular (e.g., an “inverted V” graph). This is necessary so that each frequency region may receive independent gain adjustment for speech to be delivered close to the optimum level. This goal can be achieved with a multiband system as well.

Recruitment compensation

Because most patients with a hearing loss exhibit different rates of loudness growth at different frequencies, different rates of gain adjustment (i.e., compression ratio) will be necessary for different frequency regions to restore normal loudness. Consequently, the requirement for independent multiple processing channels increases as the hearing loss configuration departs from a flat configuration. In theory, the more threshold changes across frequencies, the greater is the need for more independent processing channels.

Monitoring one’s own voice

When a person speaks, his/her voice measured at the ears has a different spectrum than when it is measured at a conversational distance of 1 m. It is about 15 to 20 dB SPL higher in the low-to-middle frequency region and about 5 dB SPL lower in the high frequency region than that measured at a conversational distance. This is due to a combination of distance factor and radiation characteristics of vocal production (Dunn...
If the patient wears a single channel linear hearing aid, the excessive low frequency from the patient's own voice could result in upward spread of masking on the high-frequency consonants and render them inaudible. However, if the speaker wears a single channel nonlinear hearing aid, the excessive low-frequency input would reduce the gain of the hearing aid across the whole spectrum. Although the threat of upward spread of masking is minimized, the gain of the high-frequency sounds is also reduced. The end result in both cases is that the high-frequency sounds may be inaudible. If speech production is a function of how well the speaker monitors his/her own voice, a single channel device may compromise the speaker's speech production skill.

A multiple channel hearing aid is useful in that different frequency regions can be amplified using different compression settings. For example, one may maintain the same gain for both the high- and low-frequency regions. However, one may increase the compression ratio (even more gain reduction at high inputs) for the low-frequency region and decrease the compression ratio in the high-frequency region to minimize the spread of masking of the low-frequency input signal on the high-frequency input signal, and to provide separate processing between the two frequency regions. The ability of a multiple channel hearing aid in allowing the wearers to monitor their own voice may be one of the reasons why improvement in speech production skills was seen in children wearing a three-channel digital hearing aid (e.g., Kuk et al., 1999c).

occlusion management
Along the same line, the higher input of the speaker's voice to his/her own hearing aids would also result in a higher output in the person's ear canal. This could contribute to the perception of "hollowness" or "barrel effect" that is commonly known as the occlusion effect. In addition to many other approaches like venting and increasing the canal length of the hearing aid, one common approach to minimize this effect is to reduce the amount of low-frequency gain (Kuk, 1990; Sweetow and Valla, 1997).

Gain reduction in the low frequency, although effective in minimizing the occlusion perception, may lead to undesirable results if one fails to control the precise frequencies at which gain reduction occurs. Fletcher (1953) reported on the relative contribution of each frequency region to overall "intelligibility" and "overall loudness." He showed that frequencies below 500 Hz contribute substantially (60%) to overall loudness perception (or the "boomy" sensation), whereas the same frequencies contribute less than 5% to overall intelligibility. That is, if one can reduce gain below 500 Hz successfully without affecting gain at higher frequencies, one could reduce the perception of hollowness with minimal effect on intelligibility. The difficulty is the ease of reducing gain below 500 Hz without affecting gain above that frequency region.

In a single channel system, reducing gain below 500 Hz may also reduce gain above 500 Hz. Indeed, the more gain reduction below 500 Hz, the more gain reduction at the nearby mid-frequencies. This is also a problem with a dual-channel system that uses a fixed crossover frequency. For example, if the crossover frequency between the two channels is fixed at 1500 Hz, any gain change in the low-frequency channel would affect frequencies up to 1500 Hz. The frequency region between 500 and 1500 Hz, however, contributes as much as 55% to speech intelligibility and 36% to overall loudness. By lowering gain in the low-frequency channel, one would have minimized the hollowness perception, but speech intelligibility may suffer also. The ideal solution is to have enough independent channels so that gain modification at 500 Hz and below does not affect gain in the nearby mid-frequencies. Alternatively, movable crossover frequencies may be used to vary the bandwidths of each processing channel.

noise reduction algorithm—analysis and implementation
Many DSP hearing aids today include noise reduction algorithms that are capable of (1)
identifying “speech” and “noise”, and (2) differentially amplifying “speech” and “noise.” Although there are many different algorithms that can achieve these goals, their effectiveness depends in part on how well they can distinctly identify speech and noise signals, and how well noise is attenuated without affecting speech. The number of channels in the DSP hearing aid sets a limit on its effectiveness.

The number of channels in the hearing aid can limit the resolution of the noise analysis and its treatment. In general, increasing the number of channels (thus narrowing bandwidth) increases the precision of the analysis. When a broad bandwidth is used (fewer filters), details of the speech signal may be missed. When a narrower filter is used (many filters), details of the signal would be retained for a more precise analysis.

The precision of the speech/noise treatment can be improved by increasing the number of independent processing channels. Imagine the same dual-channel hearing aid with a crossover frequency at 1500 Hz. Let’s assume that a narrow band of noise that is centered at 500 Hz occurs at the same time as a female talker who has a peak in her speech spectrum around 1000 Hz. The result of the analysis calls for a gain reduction in the low-frequency channel. However, the magnitude of the 500 Hz noise and the desirable speech around 1000 Hz will be attenuated. The result is a diminution of noise as well as speech. Thus the SNR may not improve.

However, in a 15-channel system, only the 500-Hz channel would incur gain reduction. This will spare the speech information at 1000 Hz. In addition, the overall loudness of the output is lowered because of gain reduction at 500 Hz. This would increase listening comfort while preserving intelligibility. For individuals who may be easily affected by the spread of masking, this may even improve their speech recognition ability as the spread of masking is minimized through gain reduction.

Despite the many potential advantages offered by multiple channel hearing aids, one inherent limitation of having many independent channels is the increased risk of temporal and spectral smearing. Such risks increase dramatically as the number of independent channels increases, and especially in a fast-acting (i.e., short attack time and release time) WDRC hearing aid.

Figure 5–8 shows the effect of the number of channels on the spectral level of the vowel /i/. The formants of the vowels are indicated by the peaks in the figure. The intensity difference between the formants and between the peak and the “trough” serves as cues for speech identification. As one can easily see, a two-channel compression system causes a small reduction of the spectral difference. As the number of channels increases to 10, however, significantly more reduction of peak-to-trough difference is seen. That is, spectral smearing increases as the number of channels increases.

Figure 5–9 shows the corresponding smearing (or contrast reduction) in the temporal or intensity domain. Again, the intensity difference between the peak and the valley of the waveform (i.e., envelope) could serve as cues for speech understanding. In this case, the reduction (or smearing) worsens as the number of channels increases. Because patients with more than a moderate
degree of hearing loss rely more heavily on the temporal contrasts for speech understanding, such smearing could also affect speech understanding (Van Tasell et al, 1987).

Several studies have examined the effect of the number of compression channels on speech recognition. Many showed a decrease in speech recognition scores in the multichannel device or as the number of channels increases in the multichannel hearing aid (Bustamante and Braida, 1987; Plomp, 1988; Drullman et al, 1996). However, several authors (e.g., Mangold and Leijon, 1979; Moore et al, 1985; Yund and Buckles, 1995a–c) showed improvement in speech recognition scores with multichannel compression processing. For example, Yund and Buckles (1995b) showed that speech recognition score improved significantly (48% vs. 52%) as the number of channels in a compression system increased from four to eight. Interestingly, the same authors (Yund and Buckles, 1995c) in a related study showed that speech recognition for an eight-channel compression system was not significantly different from that for a linear hearing aid (50.4% vs. 51.6%). Clearly, the findings on the impact of the number of channels on speech recognition are mixed at this time. Although there may be theoretical advantages with using more channels, one may encounter diminishing returns as the number of channel increases. Worse yet, negative artifacts (e.g., smearing), may become more noticeable if careful consideration is not paid to the design process. The evaluation protocol, the subject population, and the design of the multichannel compression hearing aid (e.g., slope of filters, channel coupling, etc.) may lead one to different observations and conclusions.

More Sophisticated Noise Reduction Algorithms

Conventional compression hearing aids have been used for noise reduction purposes (Dillon, 1996). By definition, noise reduction hearing aids are those that are capable of changing their frequency gain characteristics as a result of the speech/noise analysis. In essence, noise reduction hearing aids are compression hearing aids that involve not only an overall gain change, but also a frequency response change. Examples of conventional compression hearing aids that serve as noise reduction hearing aids are the bass increase at low levels (BILL) and treble increase at low levels (TILL) circuits, which decrease the low-frequency gain (for BILL) and high-frequency gain (TILL) as input level increases (Killion et al, 1990). In these two ex-
Examples, the frequency response as well as
the overall gain of the hearing aid change as
the input level changes. The assumption for
“noise” is the high overall input level.

Instead of using the overall level of the
input to signal gain reduction, many current
digital hearing aids use sophisticated algo-
rithms to estimate the nature of the input
signal. One approach that is used by several
commercial digital hearing aids is the use of
the modulation index of the input as a basis
for speech/noise estimation. The rationale
is that speech modulates greatly over its
course (e.g., Plomp, 1983) whereas a typical
noise does not vary greatly in its intensity
over time. This is seen on the left of Figure
5–10, which shows the waveform envelopes
of a speech segment (top left) and a car noise
(bottom left). It is clear that over the time pe-
riod sampled, the intensity of the speech
segment varied significantly, whereas the
noise did not show much variation. Thus,
speech has a high modulation index, whereas
noise has a low modulation index. By deter-
mining the modulation indices of the input
signals, one can determine if it is speech or
noise so that appropriate gain adjustment
may be made afterward. In general, gain ad-
justment (typically reduction) beyond that
from compression will be made in the fre-
quency channels where a noise is detected.
No gain changes beyond that from compres-
sion will be made if speech is detected.

An alternative approach to estimate the na-
ture of the input (i.e., speech or noise) is to de-
terminate its level-distribution function, which is
a plot of the percentage of time sounds within
a specific frequency channel occur at a partic-
ular intensity (Ludvigsen, 1997). For a stimu-
lus that does not vary greatly in its intensity

![Figure 5-10](image_url)

**Figure 5-10.** Examples of waveforms showing high modulation (speech, top left) and
low modulation (car noise, bottom left). On the right are the corresponding level distribu-
tion functions of the same listening environments. Note the bimodal distribution of speech
and unimodal distribution of car noise.
over time (i.e., noise), it will appear to have a unimodal distribution on a level distribution function. This is the same as saying that a particular intensity level has a high frequency of occurrence. By the same token, because a speech signal shows intensity fluctuation over time, its level distribution function will have a multimodal distribution to reflect that different intensity levels occur in the input or that significant fluctuations in intensity occurred. For the most part, the use of modulation index or level distribution function yields similar estimation of speech or noise. The right panel of Figure 5–10 shows the level distribution functions for the speech and noise inputs.

The use of the results of a statistical detector instead of a level detector (peak-to-peak or root-mean-square) to signal the onset of gain adjustment at different frequencies means that the static I-O curve is only partially reflective of the actual performance of the hearing aids in real life. More specifically, it is restricted only to speech conditions and not to noise conditions. In addition to a different real-life gain, the concept of compression threshold (onset of gain reduction) would need to be modified by the nature of the input level as well as its overall level. This has a significant impact on how such hearing aids are evaluated today, as will be discussed in the following sections.

**Need to Reconsider Current Verification Strategies with Nonlinear Hearing Aids**

The advances in multichannel nonlinear hearing aid technology raised issues on its optimal verification and validation. Specifically, because many of the evaluative tools that are used today have their origins from single channel linear hearing aids, difficulties may arise when they are applied directly to current multichannel nonlinear hearing aids. The distinction between validation and verification

Validation involves an assessment of the wearer’s performance with the hearing aids. For example, speech recognition tests, daily questionnaires, and subjective quality judgment in the form of either category rating or paired comparison are validation tools. The sensitivity and reliability of these validation tools to evaluate today’s multichannel nonlinear hearing aids, although an important consideration, will not be discussed in this chapter. The reader is referred to Chapter 5 of the companion text, Strategies for Selecting and Verifying Hearing Aid Fittings, 2nd edition (Thieme Medical Publishers) for a detailed description of these tools.

Verification refers to the physical measurement of the output of a hearing aid. The goal is to determine if it matches some predefined standards. An example of verification is measuring the electroacoustic output of a hearing aid to determine if it agrees with the data on the specification sheet. Other tools or indices of verification include measuring coupler output to determine if it matches a prescriptive output/gain target, or measuring the real-ear output of a hearing aid to determine if it matches a specific target for gain/output. Some also include the determination of the aided threshold as a verification metric. Regardless of where the measurement is performed (i.e., real ear or coupler), a decision on the suitability of the hearing aid is made as a consequence. Adjustment will be made to those hearing aids to meet the criteria. Verification tools do not yield information on the wearer’s performance with the recommended hearing aid settings. Many, however, assume the contrary and expect satisfaction when the verification criteria are met. The reader is referred to Chapters 2 and 3 of the companion text, Strategies for Selecting and Verifying Hearing Aid Fittings, 2nd edition (Thieme Medical Publishers) for a detailed description of these tools.

The availability of current nonlinear hearing aids has created two areas of concern about verification efforts: the ability to reliably measure the output of such hearing aids, and the interpretation on the adequacy of the output relative to prescriptive targets.

**Hearing Aid Factors That Affect Reliability of Measure**

Reliability of the measure, whether matching the output to specification data according to ANSI protocol (e.g., ANSI, 1996), or match-
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ing it to a prescriptive target using either coupler measures or real-ear approaches, can be affected by the characteristics of the hearing aid processing. Although many nonlinear hearing aids use digital signal processing, not all such hearing aids create the same difficulties in measurement because of differences in their processing features. Special precautions are needed to account for nonlinear processing. The special processing is addressed in the following subsections.

noise reduction algorithms

Many current multichannel nonlinear hearing aids include noise reduction algorithms that use statistics to estimate the nature of the incoming signal to determine if additional gain adjustment across frequencies is necessary. As was discussed earlier, many use either the modulation index or the level distribution function to make such decisions. Although the basis of the speech/noise decision is fairly similar among manufacturers, the aids do differ at the input level at which noise reduction is active (high input level only or all input levels), the maximum amount of gain adjustment across frequencies, as well as the speed at which the time-varying filter changes its frequency response characteristics.

The types of stimuli that are routinely used for verification purposes (real ear or coupler) could be problematic with these current hearing aids. This is because the typical stimuli, be they sinusoids or broadband noise, are stationary in nature. That is, their intensity level does not change during the course of the stimulus presentation. Although any of these signals would be acceptable for conventional hearing aids, all of these signals would be regarded as “noise” by current nonlinear hearing aids that include noise reduction algorithms. Figure 5–11 shows the gain of a three-channel noise reduction hearing aid to a broadband noise with the noise reduction algorithm deactivated and activated. As much as 12-dB gain reduction is seen when the noise reduction algorithm is activated. See Chapter 3 in the companion text, Strategies for Selecting and Verifying Hearing Aid Fittings, 2nd edition (Thieme Medical Publishers) for a detailed description of issues related to real-ear measures of DSP hearing aids.

The amount of gain reduction varies across manufacturers. In addition, the SNR of the input as well as the duration of the stimulus could affect the measured gain level. In general, more gain reduction is as-

Figure 5–11. Frequency-gain response of a digital hearing aid to a broadband speech noise when (a) the noise reduction algorithm is not active, and (b) the noise reduction algorithm is active.
associated with inputs that have lower SNRs. The duration of the stimulus interacts with the estimation time used by the noise reduction algorithm to yield varying output. If the stimulus duration is significantly shorter than the noise estimation time, no gain adjustment is made because the noise reduction algorithm has insufficient time to make an estimation. If the duration of the stimulus is about equal to the estimation time, gain reduction occurs. Gain reduction continues to the maximum for the specific SNR as long as the stimulus is present. For example, a three-channel digital hearing aid that the author has experience with takes approximately 10 seconds to be fully activated. The potential effect of the noise reduction algorithm will not be seen if a stimulus is presented for only 5 seconds. If, however, the stimulus is presented for 15 seconds, gain from the hearing aid could be reduced by as much as 12 dB.

This observation suggests that one must control the duration of the stimulus presentation during verification efforts. Otherwise, the same hearing aid settings could result in different output. If the purpose is to examine the gain of the hearing aid without the action of the noise reduction algorithm, then the stimulus must be a broadband stimulus that is presented for a sufficiently long time to activate the compression circuit but not long enough to activate the noise reduction algorithm. The exact time would be dependent on the specific manufacturer. Typically, a stimulus that is about 3 to 5 seconds in duration is acceptable for many hearing aids. If, however, the purpose is to examine the gain with the noise reduction fully activated, stimulus duration of over 15 seconds will be necessary.

There are simpler methods to bypass the action of the noise reduction algorithm during verification. Most fitting software has the option to allow deactivation of the special features (e.g., noise reduction) during verification. This would allow reliable testing of the hearing aid output. Another option is to use a specially designed broadband stimulus, the International Collegium of Rehabilitation Audiology (ICRA) signal for testing. The ICRA signal is a speech-shaped noise that is randomly modulated so it appears as speech to current hearing aids with noise reduction algorithms (see Chapter 1). This would prevent the hearing aid from lowering its gain and avoid variation of the output even with different stimulus duration. Several manufacturers of testing equipment also included this stimulus as an option.

attack and release time

The release time of nonlinear hearing aids may affect the reliability of the coupler measure (Revit, 1994). The problem is aggravated by some current nonlinear hearing aids that use even longer attack and release times (more than 2 seconds) than were considered by the manufacturers of hearing aid test systems. This is especially true when automatic test sequences are allowed. The output of the hearing aid varies depending on when the test stimulus is presented in relation to the recovery phase of the hearing aid. For example, Figure 5–12 shows the frequency gain curves at input levels from 50 to 80 dB SPL in 10-dB steps from a hearing aid that has a maximum release time of 20 seconds. Figure 5–12a shows that no gain difference is seen across the four input levels when the delay between stimulus presentation is 0.5 second. One may have concluded that the hearing aid is a mild gain linear hearing aid. Figure 5–12b shows the results with a 2-second delay between presentations, whereas Figure 5–12c shows the same at a delay of 5 seconds. Significant deviation in the gain of the hearing aid is seen across the three graphs. This shows that if the delay between presentations is shorter than the release time of the hearing aid, one may obtain variable results depending on when the subsequent stimulus is presented during the gain recovery phase of the nonlinear hearing aid. In general, the delay between stimulus presentations must be longer than the maximum release time of the hearing aid to minimize variability in output. For hearing aid test systems that provide automatic stimulus
presentation, it would be important to set the automatic test sequence so that sufficient delay between presentations is introduced.

Low compression threshold

The test protocol recommended by the ANSI working group to test the attack time and the release time of a nonlinear hearing aid is to use a stimulus that changes its intensity level from 55 to 90 dB SPL. The choice of such intensity levels assumes that the nonlinear hearing aids being evaluated are compression limiting or automatic volume control devices. These types of compression hearing aids typically have a high compres-
sion threshold (above 60 to 65 dB SPL). The use of this stimulus to measure the dynamic characteristics of a WDRC hearing aid may be questionable. This is because the threshold of WDRC hearing aids, even the conventional ones, is often below 55 dB SPL (from 40 to 55 dB SPL). This suggests that the hearing aid is in compression even for the 55 dB SPL portion of the test signal. This is even truer with current digital signal processing hearing aids where many have CTs below 30 to 40 dB SPL. A different signal level is needed if one wishes to characterize the dynamic aspects of current compression hearing aids with low CT (below 40 dB SPL). The issue is complicated even more when many of these multichannel devices have different dynamic characteristics and expansion processing in each channel. More than one test frequency may be needed in the future.

Multiple Channels

The choice of stimulus type (sinusoid or broadband noise) makes minimal difference on the obtained frequency-gain curves when testing a linear hearing aid at typical input levels. For example, Figure 5–13 shows a linear power hearing aid tested with a pure-tone sweep, an ANSI composite noise as implemented on the Frye 6500 test system (ANSI, 1992), and an ICRA noise implemented on the Frye 6500 system. Note that identical frequency response curves are obtained.

Testing nonlinear compression hearing aids, even single channel devices, requires special consideration. Preves (1996) cautioned about the artificial effects of “blooming” in the output frequency response when a sweep-tone stimulus is used as the test signal instead of a broadband noise. Because of such artifacts, it has been suggested that a stationary broadband noise with defined characteristics be used to measure the frequency response of nonlinear hearing aids (ANSI, 1992).

The availability of multichannel nonlinear hearing aids brings even more variability in the measured outcome when different stimuli are used. One such stimulus is the ICRA noise that was discussed previously. Figure 5–14a shows the gain of a three-channel WDRC hearing aid measured with a 60 dB SPL sweep-tone (ANSI, 1996). The same hearing aid is also tested with a 60 dB SPL composite noise specified by ANSI (1992) (Fig. 5–14b) and a 60 dB SPL ICRA noise (Fig. 5–14c). In addition to the blooming effect seen in Figure 5–14a (over Fig. 5–14b), significant difference in the gain values is observed between Figure 5–14b and 5–14c around the 1000-Hz region. This is because the ANSI noise and the ICRA noise differ in their frequency composition. In a multichannel hearing aid where each channel can have its own compression threshold and compression ratio, a difference in the input levels among frequencies would suggest different gain. This variability in output with stimulus types raises the issue of (in)accuracy of our target match across clinical sites or within the same site that uses more than one type of test signal as offered by some test systems. This calls for a more standardized noise for ease of data comparison across facilities and for verifying the output/gain of multichannel nonlinear hearing aids. At a minimum, the type of stimulus used in verification must be reported.

Factors Affecting Interpretation of Target Match

A criterion that is commonly used in the fitting of hearing aids (linear and nonlinear) is to determine if the output (and/or gain) of the hearing aid matches a prescribed target. This measurement can be obtained via probe-microphone measurement or through coupler measurement with appropriate real-ear-to-coupler corrections. It is often assumed that real-world satisfaction is guaranteed when the output (real ear or coupler) of the hearing aid meets a specific gain target. Although many clinicians recognize that strict target matching is impossible to achieve and may not even be desirable, there are additional reasons why one may need to be even more cautious when it comes to current multichannel nonlinear hearing aids. The author is not ignoring the many benefits provided.
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by prescriptive formulae. Rather, the point of this discussion is to challenge our current thinking so that suitable adjustment to our current practice can be made to reflect the state of technology.

Figure 5–13. Frequency gain curves of a linear power aid for different stimuli: (a) pure tone sweep; (b) Frye 6500 composite speech noise; (c) Frye 6500 International Collegium of Rehabilitative Audiology (ICRA) noise. Note the same frequency response curve in all situations.

assumptions of prescriptive targets

Most prescriptive formulae for linear hearing aids are based on the assumption of amplifying normal conversational speech (with an
overall level of 65 to 70 dB SPL, in quiet) to the wearer’s most comfortable listening (MCL) level. Suitable adjustment to the low- and high-frequency gain is made to minimize amplifying too much room noise (in the low frequency) and the occurrence of feedback in the high frequency (Skinner, 1988). Differences exist among various formulae from the difference in the assumption of the speech spectrum, as well as modification based on empir-
ical evidence. Regardless of the formula chosen, frequency gain at one input level (i.e., conversational level) is recommended for daily use in all listening environments.

The fitting philosophies for nonlinear hearing aids are more diverse. Some advocate the use of individual loudness scaling to set the individual compression parameters on the nonlinear hearing aids. Examples are the Independent Hearing Aid Fitting Forum (IHAFF) approach (Valente and Van Vliet, 1997) and the loudness growth by octave band (LGOB) approach (Allen et al, 1990). These approaches aim at restoring normal loudness in the impaired ear. Some are prescriptive approaches that use the ratio of the wearer’s residual dynamic range to the dynamic range of normal hearing listener to define the compression characteristics. An example is the desired sensation level (DSL[i/o]) approach formulated by Seewald and his colleagues (Cornelisse et al, 1995). Some take an eclectic approach to prescribe gain (and thus compression ratio) at different input levels (Fig6, 1996). Many of these approaches use a loudness normalization philosophy (Byrne, 1996). Still others prescribe gain by attempting to ensure equal loudness at each frequency region (e.g., National Acoustic Laboratories, 1999). See Chapter 1 in the companion text, Strategies for Selecting and Verifying Hearing Aid Fittings, 2nd edition (Thieme Medical Publishers) for a detailed description of each method. Unlike the prescriptive targets for linear hearing aids, frequency gain recommendations at three input levels are provided for a calculation of the desired compression threshold and ratio. But similar to the linear targets, all the targets assume listening in quiet and that the recommendation is applicable to all hearing aids (i.e., they are generic).

**should there be only a single target?**

To meet the prescriptive requirement, a linear hearing aid has to meet the target gain for conversational level ~65 db SPL only. It is assumed that the use of the volume control (VC) on the hearing aid would compensate for the overall level difference for the low and high input levels (Byrne, 1996). The assumption is that the frequency response requirement for low and high inputs is similar to the medium input.

Several studies that examined the preferred frequency responses under different listening conditions raised doubts about the assumption of a fixed frequency response for all input levels. Kuk et al (1994) measured the preferred frequency responses in hearing-impaired subjects across listening conditions that differed in the SNR while maintaining the same overall input level. Their results showed that most subjects preferred a broad frequency response when the SNR of the listening environment was positive. As the SNR became poorer, however, subjects preferred less low-frequency gain as well as lower overall gain from the hearing aid. This suggests that the preferred frequency gain on a hearing aid is determined not only by the overall input level but also by the SNR of the input signal.

Keidser et al (1995, 1996) also showed that patients with a hearing loss preferred frequency gain characteristics that are different from the National Acoustic Laboratory’s (revised) (NAL-R) (Byrne and Dillon, 1986) target in situations other than quiet. In situations where low-frequency noise or babble noise dominates, a steeper response than the NAL-R target is preferred. In situations where high-frequency noise is present, a flatter response than the NAL-R target is preferred. Because real-life listening situations of hearing aid wearers will necessarily include many types of noise, these studies question the limited value of having a fixed frequency-gain response for all environments. Fixed target at one level or the same frequency response target at multiple input levels may not be optimal. Targets must account for not only the input level effect but also the SNR effect in daily situations. That is certainly an impetus for the design of the so-called noise reduction hearing aids that respond not only to the overall input level, but also to the SNR of the listening environments.

Nonlinear fitting targets, especially those measured on an individual basis, have the appearance of higher face validity in that the level effect is considered. The effect of SNR,
However, the fitting procedures of these approaches require the patients with a hearing loss to perform loudness scaling in which their perceived loudness of discrete, stationary tone bursts or narrow band noise is used to define the static compression characteristics of a hearing aid (Kuk, 2000). Although loudness scaling is certainly one approach to study loudness perception, its use in fitting a nonlinear compression hearing aid, especially modern multichannel nonlinear hearing aids, may be questionable.

Loudness normalization is based on the premise that the manner of amplification would restore normal loudness perception in the patient with a hearing loss. Because of the difference in the degree of recruitment among frequency regions (other than perhaps a flat hearing loss configuration), compression hearing aids would necessarily require many independent processing channels so that each can perform independent processing for a restricted frequency region. A compression hearing aid that can truly restore loudness normalization would require as many channels as there are independent auditory processing units. Even if one approximates each critical band as one independent processing unit, it would require at least 15 to 20 independent channels in a nonlinear compression hearing aid to achieve loudness normalization at all frequencies. Many multichannel compression hearing aids do not have enough channels to qualify for this requirement.

Is loudness normalization critical?

The importance of loudness normalization was questioned by Byrne (1996), who argued that loudness normalization does not guarantee optimal speech intelligibility for the hearing aid wearer. Rather, he argued that the process of loudness normalization may have placed too much emphasis on the frequency-intensity domain and that undesirable side effects may occur to offset any potential advantages of loudness normalization. This may be the case when the nonlinear hearing aids use fast-acting compression in many channels. One such negative effect is the potential of smearing in the temporal and spectral domains. This could obliterate some of the cues available in the input for speech understanding (Van Tasell, 1993).

Byrne (1996) also questioned the “unimportance” of normal loudness relationship. He indicated that the long-term spectrum of speech and the short-term spectra of sounds vary considerably among talkers and listening environments. Yet, as long as they are audible and occur within the dynamic range of the listener, most people do not have any problem understanding them. This directly questions the need to maintain a “normal” loudness relationship as advocated by the loudness normalization fitting approaches. Other investigators also echoed similar opinions through independent studies (Horwitz et al, 1991; Van Tasell, 1993; Van Buuren et al, 1995). The opinions of these researchers suggest that strict loudness matching in a nonlinear hearing aid may not be critical to provide optimal results.

The effect of multiple channels has not been considered in the targets

Many of the generic prescriptive targets were generated based on theoretical constructs (e.g., DSL [i/o], Cornelisse et al, 1995) or empirical evidence derived from existing hearing aid wearers (e.g., NAL-RP [including profound loss], Byrne et al, 1990). It is reasonable to expect that the generated targets reflect the dominant hearing aid technology of the time. Given that assumption, it would follow that many of them are applicable only to single channel linear hearing aids. Issues related to multichannel compression hearing aids would probably not have been considered in the formulation of these targets.

Kuk and Ludvigsen (1999a) provide a detailed illustration on the output effect of the number of channels in a multichannel compression hearing aid. In summary, these investigators showed that two compression
hearing aids that have the same static I-O curve (i.e., same gain, compression threshold, and compression ratio) but differing in the number of channels would yield different output to a broadband stimulus in real life. This would be the case when one matches the output of both hearing aids to the same gain targets during verification with a tonal stimulus. The one with more channels would sound louder than the one with fewer channels to complex stimuli in real life. The output difference between the two hearing aids increases as the difference in the number of channels between them increases. The difference also increases as the compression ratio of the two hearing aids becomes higher. Linear hearing aids, on the other hand, do not show this output difference even if they differ in the number of channels.\(^2\)

Kuk and Ludvigsen’s (1999a) observations have several implications. First, it provides another piece of evidence to suggest that tonal stimulus is not a good stimulus to use to adjust gain on compression hearing aids, especially ones with multiple compression channels. Second, it suggests that current generic fitting targets, even if they truly predict real-life performance, must be updated to accommodate the difference in the number of channels seen in today’s hearing aids. Specifically, this means that one must modify the target values to include hearing aids with multiple channels. As of date, only the NAL-NL1 (National Acoustic Laboratories, 1999) prescriptive target has considered the effect for up to four channels. Such limitation in current prescriptive targets should caution one against strictly matching the output of a multichannel hearing aid to a prescriptive target.

### The effect of noise reduction has not been considered in the targets

Another area of concern in gain verification is the availability of noise reduction algorithms in many current hearing aids. Previously it was discussed how this mechanism could make the verification/measurement of output variable. This section questions the validity of the gain targets for these hearing aids, and examines how such hearing aids should be verified.

The noise reduction algorithm in most multichannel nonlinear hearing aids works by reducing or modifying the prescribed gain for a quiet listening situation. For such hearing aids to provide optimal listening in quiet and in noise, their manufacturers must have some knowledge of optimal gain requirement in these situations before they specify the amount of gain adjustment to be provided by the noise reduction algorithm. Indeed, several manufacturers of digital multichannel hearing aids have developed proprietary fitting formulae based on extensive in-house research and clinical data. Their proprietary gain targets have considered the dynamic changes in gain requirements beyond what would have been considered by generic prescriptive formulae.

A clinician who unknowingly adjusts the parametric settings on such hearing aids to match a generic target could risk compromising the hearing aid fit for the wearer—in quiet, in noise, or in both types of environments. Consequently, some manufacturers recommend against a strict target match. This may be the same reason why some researchers reported that the preferred gain of adult hearing aid wearers was closer to the manufacturer’s recommendation than to a generic target’s (Stelmachowicz et al, 1998). Because not all manufacturers of digital hearing aids have gone through this research effort, clinicians need to know the rationale behind the hearing aid gain target prior to deciding if a generic fitting target should be matched.

An equally complex issue is whether the noise reduction algorithm should be deactivated or should remain active during target matching (if one decides to match its output to a generic target). If one assumes that all prescriptive targets are generated for typical noise-free listening conditions, then

\(^2\)A “multichannel linear” hearing aid is essentially an equalizer and not a true multichannel hearing aid. A true multichannel hearing aid refers to one with at least two channels of independent compression.
the noise reduction algorithm should be bypassed during verification/target matching. The caveat is that the output may be “inflated” relative to a generic target if the manufacturer’s proprietary target has considered the effect of the noise reduction (assuming that the generic target has corrected for channel and release time effects also). In that case, one should guard against lowering the output/gain to meet the target. If one matches the target with the noise reduction algorithm activated, one may be examining the impact of noise on the hearing aid performance. One may receive less output than the prescriptive recommendation. If one proceeds to increase the gain setting to match a target, it could result in too much gain for everyday situations. These are complex issues that need further deliberation. Bearing in mind that a basic objective in any acceptable hearing aid fitting is to ensure audibility of the softest sound and comfort for the loudest sound, it would seem appropriate that the noise reduction algorithm be deactivated during verification to examine if the criterion of audibility is first and foremost achieved. Testing with noise reduction activated may be desirable to examine the comfort issue when loud sounds are present, but it should be a secondary objective at this point. Kuk and Ludvigsen (1999b) discuss these issues.

The Effect of Attack Time and Release Time Has Not Considered in the Targets

It was discussed earlier that the static characteristics on nonlinear hearing aids are selected based on either a prescriptive approach (e.g., FIG6, DSL[i/o], NAL-NL1) or an individual loudness rating approach using short, steady-state tone bursts or narrow bands of noise (e.g., LGOB, IHAFF). None of these approaches considered the effect of time constants (attack time and release time) on the subjective perception of the output of a nonlinear hearing aid.

It was discussed earlier that the effective compression ratio of a compression hearing aid is affected by the time constants of the hearing aid and the input acoustic signal. Real-life signals fluctuate greatly in their intensity levels. As one would predict, the shorter the time constants (or faster attack and release), the better is the nonlinear circuit in following the intensity variations of the input signal to ensure audibility of the weaker sounds. It also suggests an effective compression ratio that approximates more closely the nominal (or static) compression ratio. The intensity difference among speech components, however, will be decreased and ambient noise more noticeable during pauses in the speech input. In contrast, a longer release time (greater than 1 second or so) would preserve the intensity relationship of the speech components better. The effective compression ratio, on the other hand, would regress toward 1 (or linear) (Verschuure et al, 1996). Figure 5–15 shows the difference in waveform for the same speech input processed by two different release times. A linear representation of the same signal is shown for comparison. In the first case where a release time of 200 msec is used, clear reduction in the intensity difference is noted. In the second case where the release time is 10 seconds, the intensity difference is mostly retained. Subjectively, hearing aids with a longer release time sound clearer and more natural than hearing aids with a shorter release time (Neuman et al, 1995; Hansen, 2001). Spectrally, a longer release time results in an increase in the mid-frequency output over a shorter release time.

There are several implications from the study of the effect of release time on the effective compression ratio (and subjective preference). First, if two nonlinear hearing aids are identical in all but their time constants, the one with a longer attack and release time would sound more natural and pleasant than the one with a shorter attack/release. This suggests that matching the static I-O curve is no guarantee of subjective acceptance. Second, because the release time changes the overall loudness of the amplified sound, modification to the prescriptive target is necessary to account for the loudness effect resulting from a difference in release time. Such has not been considered in any generic prescriptive targets. Third,
if release time changes the dynamic I-O characteristics of the hearing aid, one may question the value of measuring loudness growth function for the fitting of compression parameters. After all, the effective compression ratio will not be the same as the static compression ratio that is determined through loudness growth procedures. And if the goal is to determine a general long-term compression ratio, several studies have confirmed that average loudness growth functions (and static compression ratios) can be predicted accurately and efficiently with threshold data (Hellman and Meiselman, 1993; Jenstad et al., 2000).

Reconsidering the Concept and the Value of the Aided Threshold

A frequently used verification tool, that of measuring the functional gain and/or the aided threshold, should be reconsidered. Functional gain is the gain provided by the hearing aid when it is worn over the unaided condition. It is obtained as a difference between the aided and the unaided sound field thresholds under the same test conditions. This index is not new. It has been used with linear hearing aids as a way to quantify the improvement in auditory sensitivity. However, many may have refrained from such claims because as one adjusts the volume control position on the linear hearing aid (as would be typical in real life), the magnitude of the aided threshold (and functional gain) changes. This significantly diminishes the meaningfulness of the aided threshold. When used with linear hearing aids, it was often cited that the magnitude of the functional gain was similar to the insertion gain that the wearer received for conversational input (Pascoe, 1975; Hawkins and Haskell, 1982).

In reality, one should recognize that the aided threshold represents the lowest input level that one can hear with a hearing aid at the specific gain setting. Because one would likely adjust the VC on a linear hearing aid in real life, the aided threshold could change from the value determined during sound field threshold measures. Thus, not many would claim that the aided threshold represents the softest sound that a wearer can hear.
hear with a linear hearing aid. If one can accept, however, that the design of a nonlinear hearing aid should free its wearer from level adjustment (via the VC), the aided threshold would represent the softest sound that the wearer hears with the nonlinear hearing aid. Such assumption is reasonable because the signal processing in many modern DSP nonlinear hearing aids has considered the changes in gain requirement across various input levels and SNR conditions (through noise reduction algorithm). Thus, the need for a VC is minimal (although some studies reported that a percentage of wearers still prefer to have a VC on their nonlinear hearing aids for safety and control reason). The aided sound-field threshold that is measured in the clinic should represent the softest sound that the wearer can hear with the nonlinear hearing aid. Knowing the softest audible sound can be extremely useful for documentation of hearing aid benefit (improvement in audibility/sensitivity or diminishment of impairment), or as a counseling tool to set proper expectations.

For example, one way to evaluate if a nonlinear fitting is optimal is to compare the value of the aided threshold with the average speech spectrum displayed on an audiogram. In a well-calibrated sound field, aided thresholds that are around 20 dB HL across frequencies would suggest that the softest part of the speech spectrum (which occurs around 20 dB HL across frequencies) is audible. A higher (or poorer) aided threshold would suggest that some of the speech spectrum may not be audible. In light of the average speech spectrum level, one may use the absolute value of the aided threshold as a target. For listening to the softest part of normal speech, an acceptable target would be an aided threshold of 20 dB HL across frequencies. An aided threshold that is lower should be achieved if better audibility is desired. Note, however, that this discussion is applicable only to nonlinear hearing aids. The same conclusion must not be applied to linear hearing aids, especially when fitting someone with a severe-to-profound hearing loss (Kuk, 2001b).

One can compare the results of the fitting of new nonlinear hearing aids with the wearer’s own linear hearing aids. If the output of the linear hearing aids and the new nonlinear hearing aids are matched to a conversational speech input, one would expect that the nonlinear hearing aids yield a lower (or better) aided threshold than the linear case. This has been demonstrated in Figure 5–2. If the aided thresholds for both hearing aids are the same, then either the linear hearing aids have given the wearer more gain than required, or the nonlinear hearing aid is not prescribing enough gain. In both cases, the wearer will complain of insufficient loudness from the nonlinear hearing aids.

The value of the aided threshold may serve as a clue for troubleshooting. It is not uncommon for experienced linear hearing aid wearers to complain that sounds are not loud enough with their new nonlinear hearing aids. Such complaints may indeed reflect insufficient gain at all input levels, or simply a reduction in gain at a high input level relative to the linear hearing aid experience. One approach to decide if further gain adjustment is needed is to examine the value of the aided thresholds. Wearers who have aided thresholds around 20 dB HL would have been ensured of adequate audibility and are probably reacting to the reduced loudness of nonlinear hearing aids to high input sounds. In such a case, counseling or adjustment of gain for high input sounds would be necessary. On the other hand, if the aided thresholds with the nonlinear hearing aid are similar to or poorer than the aided threshold of the linear hearing aid, adjustment of overall gain would be necessary. See Kuk (2001a) for a detailed description of the considerations in troubleshooting modern nonlinear hearing aids.

**Additional variables affecting the aided threshold**

One criticism of the concept of sound-field threshold measurement concerns the variability of the measured index (Humes and Kirn, 1990). Walker (1995) summarizes the environmental factors that may confound
the results of sound-field measures. This section concentrates on the additional considerations when one attempts to measure the aided threshold of a modern nonlinear hearing aid. The key factor affecting the accuracy and reliability of the measure is the time constants of the nonlinear hearing aids, and especially for devices with a low compression threshold (below 40 dB SPL).

First, because a compression hearing aid decreases its gain as input level increases, the typical bracketing approach in threshold estimation may yield more variable results when the real aided threshold is at or above the compression threshold of the nonlinear hearing aid. In this case, the input stimulus that is just above the CT receives lower gain than it would otherwise if it were below the CT. This may artificially raise the value of the aided threshold and/or add to the variability of the measurement. An alternative approach is to present the stimulus in an ascending manner once the aided threshold is approximated with a bracketing approach. This could minimize variability in the aided threshold measure.

Recall that the attack time is the time it takes a nonlinear hearing aid to settle into the reduced gain state when the input is above the CT (or compression active). Consequently, if the stimulus level is below the compression threshold, the duration of the stimulus would have no effect on the gain (and output) of the hearing aid. However, when the stimulus level is above the CT, the duration of the signal could interact with the attack time and affect the output. When the stimulus duration is significantly longer than the attack time, the initial part of the stimulus may receive the full gain of the hearing aid and the remaining part of the stimulus the reduced gain. If the initial part is significantly above the wearer's threshold, the wearer may indicate a threshold response. Otherwise, the stimulus level has to be raised until the reduced gain part is above the wearer's threshold. The drawing on the left of Figure 5–16 illustrates the situation.

In contrast, in hearing aids using a long attack time, the same stimulus would receive the same gain from the hearing aid. However, a higher output may be possible (resulting from a longer duration of linear gain), and a lower aided threshold results for the same audiometric threshold. Thus, two hearing aids with identical I-O characteristics could have different aided thresholds if they are significantly different in their attack times. Thus, knowledge of the attack time used by a compression system is useful. Given that most WDRC hearing aids use a relatively short attack time (less than 10 msec), it is foreseeable that the aided threshold is measured with the hearing aids in a reduced gain state unless the hearing aid has a high CT. Otherwise, the obtained threshold may be higher than the real threshold. For WDRC systems using a low CT, the use of a long attack time may slow the gain reduction and circumvent the problem of achieving a higher aided threshold. Considering the effect of temporal integration, CT, and attack time of most hearing aids, a stimulus duration of approximately 1 or 2 seconds should be used for reliable measure.

As discussed in the previous section on coupler and real-ear measures, the delay between stimulus presentation could also affect the reliability of the aided threshold. For most commercial WDRC hearing aids that
use short attack and short release times, no special considerations are necessary to pace the stimulus presentation accordingly because the typical delay between presentations is longer than the release time of the hearing aid (less than 200 msec). However, the delay between presentations can be more critical in a slow-acting WDRC system where the release time can be several seconds. If possible, one should wait for the period of the release time before the next presentation to minimize variability from presentation at various gain levels. But if one assumes that the difference between presentations is typically 5 dB, the minimum required delay between presentations can be as brief as 1 or 2 seconds, even for a slow-acting WDRC hearing aid.

In summary, the attack time and the release time on a nonlinear hearing aid may affect the reliability of the obtained aided threshold measure. Typically, an ascending approach with stimulus duration of about 1 or 2 seconds and delay between presentations of 1 or 2 seconds is appropriate to ensure a reliable outcome.

**Conclusion: What to Do with Current Nonlinear Hearing Aids**

Nonlinear hearing aids are different from linear hearing aids in that different gain values are assumed at different input levels. This provides significant advantages over linear hearing aids in meeting the wearer’s listening needs. At the same time, various aspects of its processing have not been considered in conventional verification approaches. Even for the simplest single channel WDRC hearing aids, their verification and validation would require special considerations beyond that of a linear hearing aid. The use of multichannel nonlinear hearing aids further complicates existing verification and validation efforts. The effects of the number of channels, time constants, compression threshold, and the presence of special features like noise reduction require a modification of our current verification practice to accommodate these advances. More advances are forthcoming to result in even more modifications to our current protocol.

These changes require clinicians to reconsider the appropriateness of their verification protocols for the chosen hearing aid. Knowledge of the static input-output curves of the selected hearing aid is needed, but is not enough. How the dynamic characteristics such as time constants and the noise reduction feature relate to the temporal characteristics of the hearing aid/real-ear test system should be thoroughly understood.

It is critical that hearing aid manufacturers provide explanations of the use of their hearing aids. How their hearing aids can be fitted accurately and evaluated fairly should be clearly articulated. Specific fitting and verification (and validation) protocols, if they deviate from the generally accepted practice, should be communicated to the clinicians who fit the hearing aids.

It is time that the profession reevaluates its protocol for fitting and verification of nonlinear hearing aids. The current approaches, although arguably appropriate for linear hearing aids, are not appropriate for use with modern multiple channel nonlinear hearing aids without significant revision and modification. Although clinicians must verify and validate the results of their clinical fittings, care must be applied in interpreting the results of their efforts. The profession must evaluate and agree on a common set of stimulus and test protocols so that some level of agreement in verification and validation is shared.

**References**


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