Fitting Strategies for Noise-Induced Hearing Loss

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Introduction

Of the approximately 28 million people in the United States with significant hearing loss, 10 million have hearing loss attributable to high-intensity noise (NIDCD, 1993). Noise-induced hearing loss (NIHL) is the most common occupational disease, with over 30 million workers exposed to hazardous noise levels that could result in hearing loss (NIOSH, 1996).

Potentially hazardous sources of excessive exposure to noise include occupational noise and sociocusis (i.e., hearing loss from loud noise at social activities, sports events, and fitness clubs, and from children’s toys and music generated from personal stereo headsets and concerts). Additionally, 50 million Americans use firearms, expanding the number of citizens exposed to high levels of noise (Kramer, 1990). The Occupational Safety and Health Administration (OSHA) mandates control of noise in the workplace to protect workers’ hearing. Unprotected exposure is limited by OSHA to 90 dBA for an 8-hour period (OSHA, 1983). For every 5-dB increase in intensity, the time is halved (i.e., 4-hour limit at 95 dBA, 2-hour limit at 100 dBA, etc.). Hull noted that 126 dBA, an intensity sometimes produced at rock concerts, can damage the ear in 1.85 minutes (Radcliffe, 2000). Clark et al (1998) evaluated children’s toys and found that of the 30 tested, seven produced average measures exceeding 120 dBA in the ear canal and several produced levels at 130 dBA or greater, clearly a health risk to children.

According to the Oticon Research Center (Oticon, 1989), a study performed by the Institute of Hearing Research in Nottingham, England, involving 25,000 randomly selected adults found that approximately 1% of the population experiences bilateral high-frequency hearing loss. If the statistics are recalculated to include unilateral loss, the prevalence increases to over 3%. Since NIHL is often insidious, it is impossible to determine how many citizens exposing themselves daily to harmful levels of noise may someday develop the symptoms that commonly accompany NIHL.

Despite these alarming prevalence rates, patients whose audiometric configurations suggest that the cause of their hearing loss is due to excessive noise exposure are currently woefully underserved by the use of amplification. This chapter explores several reasons that may help explain the lack of success in the past with conventional hearing aid fitting approaches for patients with NIHL, examines fitting alternatives that have recently been introduced for this population, and offers psychoacoustic considerations and practical fitting strategies for patients with NIHL.
Terminology and the Course of NIHL

A sudden loss of hearing from a single exposure of noise is called acoustic trauma. The gradual deterioration of hearing sensitivity occurring from several years of excessive noise exposure is referred to as noise-induced hearing loss. Understanding the nature of the hearing loss caused by exposure to intense noise can help formulate methods of providing amplification and aural rehabilitation. Mills (1988) offers this explanation of the nature of hearing loss following exposure to impulse versus continuous noise:

The nature of the hearing loss produced by acoustic impulses is probably different from the nature of the hearing loss caused by continuous exposures to steady state noise at moderate levels. That is, the excessive displacement of a basilar membrane by an intense acoustic impulse may produce mechanical ripping and tearing, whereas injuries produced by lower level acoustic signals may be caused by metabolic, biochemical or vascular effects, including the depletion of energy stores, mechanically induced changes in the shape of the tectorial membrane, and vasoconstriction within the cochlea. [p. 264]

The audiometric configuration resulting from noise exposure may be similar for either traumatic loss or NIHL (Ward and Frick, 1969). Similarly, the audiometric configuration may mimic that produced by presbycusis. In each case, the resultant hearing loss can present the listener with significant communication difficulties. A typical progression of hearing loss as a function of exposure to industrial noise is shown in Figure 6–1 (Newby, 1979). Maximum damage seems to occur on the portion of the basal turn of the cochlea that tonotopically corresponds to 3000 to 6000 Hz. Originally it was thought that the maximum hearing loss occurred in the 4000-Hz region because the hair cells in this region were more susceptible to damage. Lehnhardt (1965), however, reported that auditory fatigue and recovery at 4000 Hz were no different than at other frequencies. Therefore, it is more likely that initial damage is greatest in the 4000- to 6000-Hz range because of the combination of the normal ear-canal resonance, occurring at approximately 2700 Hz, and because the frequency region most affected is usually one-half to one octave above the frequency of the offending noise. As the exposure continues, the “notch” typically seen at 3000 to 6000 Hz tends to broaden to the frequency region below 3000 Hz, and the magnitude of hearing loss at 3000 to 6000 Hz increases.

The audiometric configurations of NIHL cover a wide range. For the purpose of clarity, however, only three representative audiometric configurations of NIHL are dis-

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**Figure 6-1.** Typical progression of noise-induced hearing loss (NIHL) as a function of years of exposure. (Adapted from Newby HA. *Audiology,* 4th ed. Englewood Cliffs, NJ: Prentice-Hall, 1979, pp. 322–325.)
cussed in this chapter. Type I, which may be considered representative of an audiometric configuration typical of only several years of exposure to noise, is characterized by near-normal hearing through 2000 Hz. Type II, which may be considered representative of many years of excessive exposure to noise, is typically illustrated by audiograms revealing hearing loss extending into the lower frequencies (i.e., below 2000 Hz). Type III, which is less common, represents the more extreme case in which hearing is near normal for the low frequencies only, and the audiometric configuration has a precipitous slope into the high frequencies. These three audiometric configurations are illustrated in Figure 6–2. For comparison purposes, note the expected age level hearing thresholds illustrated in Figure 6–3 (ANSI, 1996).

For further information regarding the evaluation and determination of occupational hearing loss, go to the Web site www.entusa.com/handicap.htm.

NIHL Candidacy for Amplification

If one were to apply the conventional guidelines concerning candidacy for amplification developed nearly 30 years ago, it is obvious that only type II and some type III patients with NIHL would be considered prospective hearing aid users (Hodgson, 1977). Griffing (1992) reported that when hearing handicap scales are provided by patients who might not have met candidacy criteria by conventional guidelines, a pattern emerges suggesting that a large number of these individuals experience significant difficulties in communicating. In addition, numerous studies have suggested the possibility of auditory deprivation, in which word recognition skills may deteriorate at a faster rate for unaided versus aided ears of listeners with hearing loss (Silman et al, 1984; Gelfand, 1995). Auditory deprivation also is demonstrated by neurophysiologic studies indicating that central auditory changes occur secondary to reduced neural input (Palmer et al, 1998; Emmer, 1999; Palmer, 1999). Additional reports by Sweatow et al (1991) reflect benefit for type I patients whose only complaint is the presence of tinnitus and who therefore might not be considered candidates for amplification strictly on the basis of degree of hearing loss. In reality, the only valid measure for determining candidacy for amplification is the extent of the communication problems experienced by the individual. Certain occupational demands (e.g., those experienced by judges or psychiatrists) further expand the need for amplification to an increasing number of type I patients with NIHL.
Why Amplification Has Been Unsuccessful for Patients with NIHL

Historically, there have been a number of factors contributing to the lack of confidence shared by the public and hearing health care professionals regarding the benefit derived from hearing aids for the NIHL population: (1) lack or misuse of handicap scales, thus rendering inappropriate assumptions regarding candidacy; (2) misinterpretation of medico-legal formulas that underestimate the effect of hearing loss on word recognition because they are based only on the average hearing loss from 500 to 2000 Hz; and (3) concentration of results based on mean group data rather than individual data, which can be misleading because success with high-frequency amplification may vary among individuals, even those having similar audiograms. Perhaps, however, the most compelling reason for this bleak attitude is the history of poor results obtained by patients with NIHL wearing conventional amplification.

Six additional factors that may have contributed to the failure of amplification for NIHL patients emerge and are explored in the following sections: (1) lack of motivation, (2) primary focus on word recognition measures, (3) inappropriate match between the needs of NIHL patients and current technology, (4) focus on transparency in the frequency domain rather than the amplitude domain, (5) excessive high-frequency emphasis, and (6) inadequate counseling.

Lack of Motivation

Hearing aids are often dispensed to patients who lack motivation for amplification. A poorly motivated patient is a poor candidate for amplification regardless of the degree of hearing loss (Alpiner, 1987). Vanity, along with the unfortunate, but undeniable, social stigma often associated with hearing aids, combined with normal word recognition scores in quiet conditions, produce listeners with hearing loss who are resistant to hearing aids. These potential candidates for amplification may have heard stories concerning hearing aids that make sounds uncomfortably loud, but not clear; consequently the aids are confined to dresser drawers. Furthermore, one must consider the cost-to-benefit ratio (i.e., that the expense associated with hearing aids may be unacceptable to a potential user who denies having anything more than a slight problem).

Primary Focus on Word Recognition Measures

There are four primary reasons listeners with hearing loss have decreased word recognition
skills: (1) reduced audibility of the speech cues that are important for correct recognition of words; (2) cochlear distortions that are manifested as reductions in frequency selectivity, temporal resolution, gap detection, and frequency and/or intensity discrimination; (3) central auditory nervous system deficiencies; and (4) deficits in cognitive processing. It is important to differentiate peripheral from central pathology as the cause for reduced word recognition ability. For central problems, or central problems coexisting with peripheral involvement, one must consider additional issues such as whether assistive listening devices are more appropriate than wearable amplification, whether hearing aids in addition to assistive listening devices are necessary, and whether the usual preference for binaural amplification is appropriate.

Regardless of the cause of diminished word recognition ability, word recognition measures, particularly in quiet, are poor prognostic indicators of success with amplification for patients with NIHL. For many of these patients, for example, word recognition scores are excellent when the words are presented in quiet. Thus, it is not possible to demonstrate improvement with amplification in these easy listening conditions. Difficulty in word recognition may arise, however, in noisy environments. Crandell (1991) reported data suggesting that pure-tone sensitivity and/or word recognition in quiet are poor predictors of ability to understand speech in noise. In his investigation, subjects with similar audiometric threshold configurations (mild to moderate sensorineural hearing loss) were tested with the revised speech perception in noise (SPIN) high-predictability sentences (Bilger et al., 1984) and multi-babble noise competition presented at levels of 75- and 85-dB sound pressure level (SPL). Nonsignificant relationships ($r = 0.34$ at 75-dB SPL noise level, and $0.09$ at 85-dB SPL noise level) were found between recognition scores in quiet and noise.

It is highly likely that comfortable listening in noise, as well as a release from the stress and fatigue produced by straining to hear, may be more important determinants of success with amplification than improved word recognition scores for patients with NIHL. For example, type I listeners may have excellent word recognition even without amplification. Type III listeners may not receive word-recognition score increases with amplification because of cochlear distortion and insufficient audibility even with high-frequency amplification. Even in the presence of better word recognition scores, unwanted effects such as excessive loudness, the occlusion effect, or internal noise levels generated by the hearing aids may lead to rejection of amplification.

Inappropriate Match between the Needs of NIHL Patients and Current Technology

Given our current understanding of outer hair cell (OHC) function and resultant loudness growth patterns, it is hard to believe that it has only been since the mid-1990s that the majority of hearing aid fittings in the United States have utilized compression amplification. Prior to that, most fittings incorporated linear signal processing.

If the normal cochlea was, as once thought, linear, passive, and broadly tuned, then linear amplification would be appropriate for most hearing loss. We now know, however, that this is certainly not an accurate description of the cochlea. The input-output function of the cochlea is nonlinear. At input sound levels of less than 60-dB SPL, a doubling of amplitude produces far less than a doubling of displacement of the basilar membrane. The active mechanical process of the OHCs “amplifies” these sounds while also sharpening frequency selectivity. Loudness recruitment results from damage to, or loss of, the active process in the cochlea that enhances sensitivity for low-input sound levels (below 60-dB SPL). Because this process is nonlinear, it results in an amplification of the basilar membrane response to lower intensity sounds (below 60-dB SPL) while leaving higher intensities (greater than 60-dB SPL) relatively unamplified. If this active process is compromised, the response to these lower input signals is unamplified and the absolute...
threshold is elevated. However, the response to the higher-level signals remains nearly the same as in the normal ear (Kates, 1991). An example of this concept is shown in Figure 6–4. In this example, the impaired listener would require 25 dB of gain for an input signal of 40-dB SPL, but only 1 dB of gain for an input signal of 80-dB SPL to achieve "normal" loudness. Results such as these can be established using loudness growth tests currently available in the software of several manufacturers' products. Examples include the ReSound LGOB, Phonak Loudness Perception Profile Test, Danavox ScalAdapt Loudness Scaling, and Sonic Innovations Dynamic Range Measurement. The Contour Test (Cox, 1995) is another loudness test that is available and used in certain prescriptive formula approaches such as the Independent Hearing Aid Fitting Forum (IHAFF) (Cox, 1995).

The use of nonlinear amplification attempts to solve two important fitting problems. One, compression minimizes the likelihood that amplified sound will reach the wearer's loudness discomfort level (LDL), thereby forcing the user to lower the volume control setting and resulting in inaudible components of the speech signal. Two, it prevents moderate input levels (e.g., 65- to 70-dB SPL) from driving the aid into saturation. It is believed that an important cause of decreased word recognition for listeners with sensorineural hearing loss in noisy situations is the presence of harmonic and intermodulation distortion (see Chapter 1) produced by hearing aids driven into saturation (Fabry, 1991). Thus, compression in itself likely does not enhance word recognition, but, rather decreases distortion (Kuk, 1996).

**Focus on Transparency in the Frequency Domain Rather Than the Amplitude Domain**

The so-called Harvard Report (Davis et al, 1947) suggested that most hearing aid users...
would benefit from hearing aids in which the frequency-gain response incorporated a 6-dB rise per octave. That philosophy has been replaced by the concept of selective amplification; that is, that optimal hearing can be achieved through the use of specified electroacoustic performance characteristics based on the magnitude and configuration of the individual hearing loss. The principle of transparency in the frequency domain implies that if a listener with hearing loss has frequency regions of normal hearing, attempts should be made to provide little or no amplification for that frequency region. This is, in fact, reflected in all of the prescriptive formulas currently in use. It is incorrect to assume, however, that there is a direct relationship between the degree of hearing loss at a specific frequency and the amount of gain required at that frequency for a given individual. Collection of loudness growth data on individuals will illustrate this disparity.

Excessive High-Frequency Emphasis

A goal of hearing aid fittings is to provide sufficient gain to render all speech sounds audible. In an attempt to ascertain the minimum high-frequency characteristics necessary to provide access to the important spectral cues for the sounds in English, Boothroyd and Medwetsky (1992) analyzed the /s/ phoneme because of its high-frequency content and importance to word recognition. They found that the lowest prominent spectral peak for the /s/ phoneme was 4300 Hz for males and 7200 Hz for females. Furthermore, they found that the frequency of the lowest prominent spectral peak varied by as much as 1000 Hz depending on coarticulation effects. For example, the /s/ phoneme in the /i/ context was 1000 Hz higher than it was in the /u/ context. Thus, if a hearing aid has an upper frequency limit that is lower than the lowest prominent spectral peak (i.e., 4300 Hz for male speakers and 7200 Hz for female speakers), there may be occasions when the listener will fail to hear the /s/ sound or may confuse it with the /f/ or /th/ phonemes. Some individuals could make these distinctions simply on the basis of the cues provided by formant transitions, but Zeng and Turner (1990) found that subjects with hearing loss did not use these formant transition cues as efficiently as normal listeners and needed to hear the specific fricative spectra to provide correct identification.

Research findings are contradictory on whether it is necessary, or even desirable, however, to extend the high-frequency response of wearable amplification for listeners. Pascoe (1975) demonstrated, under controlled lab conditions, the advantage of extending bandwidth beyond 3000 Hz for improved word reception in subjects with hearing loss. Murray and Byrne (1986) varied the upper cut-off frequency among 1500, 2500, 3500, and 4500 Hz and had normal-hearing subjects as well as subjects with hearing loss judge intelligibility and “pleasantness.” Normal listeners judged the wider bandwidths (e.g., upper cut-off frequencies at 4500 Hz) as better on both intelligibility and pleasantness. The subjects with hearing loss, however, seldom reported additional benefit when the bandwidth was extended beyond 2500 Hz. Unfortunately, it was difficult to ascertain from this experiment whether the high-frequency gain was sufficient above 2500 Hz to allow the speech signals to be audible for the listeners with hearing loss. Sullivan et al (1991) indicated, on the other hand, that the subjects with hearing loss experienced improved word recognition with the addition of spectral information above 2000 Hz. They did not divide the spec-
tral information above 2000 Hz into smaller bands, however, as did Murray and Byrne (1986). Rankovic (1989) stated that, for the majority of her subjects, word recognition increased as the high-frequency region was amplified; however, for subjects with sloping high-frequency loss, the conditions that produced the greatest high-frequency gain resulted in decreased aided performance. Skinner (1988) stated that increasing high-frequency gain by more than 35 dB above low-frequency gain caused a decrement in performance for some, though not all, listeners. She speculated that spectral balance is needed to maintain optimal performance.

In another study, Sullivan et al (1992) reported at least three factors affecting performance of listeners with steeply sloping losses: (1) the overall gain of the aids, (2) the presence of background noise, and (3) the type of performance measure. In quiet, amplification with the broadest response was reported to provide the best performance. This finding is similar to that reported by Punch and Beck (1986), who demonstrated that both subjects with normal hearing and subjects with hearing loss associated better sound quality with speech containing low-frequency energy.

Sullivan et al (1992) found that the lack of high-frequency spectral information can be compensated for by the addition of low- and midfrequency spectral information. They also reported that high-frequency energy was more critical for syllable recognition in noise than in quiet. They concluded that, as audible bandwidth increases above 2000 Hz, word recognition performance improves, but subjective judgments of word intelligibility do not. This is in contrast with normal listeners, who report improvement in sound quality with increased bandwidth.

Lest one believe that studies finding decrements in performance with increased bandwidth may be related to earlier hearing aid technology that contained restricted bandwidths, recent investigations also indicate that high-frequency amplification for individuals having thresholds poorer than 55- to 60-dB hearing level (HL) at 3000 Hz and above may perform better without amplification in those high-frequency regions (Ching et al, 1998; Turner and Cummings, 1999).

It may be helpful to consider more than threshold and loudness growth patterns in determining the need for gain at a particular frequency. This is particularly applicable for type III patients with NIHL. Masking experiments can be used to determine if a patient has a complete loss of functioning neurons over a certain range. Moore (2001) provides a detailed review of these various masking procedures. Moore et al (1985) documented an experimental procedure in which the threshold of a tone located between two bands of noise with a notch between them may increase as the notch increases, instead of decreasing in the usual way. The authors theorized that this would occur if the signal frequency falls in a range where there are no functioning neurons. This experimental procedure can be used to reveal a reduction or complete absence of high-frequency neurons. Recently, Moore et al (2000) and Moore (2001) described threshold equalizing noise (TEN), an acoustic stimulus that can be used to identify “dead regions.” When high-frequency neurons are not functional, providing high-frequency amplification to restore audibility is questionable. Furthermore, patients without functioning high-frequency hair cells report that high-frequency sounds such as pure tones are heard as “noise” (Moore, 1991; Sweetow, 1994). In such cases, it may be better either to filter out the high frequencies or to use a transposing filter (Velmans and Marcuson, 1983). Recently AVR Communications Ltd. developed earlevel instruments (ImpaCt and Logicom-20) with “proportional frequency compression.” This technology moves high-frequency speech information to a lower frequency region in which hearing is better (Davis, 2001). Specifically, the “dynamic speech recoding” process for these instruments identifies speech characteristics to determine if the particular incoming speech sounds need to be frequency compressed, and then the selective proportional frequency compression process ensures that only desired high-
frequency phonemes like /s/, /f/, /t/, /k/, /sh/, and voiceless /th/ are moved into the lower frequency regions of residual hearing (Davis, 2001).

Adding to the uncertainty regarding the advisability of providing high-frequency amplification for patients having severe hearing loss is the fact that producing significant amounts of high-frequency gain has traditionally been problematic because of feedback and distortion issues. Lowering the volume control to reduce feedback will decrease the intensity for audible speech in the frequency region in which the patient has audible hearing. As discussed later in this chapter, these limitations have been minimized by digital feedback reduction and advanced compression, and, it is hoped, will be removed in the near future as more hearing aids incorporate digital technology.

Given these factors, it may be prudent to concentrate efforts on obtaining a better midfrequency response, or a response in frequency regions in which the patient has hearing loss less than 60-dB HL. This is particularly applicable to type III patients with NIHL who perceive high-frequency sounds as "noise" or as annoying. Staab (1988) stressed the importance of midfrequency amplification (500 to 2000 Hz). He reviewed several studies that showed that lower frequencies contribute more to perception of continuous discourse than to the identification of nonsense syllables. The articulation index (French and Steinberg, 1947) suggests that 70% of the cues for word recognition are located between 500 and 2000 Hz. In fact, the frequency band of 500 to 1000 Hz contributes 35% to the power and 35% to the recognition of speech. Staab also emphasized the importance of formant transition (i.e., the shifts in spectral energy composition when moving from one phoneme to another) perception in making consonant distinctions. This information is centered in the midfrequency regions. Staab concluded that too much emphasis is placed on attempting to establish electronic high-frequency emphasis and more should be placed on establishing a smooth frequency-gain response in the midfrequency region. Of course, this argument might not be applicable to all type II and many type I patients for whom efforts at rendering high-frequency consonants audible should be successful.

Inadequate Counseling

Because many patients with NIHL deny the presence of a hearing loss or lack sufficient motivation to treat a hearing loss, they often demand to be convinced of the improvement hearing aids can provide. Because the major goal of amplification is to facilitate the ease of communication, some type I patients may be disappointed when they initially experience only minimal benefit from amplification. Proper counseling to establish realistic expectations and educate patients regarding potential benefits and limitations can minimize this frustration. Patients must be educated that prediction of long-term benefit from amplification is tenuous at best because of the initial adjustment and learning process that takes place. According to Berger and Hagberg (1982), most hearing aid wearers require several weeks before they adjust to hearing aids. Barford (1979) hypothesized that a preliminary auditory analysis consisting of converting incoming acoustic signals into neural impulses is followed by a "recognition device," which matches these neural impulses to previously learned information to properly recognize phonemes. Frequency-dependent hearing aids (i.e., hearing aids that provide different gain for different frequencies) modify the speech input and thus change the output for the preliminary auditory analysis by producing new patterns. Therefore, it takes considerable time for the user to adapt to the new pattern and to learn new "recognition" cues.

Patients also should be informed that there might be no measurable improvement in word recognition performance in quiet, even if the hearing aid seems beneficial to the listener. Hafter and Schlauch (1992) provided an explanation for this apparent discrepancy. They reasoned that this occurs because the hearing aid evaluation process
itself may be reflected, rather than the benefit from the hearing aids. Recognition scores improve for laboratory tasks that are void of extraneous demands on attention. Word recognition testing in quiet environments is an example of such a low-demand task. Subjects in this situation might prefer amplified speech because it is easier to process; however, performance may not be improved because the hearing aids are not providing enough new information to the patient that could not be accessed simply by applying full attention. In other words, the hearing aids help in quiet by allowing the subject less effort, but recognition scores, which were near the maximum when tested unaided (i.e., ceiling effect), are not significantly improved. The true test occurs in noise or in high-demand environments where attention is resource limited. In these conditions, the information provided by the hearing aids may no longer be redundant with processes the listener could have accessed on his own by paying full attention, so now the devices may be shown to be of value. Thus, testing must be done in high-demand conditions, such as those created by adverse (i.e., negative) signal-to-noise ratios (SNRs).

**Fitting Strategies for Patients with NIHL**

Knowledge correlating clinical data to the establishment of electroacoustic parameters for a given individual remains inadequate. Given the complexity and variety of needs exhibited by patients with NIHL, however, the flexibility provided by digitally programmable and digital hearing aids can be particularly advantageous to the fitter. Prior to the advent of this technology, a number of trade-offs had to be considered. These included such issues as noise reduction versus speech intelligibility, intelligibility versus tinny speech quality, gain versus loudness comfort, and feedback versus high-frequency gain. Some of the primary advantages of this newer technology include flexibility in frequency shaping, microphone noise reduction, digital noise reduction, multichannel adaptive compression, user controlled multiple memories, digital feedback reduction, and directional/multiple microphones. With these features, the necessity for selecting one component of the aforementioned trade-offs at the expense of the other is no longer always necessary.

**Frequency Shaping**

Traditionally, in fitting conventional hearing aids, the use of potentiometers in combination with a variety of earmold configurations allowed the fitter to achieve a fairly wide range of frequency-gain responses. These were appropriate for the easier to fit type II NIHL patient, but were not always appropriate for the type I patient, for whom any degree of low-frequency gain could be offensive, or for the type III patient, for whom sufficient high-frequency gain might not be achievable without producing distortion or feedback. Recognizing this fact, in 1990, one manufacturer (Oticon) introduced a postauricular hearing aid (E43) with a frequency-gain response that rose 24 dB/octave above 2000 Hz and provided essentially no gain below 2000 Hz. This hearing aid was coupled to a specially designed earmold that produced little, if any, alteration of the real-ear unaided gain (REUG). It is interesting to note that during field testing of this hearing aid, Oticon found that 37% of the subjects with hearing loss confined to 3000 to 8000 Hz reduced the volume control to avoid listening to the internal noise generated by the hearing aid. The remaining 63% of the subjects adjusted the volume control to obtain satisfactory high-frequency amplification. The manufacturer also reported that most of its test subjects preferred real-ear gain equivalent to 40% of their hearing loss at 4000 and 6000 Hz. This finding underscored the need for a different prescriptive formula for the type I population.

Some conventional analog hearing aids also provided high-frequency reduction tone controls. Earlier, it was suggested that excessive high-frequency amplification might be
unnecessary and even detrimental to word recognition ability for certain listeners with type III precipitously sloping NIHL. If this high-frequency gain can be filtered out without adversely affecting audibility to the extent that speech becomes unclear, a more satisfactory fitting might result.

The frequency-gain response selected by the fitter was often based on prescriptive formulas. Unfortunately, the use of these formulas was often inadequate for predicting the "desired" fitting parameters. A variety of reasons accounted for this shortcoming. First, subject preference and performance with hearing aids change as intensity levels are varied. Tecca and Goldstein (1984) reported that subjects prefer more low-frequency gain at a comfortable listening level, but less low-frequency gain with higher input levels. Near threshold a patient might prefer a "four-fifths gain rule," but as the LDL is approached, the preferred gain might be one-fifth of threshold, or even no gain at all.

Second, until recently, prescriptive formulas were based on linear amplification. For hearing aids using compression, increasing the input level results in decreasing gain within the compression range. Thus, the target is a function of input intensity level, and the input levels that are most appropriate for defining the valid target have not yet been firmly established.

Third, the use-gain preferred by some patients with NIHL may not be well predicted by any current prescriptive formula. Mueller et al (1991) suggested that the National Acoustics Laboratories-Revised, Profound (NAL-RP) formula (Byrne and Dillon, 1986) and most other linear prescriptive formulas prescribe real-ear gain that is considerably greater than the use-gain preferred by listeners whose hearing loss exhibited normal hearing through 2000 Hz. In fact, they found that the preferred use-gain was only 20% of the hearing loss at 3000 Hz, far less than the use-gain prescribed by NAL-RP or the other formulas. In their experiment, however, Mueller et al reported that most of their subjects' hearing aids produced excessive gain in the 2000-Hz region and that this may have accounted for the patients' preference for a lower overall volume. One must recognize that even with current technology, there are limitations in providing adequate gain for the higher frequencies without providing any gain for the lower frequencies (e.g., below 2000 Hz). It is interesting to note that although some listeners in the Mueller et al study preferred as little as 5 to 8 dB of real-ear insertion gain (REIG), 10 of the 12 inexperienced hearing aid users preferred amplification and indicated improvement in communication. Thus, when fitting a patient with NIHL, it is important to determine whether exclusive amplification of the high frequencies is most useful for the listener, or whether amplification should also be provided in the midfrequency region.

Fourth, some conventional hearing aid users complained of hearing circuit noise. If these noise levels are audible to the type I NIHL listener, the hearing aid may be rejected (Chapter 1 of Hearing Aids: Standards, Options, and Limitations, 2nd edition, Thieme Medical Publishers, contains an expanded discussion on equivalent input noise level).

Given the above findings and the wide range of subjective preferences that cannot be predicted by prescriptive formulae, the use of programmable and digital instruments can minimize many problems. For example, multiband programmable hearing instruments, particularly those with three bands or more, can be especially useful for type III patients because gain can be minimized for the low frequencies, maximized for the midfrequencies, and then minimized for the high frequencies (if so desired). If, on the other hand, patients can capitalize on high-frequency amplification, then the gain in that frequency band apparently can be programmed. For patients having unusual REUGs, atypical frequency shaping can be fashioned to the individual's needs. Multiple frequency bands also can be utilized to shape the frequency-gain response by altering the crossover frequencies to overcome interband restrictions (i.e., 30-dB maximum difference between bands). In addition, by using a combination of expansion and com-

**Adaptive Amplification**

Adaptive amplification means that gain is automatically adjusted based on the characteristics of the input signal. The objectives can be to enhance intelligibility, to improve listening comfort, and/or to create noise reduction. Crandell (1991) indicated that for listeners with normal hearing, even a 1-dB enhancement in SNR can result in a 6 to 8% improvement in word recognition when evaluated with sentences from the SPIN test (Bilger et al, 1984). Hawkins (1985) reported an articulation function that rises at a rate of 7%/dB. In other words, a 5-dB improvement of SNR theoretically could provide as much as a 35% increase in word recognition for listeners with sensorineural hearing loss who have good word recognition abilities. One means of providing a perceptual, albeit not a physical, SNR improvement is to use hearing aids that automatically reduce low-frequency gain in response to the level of the input signal. That is, at input levels below a defined kneepoint (usually less than 65-dB SPL) the hearing aids provide a broad, flatter frequency response. As the input level increases, however, the signal processing provides progressively less low-frequency gain without altering the high-frequency gain. There are two major limitations to this approach. First, not all background noise has spectra limited to the low frequencies, and, second, these hearing aids would not be appropriate for type I NIHL patients for whom virtually no low-frequency gain would be desired, even when listening in quiet.

Research findings on hearing aids with automatic low-frequency reduction have been mixed. Most studies (e.g., Stein and Dempsey-Hart, 1984; Van Tasell et al, 1988) suggest that they do not usually result in significant improvement in word recognition in comparison to conventional linear hearing aids unless (1) the conventional aid provides an inappropriate frequency-gain response or excessive maximum output for the subject’s needs, or (2) the background noise is limited to the low frequencies. It is likely, therefore, that patients with type II NIHL who indicate a preference for aids with automatic noise-reduction circuitry are reacting in a positive manner to reduced listening efforts in noise and/or improved sound quality in comparison to listening through linear hearing aids.

Adaptive amplification also can be obtained using hearing aids containing K-Amp circuitry. For these aids, the high frequencies are emphasized for low-input intensities, but, as the input level increases, the frequency-gain response flattens, approximating the subject’s REUG (Killion, 1990). This is another example of achieving transparency in the amplitude domain for signals that do not require any signal processing to make them audible to the listener with hearing loss.

Programmable hearing aids have the capacity to incorporate adaptive low-frequency reduction [also termed *bass increase at low levels* (BILL)] and/or adaptive high-frequency emphasis [also termed *treble increase at low levels* (TILL)] (Preves, 1999). Programmable aids can be adjusted to provide BILL merely by setting a low compression kneepoint (e.g., 45-dB SPL) in the low frequencies and a high (e.g., above 70-dB SPL) compression threshold (or even a linear response) in the high frequencies. The opposite effect (TILL) can be achieved by setting a similarly low compression threshold for the high frequencies and a linear or high compression threshold for the low frequencies. Both BILL and TILL programming (analogous to the K-Amp circuitry) would be more effective for the listener with type II than type I NIHL because of the regions of normal hearing present for type I listeners. That is, both could generate too much audible low-frequency noise in quiet environments. Alternatively, both strategies can be pro-
grammed into a single hearing aid using multiple programs.

Digital hearing aids now have the capacity to utilize adaptive amplification throughout the frequency spectrum. Digital signal processing (DSP) allows instruments to attempt a differentiation of noise from speech, not only on the basis of spectral composition, but also on the basis of temporal characteristics. Noise and speech have quite distinct temporal patterns. For example, speech modulates at a much slower rate than does noise. Thus, DSP hearing instruments assess the modulation pattern (rate and depth) of the input signal to predict whether or not that signal is primarily speech. If it is, full amplification will be provided. If not, gain will be attenuated within that frequency band. The benefits of DSP for noise reduction are not yet certain. Although some studies did not demonstrate improved word recognition in noise, Bray and Nilsson (2000) reported significant improvements in word recognition for this type of noise-reduction strategy when compared to conventional amplification. Readers are cautioned to recognize that some studies showing enhanced word recognition in noise used DSP devices containing directional microphones, and thus may have achieved these effects due more to directional/multiple microphones than due to DSP. However, Valente et al (1999b) found it difficult to conclude that the improvement found in their study could be totally explained by the front-to-back ratio provided by the directional microphone. They suggested that the speech enhancement algorithm used in the hearing aid might have worked in synergy with the directional microphone, thereby enhancing word recognition in noise to a greater extent than that reported in the past for analog instruments with single or dual directional microphones.

Also, it should be noted that different DSPs incorporate various time constants. Certain manufacturers adhere to the philosophy that to enhance comfort in noisy environments, time constants should be slow (e.g., longer than 150 msec). Others believe that to preserve spectral differences, time constants should be fast (e.g., 100 msec or less). This will be discussed in greater detail in the sections that follow. (Chapter 3 in Hearing Aids: Standards, Options, and Limitations, 2nd edition, contains a detailed discussion of DSP.)

**Compression**

It has been proposed in this chapter that the majority of patients with NIHL should be fit with hearing aids providing nonlinear amplification. The optimal parameters (i.e., number of compression bands, compression kneepoint, compression ratio, and time constants) remain to be resolved. Because aspects of compression are discussed elsewhere in this text, only a brief discussion of the relevance to fitting patients with NIHL will be offered here. (Chapter 5 in Hearing Aids: Standards, Options, and Limitations, 2nd edition, contains an expanded discussion on compression and expansion.)

**NUMBER OF COMPRESSION CHANNELS**

In this chapter, **bands** refer to frequencies that are grouped together for the purpose of frequency shaping. **Channels** refer to frequencies that are grouped together for the purpose of compression. For example, the Oticon DigiFocus contains seven frequency bands for flexible frequency shaping but only two channels of compression, whereas the Widex Senso C series provides three bands and three channels. In addition to accounting for different patterns of loudness growth, the advantage of multichannel compression is that invasive sounds restricted to limited frequencies will not activate compression across the entire frequency range. Therefore, even type I patients who receive minimal or no low-frequency gain can benefit from multichannel compression in that low-frequency input signals will not lower high-frequency gain and resulting consonant audibility.

A number of the digital hearing aids currently available have numerous channels (as
many as 14). Having a high number of channels makes theoretical sense in that the normal human cochlea is characterized by as many as 23 critical bands. Multiple bands, as stated earlier, allow the fitter great flexibility in frequency shaping. There is no empirical evidence, however, demonstrating that a high number of channels benefit the listener with hearing loss in terms of word recognition. Leek et al (1987) indicated that reductions in spectral contrasts might add to the difficulty in detecting the spectral features in complex sounds. To avoid this problem, bandwidths of compression should be substantially greater than those of the auditory filters of the listener with hearing loss. Considering that listeners with sensorineural hearing loss already have widened auditory bandwidths, Kates (1991) concluded that no more than three channels of compression are useful. Moore (1991) also confirmed that too many channels of compression can reduce spectral contrasts (between vowels and consonants, and even with these two groups). He reported that an ear with sensorineural hearing loss has about 10 critical bands to process information. He theorized that unless the hearing aid has more than 10 channels, it will do little to reduce the masking effects of background noise. It is important to recognize that the statements by Kates and Moore were made prior to technologic changes that now allow for the production of hearing aids with more than 10 channels.

**Compression Ratio**

Compression ratio (CR) refers to the ratio of change in input level to the corresponding change in output level. Normally functioning OHCs act as a nonlinear “cochlear amplifier,” providing up to 60 dB of gain for low-input sounds (e.g., 0 dB) and no gain for high-input sounds (e.g., 100 dB). The listener with impaired OHC function has no “cochlear amplifier” for high- or low-input sounds. Hypothetically, as a result, the listener with normal hearing might have a dynamic range on the order of 100 dB, whereas the listener with sensorineural hearing loss might have a dynamic range that is narrowed to 40 dB. Thus, it can be argued that to restore “normal” nonlinearity to the ear with sensorineural hearing loss, a CR of no greater than 2.5:1 should suffice. High CRs (e.g., 8:1) with multichannel automatic gain control (AGC) having relatively low kneepoints may degrade the relative intensity cues required to identify certain speech sounds (Plomp, 1988). Licklider and Pollack (1948), however, showed that with amplitude clipping so severe that no intensity contrasts existed among the speech sounds, speech was still intelligible to listeners with
normal hearing. It should also be considered that even if spectral contrasts are reduced because of high-frequency band compression (with linear low-frequency processing), high-frequency recruitment might compensate and reinstate the loudness differences. The audiologist should be cautioned, however, that experienced users of hearing aids may report that too much compression (be it low knee-points or high CRs) causes muffling and insufficient perceptual loudness.

**COMPRESSION TIME CONSTANTS**

Early compression hearing aids were characterized by a fixed attack and release time (i.e., the time it takes for the gain to decrease from linear to the fully compressed state, or conversely, to increase from the compressed state to the linear state). In the 1980s the concept of adaptive compression was introduced to minimize periods of inaudibility that occur as a result of listening in an acoustic environment characterized by loud transient sounds (Smriga, 1986). In this design, release time varies as a function of the duration of incoming noise. For example, a short burst of noise (such as a spoon dropping on a hard surface) produces a shorter release time than an ongoing noise (such as constant background chatter or machine noise). If the release time is too short, the SNR ratio is decreased. If the release time is too long following transient noises, the hearing aid remains in compression during a period of time in which the listener wants to listen to speech. Such compression activity may render speech components inaudible while the gain recovers to its desired level. It is interesting to note that there is not universal agreement on the optimal time constants. For example, two of the more popular devices incorporate diametrically opposed philosophies. In one, slow-acting compression is utilized to maintain comfort for the listener in a variety of environments (Kuk, 1998), whereas in the other, very fast compression is used to maintain spectral contrasts and simulate cochlear function (Bray and Nilsson, 2000).

For hearing aids that do not have adaptive time constants, it is best to have attack (and release) times that are longer for low-frequency inputs than for high-frequency inputs. This is because vowels have a longer duration than consonants, and most, though not all, noise is composed primarily of low-frequency energy. High-frequency sounds fluctuate within a time frame of 40 msec, and vowels may last for 500 msec. It is best to activate low-frequency filtering for long-duration sounds and not for fluctuating sounds of less than 40 or 50 msec. If different time constants are not used for low- versus high-frequency inputs, normal conversational speech could activate an unnecessary adaptive filtered response. These considerations are incorporated into many of the programmable analog and, to an even greater extent, in fully digital hearing instruments. (For an expanded discussion on compression time constants, see Chapter 5 in Hearing Aids: Standards, Options and Limitations, 2nd edition.)

**User-Controlled Multiple Programs**

Another useful feature available in some programmable and digital hearing aids is multiple programs. No single hearing aid is optimal for all listening environments (Libby and Sweetow, 1987). Therefore, it seems logical to offer the patient choices of acoustic responses to best interface with the environment in which listening is taking place. This concept is not new. For years, some conventional analog hearing aids have offered user-manipulated tone control switches. Some of the limitations inherent in automatically adaptable responses can be avoided by allowing users to select among programs according to their own preferences. Users often give superior ratings to hearing aids with multiple programs (Kochkin, 1996). Keidser et al (1995) showed that multiple programs that varied in frequency-response characteristics might be more useful for users with flatter audiometric configurations, as opposed to those with normal hearing in the low frequencies. Thus, type II and III NIHL
patients might be most appropriate for this feature. In another paper, Keidser (1995) demonstrated that certain combinations of frequency shaping and compression schemes enhance listening in different environments. In addition to frequency shaping, current multiple-program digital instruments also offer flexibility in numerous compression parameters, directional characteristics, feedback management, and noise reduction.

The optimal number of multiple programs to meet listening needs is unknown. There are no research data to support the concept that more program choices are better. It is possible that too many choices might confuse the user and create difficulty in rapidly selecting the “right” program. Even so, the psychological advantage of offering the patient some “control” over the listening environment is significant. Current multiple-program devices contain from two to four choices. If the device has no volume control, yet the patient desires controllable changes in volume, devices having more programs might be beneficial so that program selection acts as a pseudo-volume control.

The usefulness of multiple programs is not limited to providing choices to the user in a variety of acoustic environments. Some new users, particularly those with type I NIHL, may find high-frequency emphasis hearing aids objectionable and “tinny.” Sweetow and Mueller (1991) reported on a patient with normal hearing through 1500 Hz who effectively adjusted to a high-frequency emphasis hearing aid by initially selecting a program containing a broader response, and then progressively moving through to a higher emphasis response. The patient’s primary hearing difficulty occurred during group meetings. At the time of the fitting, target gain was calculated and the instrument was programmed. The coupler response is shown as response A of Figure 6–5. The patient objected to this response saying that it sounded “tinny” and did not think the use of hearing aids made a significant difference in his hearing ability. By manipulating the low cut and slope adjustment features, responses B, C, and D were set in the remaining three memory locations of the instrument. The patient now had the option of adding 15 dB of gain at 1000 Hz (response D), even though most prescriptive formulas would suggest that this response would overamplify the low frequencies. As the patient became accustomed to programmable hearing aid use, he gradually increased his

![Figure 6-5](image-url)
use of response A by progressing through responses C and B. The flexibility of providing four different settings led to amplification acceptance that may not have occurred had he been offered only the amplification response shown in response A.

Some type II and type III patients express difficulty hearing on the telephone. Some, but not all, instruments will contain adequate telecoil gain. Some manufacturers provide a programmable telecoil (e.g., Phonak, ReSound, Sonic Innovations) that can be beneficial in meeting the patient's needs. Widex reports a telecoil frequency-gain response that matches the frequency-gain response of the microphone. For those devices that either do not have sufficient telecoil sensitivity or do not contain a telecoil, it might be practical to reserve one program for telephone use. It might also be beneficial to frequency shape so as to limit amplification to frequencies below 3000 Hz, because most telephones do not transduce above this frequency, and so that feedback is limited. Or, digital hearing aids having feedback reduction might be employed so that high-frequency gain does not have to be compromised.

Digital Feedback Reduction

Although some of the advantages of digital hearing aids are difficult to demonstrate, the advantages of digital feedback reduction are apparent and, indeed, objectively assessed. This active approach is very different from traditional feedback management approaches in that, rather than simply reducing gain in certain frequency regions (generally high frequencies), digital feedback control seeks out and minimizes feedback by means of phase shifting technology (Edwards et al., 1998). Clinical measurements have shown that these systems provide feedback margins of at least 10 dB (Groth, 1999). This can be extremely important for NIHL patients because many users will require significant high-frequency gain but will prefer earmolds that do not occlude their canals. Once digital feedback control is introduced, however, it is important to note that some users experience distortion from certain environmental sounds. (Chapter 3 in *Hearing Aids: Standards, Options and Limitations*, 2nd edition, contains an expanded discussion on feedback reduction in DSP hearing aids.)

### Directional and Dual/Multiple Microphones

NIHL patients frequently report that their primary communication difficulty is hearing in noisy environments. Despite advances in hearing aid technology, the only true method of improving SNR is with remote microphone location, such as with assistive listening devices. Unfortunately, many hearing aid users are loath to call attention to themselves by using these instruments. An additional strategy is the use of directional or multiple microphones.

Improvement in the SNR ratio can reduce noise and improve comfort in noise. There are excellent reviews of currently available microphone arrays including directional, multiple, and beam-forming technologies (Ricketts and Mueller, 1999; Schuchman et al., 1999; Valente et al., 1999a; Wolf et al., 1999). These approaches allow for communication between microphones that can effectively minimize the gain based on the directional origin of the incoming signal. Thus, signals originating from behind the listener (i.e., azimuths of 135 to 225 degrees) can be significantly suppressed relative to sounds occurring from in front of the listener. Among the improvements in microphone technology in some of the new digital instruments is a variety of polar patterns (the directional pattern of suppression), including automatically adapting rotating polar patterns (Strom, 2000) and user-switchable omnidirectional/directional operating modes (Prewes et al., 1999). Although omnidirectional microphones are often preferred for quiet listening, significant improvement in noise is consistently shown in multiple microphone modes [i.e., objective improvement using the Hearing in Noise Test (HINT) (Nilsson et al., 1994) and subjective improvement...
using the Abbreviated Profile of Hearing Aid Benefit (APHAB) (Cox and Alexander, 1995]).

Directional microphones are not new; they were available over 20 years ago in conventional hearing aids. The use of programmable, and particularly digital, technology, however, allows for potentially enhanced performance due to greater control and flexibility in the polar pattern. In addition, multiple microphone digital and programmable hearing instruments showed superior customer satisfaction on a significantly greater number of variables compared to single microphone programmable and digital instruments (Kochkin, 2000).

Despite the advantages, it should be noted that benefits from multiple and directional microphones might be minimized by highly reverberant environments and by longer release times when maintaining sounds at a comfortable level. (Chapters 2 and 7 in Hearing Aids: Standards, Options and Limitations, 2nd edition, contain an expanded discussion on directional microphones.)

**Other Considerations**

**Earmold Acoustics**

If a listener has normal or near-normal low frequency hearing, as do most patients with type I and type II NIHL, it is usually desirable to minimize occlusion of the ear canal, within limits imposed by feedback. The unwanted enhancement of the low frequencies of the user’s voice is called the *occlusion effect*. It might be useful to have the patient vocalize words with both closed and open vowels (such as /i/ and /u/) to ascertain whether the occlusion effect is likely to be unacceptable to the user. Even some type III patients can benefit from more open coupling if the desired frequency response does not require substantial high-frequency gain. If significant amounts of high-frequency real-ear gain are required (i.e., 30 dB or more), a more closed coupling system will be necessary. Special earmolds, such as the continuous flow adaptor (CFA), Libby horns, and deeply seated earmolds may be of assistance in enhancing high-frequency gain and minimizing occlusion. Also, less occlusive earmolds can be utilized with digital and programmable hearing aids with features such as digital feedback control and multiband frequency shaping. It should be noted that there are certain trade-offs that must be considered. Specifically, directional and multiple microphones are not as effective when the ear is not occluded because of venting and low-frequency (500 to 1000 Hz) attenuation (Ricketts, 2000). (Chapter 6 in Hearing Aids: Standards, Options and Limitations, 2nd edition, contains an expanded discussion on earmolds.)

**Hearing Aid Style**

With conventional instruments it might be difficult to achieve the type of frequency response (minimal low-frequency with adequate high-frequency gain) for NIHL patients using in-the-canal (ITC), or even certain IROS in-the-ear (ITE) hearing aids. If the external auditory meatus is sufficiently large, use of an IROS configuration in an ITE may be helpful, but real-ear measures should be used to verify the amount of high-frequency insertion gain. Similarly, these measures should be used to confirm the amount of low-frequency insertion gain reduction because this cannot be extrapolated simply on the basis of vent size.

Although behind-the-ear (BTE) hearing aids might be more suitable in achieving these objectives, Surr and Hawkins (1988) reported that 73% of new users selected ITE hearing aids even when the advantages of BTE hearing aids were explained. Despite the traditional cosmetic concerns regarding the BTE style, it should be noted that (1) there has been a significant size reduction in BTEs over the years; (2) less conspicuous earmolds are now available; and (3) very small, cosmetically appealing, so-called entry-level hearing aids (e.g., the ReSound Avance) might be appropriate for many type I NIHL patients. Figure 6–6 is a photograph of this instrument.

Programmable and digital hearing aids have rendered issues of hearing aid style
less problematic than in the past. BTEs may continue to best serve the acoustic needs of many of the NIHL patients, but, again, features such as multiband frequency-shaping and digital noise and feedback reduction enable greater flexibility in choosing hearing aid style. Also, if occlusion can be overcome, completely-in-the-canal (CIC) hearing aids might meet both the cosmetic and acoustic needs of some of these patients with their high-frequency emphasis and decreased gain requirements because of closer proximity to the eardrum.

**Binaural Amplification**

For NIHL patients, as with other listeners with sensorineural hearing loss, word recognition scores measured in quiet, sound-treated rooms often are not sensitive enough to prove or disprove the notion of binaural superiority with regard to hearing aid use (Carhart, 1958). Even so, preference declarations and anecdotal reports of enhanced laterality and more comfortable listening through binaural systems abound (Balfour and Hawkins, 1992). A variety of factors might account for this subjective preference. Elimination or minimization of head shadow (the reduction in signal intensity from the side of the head that is opposite the signal) is particularly important for listeners with high-frequency hearing loss, such as the NIHL population. Improved localization and better balance of sounds results from hearing sounds from both sides. A central release from masking, termed binaural squelch (Koenig, 1950), may also be operative, resulting in better hearing in noise. With binaural loudness summation, absolute binaural thresholds are 2 to 3 dB better than monaural thresholds (Haggard and Hall, 1982). At suprathreshold levels at which listeners receive amplified sound, summation increases by as much as 6 to 10 dB (Reynolds and Stevens, 1951). Thus, a hearing aid user can achieve the same loudness perception from binaural hearing aids set at lower gain than with a monaural aid. This may greatly reduce feedback problems. One might reason that if binaural stimulation sounds louder than monaural stimulation, it would be necessary to limit the maximum power of a hearing aid to keep it from exceeding the patient’s LDL. However, Hawkins (1986) found that when subjects were asked to match the loudness of binaural and monaural stimuli, this summation effect occurred, but the subjects reported no reduction in binaural LDLS versus monaural LDLS. In fact, most indicated that the binaural stimuli could be more intense than the monaural stimuli before it produced discomfort. It follows that the dynamic range of listening is greater for binaural listening than for monaural listening.
Other factors to consider in choosing binaural versus monaural amplification include (1) the possibility of tinnitus reduction regardless of a perceived dominant side, because it raises the possibility of neural stimulation to as much of the auditory cortex as possible (Sweetow, 2001), and assures symmetrical stimulation of the central auditory pathways, preventing development of abnormal receptive fields (Jastreboff, 2000); (2) the legal implications of the potential deprivation of an unaided ear, as noted in the earlier discussion on the potential negative consequences of auditory deprivation of an unaided ear; and (3) the easier adjustment of binaural to monaural than monaural to binaural amplification.

Thus, the general rule is that, unless there is a compelling reason to avoid amplification of one ear (e.g., that the ear is normal, or is very severe and unable to benefit from a hearing aid, or if a medical contraindication exists), the standard should be trial with binaural amplification. (Chapters 8 and 9 contain an extensive discussion on the issue of binaural amplification.)

Verification and Validation Considerations
Verification of a successful hearing aid fitting for patients with NIHL should include assessment of word recognition and judgments of sound quality, as well as real-ear insertion and saturation responses and/or functional gain measures. Validation of a successful hearing aid fitting for patients with NIHL includes subjective scaling.

Assessment of word recognition and judgments of sound quality
The primary goal of amplification is to enhance communication. For some hearing aid users, this corresponds with an improvement in word recognition. For others, particularly those with type I and some with type III NIHL, the goal may be to ease listening effort. Both word recognition scores and judgments of sound quality should be obtained in quiet and in noisy environments. The ineffectiveness of word recognition testing in quiet for distinguishing one aid from another has previously been discussed. In addition, Beck (1991) discussed the limitations of word recognition tasks in noise to differentiate precisely among hearing aids. This is especially true when one utilizes monosyllabic word lists that have questionable face validity. Word recognition testing, however, can illustrate differences in hearing aid performance over a range of input intensities and SNRs (Valente et al, 1999b).

A number of attractive alternative speech procedures have been proposed. The use of adaptive speech measures (i.e., maintaining a certain subjective intelligibility level, such as 50%) of connected discourse in various SNRs may be helpful in avoiding ceiling effects (e.g., word recognition scores that are too high to show improvement), as may be characteristic of type I NIHL patients. The SPIN test (Bilger et al, 1984) utilizes high-predictability and low-predictability items and can be effectively used. It has been shown, for example, that the need for extending the frequency response into the high frequencies becomes more apparent for low-redundancy items. Cox and Alexander (1991) proposed use of the Connected Speech Test (Cox et al, 1989), but cautioned that a variety of input levels should be utilized in light of the fact that subjective preference of hearing aid-processed sound might change as a function of intensity level. Other adaptive speech measures gaining in popularity include the HINT (Nilsson et al, 1994) and the Speech-in-Noise (SIN) test (Fikret-Pasa, 1993). These tests can also be applied when assessing the effectiveness of directional and multiple microphones, and when illustrating binaural superiority and ease of listening.

Mueller and Palmer (1998) described a tool that compares the loudness judgments of an aided listener with those of an unaided normal listener. Approaches such as this can help not only in troubleshooting patient complaints, but also in counseling listeners unaccustomed to amplified signals.

Sound quality judgments can be obtained in a variety of ways. For example, Bentler and Duve (2000) utilized paired comparisons to rate speech clarity in various background noises, including street noise (0-dB
SNR), cafeteria noise (+8-dB SNR), and office noise (+8-dB SNR). Subjects rated (on a 0 to 10 scale) the clarity of speech, which they defined as “sounds clear, distinct, and pure instead of blurred and distorted.” Quality judgments also were used in a study by Arlinger et al (1998). It might be helpful to assess quality judgments at multiple intensity levels (e.g., 55-, 70-, and 85-dB SPL) and for multiple azimuths (e.g., 0, 45, 90, 135 degrees, and diffuse). Informal quality judgment assessments can be designed based on the listening situations encountered by the listener. For some listeners, quality judgments of music may be important; for others, the focus can be on family members’ voices, or even the sound of a computer keyboard. Ratings can be made using a numerical scale or a descriptive semantic differential (e.g., clear, harsh, tinny, hollow; or excellent, fair, poor, very poor).

**ASSESSMENT OF REAL-EAR INSERTION AND SATURATION RESPONSE AND FUNCTIONAL GAIN**

It is difficult to obtain a target match in the 3000- to 4000-Hz region (using any of the prescriptive formulas) due to the loss of the natural resonance caused by complete or partial occlusion of the ear canal. Because audibility in this region is so critical to word recognition, it is advisable to begin programming the hearing aid so that target gain (or adequate consonant audibility and/or comfort) is achieved at 3000 Hz and, one hopes, at 4000 Hz. Then, the low-frequency gain and slope can be set as desired. This can be accomplished either by changing the actual gain in the lower frequency band, or by raising the transition or crossover frequency in multiband hearing aids. Decisions involved in setting low-frequency slope must incorporate both subjective (the patient’s description of his acoustic environments) and objective (upward spread of masking and other masking functions) information. Priorities must be established regarding the need for low-frequency reduction (perhaps to minimize certain background noises) versus the need for second formant and formant transition information. In addition, one must take into consideration that requesting additional high-frequency gain in instances when the REUG shows a particularly large resonant peak may be fruitless because the manufacturer may have already incorporated the maximum gain possible into that circuit.

There has been a disturbing trend for some manufacturers to claim that real-ear measurements are not valid for their particular hearing aids. (The issue of real-ear measures is presented in Chapter 3.) This caution might be expressed in part because of the depth and nature of the coupling to the ear. Dirks and Kincaid (1987) noted the difficulties encountered in accurately assessing the very high frequencies depending on probe microphone placement. Additionally, it is true that prescriptive formulas were developed initially for linear hearing aids and that several unresolved questions remain concerning how target levels should vary as a function of input level and audiometric configuration. Potential shortcomings of target insertion gain for patients with NIHL notwithstanding, certain real-ear measures remain vital objectives of the hearing aid fitting verification procedure. Auditory mapping procedures such as those proposed by Seewald et al (1995) or Loven (1991) utilize the real-ear aided response (REAR) and the real-ear saturation response (RESR) to ensure audibility and to ensure that the individual’s LDL is not exceeded. Similar to the rationale for using multiple input levels for word recognition and sound quality assessment, the use of advanced compression techniques also requires that the REAR be measured using several different input levels. Prescriptive approaches such as the NAL nonlinear version 1 (NL1) (Dillon et al, 1998) and desired sensation level (input/output) (DSL [i/o]) (Seewald et al, 1995) address some of these issues. Etiologic factors (e.g., duration of the hearing loss, cochlear versus retrocochlear versus conductive components, etc.) might impact these prescriptive formulas. For example, re-
call the earlier discussion that some patients with type I NIHL prefer use-gain equivalent to only 20% of the hearing loss (the issue of prescriptive formulae for nonlinear signal processing is presented in Chapter 1; the issue of corrections for conductive hearing loss is presented in Chapter 10).

More recently it has been reported that certain DSP devices with noise reduction will be “tricked” because the hearing aid will correctly interpret the input stimulus as noise (Kuk and Ludvigsen, 1999). The use of alternative stimuli, such as random bursts or the International Collegium of Rehabilitative Audiology (ICRA) composite signal, that more closely approximate the temporal characteristics of speech can circumvent these problems (Frye, 2000). If these techniques are not available, one can use stimuli that are shorter than the critical duration required to activate the noise reduction process, or turn off the noise reduction and other DSP algorithms that would alter the true amplification characteristics if left on during testing.

If probe microphone measures are not available, one can utilize functional gain for verification. The results obtained using functional gain may not agree with those obtained with probe microphone measures. Given the very low knee points found in many current hearing aids, insertion-gain measures, typically performed with a 60- to 70-dB input, force the aid into compression. This will produce a greater amount of functional gain than insertion gain. It is important to note that functional gain with linear hearing aids can reflect gain for conversational levels; however, functional gain for compression hearing aids reflect the aid’s maximum gain at a low input level, and gain for conversational and loud input is not reflected (Kuk, 2000). Also, when testing functional gain for many DSP instruments, it is essential to use short burst stimuli, to space the presentations far enough apart that functional gain is not reduced due to compression or noise reduction processing, and to test using ascending steps when the stimulus is above the compression threshold of the stimulus. Other limitations of functional gain measures include variation in test-retest results and underestimation of functional gain due to masking from ambient noise if any unaided thresholds are near normal (as would be found in type I NIHL). If the limitations of functional gain are taken into consideration, these measures can provide valuable verification information, not only in lieu of but also in addition to that obtained with probe microphone measurements.

**SUBJECTIVE SCALING**

In view of the limitations of the verification procedures just discussed, and the fact that quality and comfort of listening might be the most important factors determining success with amplification for NIHL patients, it is important to validate aided benefit with self-assessment scales. Several scales have been developed for this purpose (Walden et al, 1984; Cox and Gilmore, 1990; Cox and Alexander, 1991, 1999; Cox et al, 1991). Whether one chooses to utilize any of the published hearing aid benefit or satisfaction questionnaires, or a tool like the Client Oriented Scale of Improvement (COSI) (Dillon et al, 1997) that directly addresses the needs of a particular patient, subjective scaling is certainly useful (Chapter 5 discusses outcome measures). It is important to remember that NIHL patients will not benefit from hearing aids that are not worn, so the bottom line is how an individual feels about the prescribed amplification, not how an individual scores on a battery of clinical measures in a controlled laboratory environment. Listener preference (both to other voices as well as to their own) might represent the highest order of sensitivity and certainly represents the most critical factor in motivating patients to use hearing aids. The needs and preferences of the individual take precedence over any single assumption based on research data. A potentially useful tool might be the Hearing Aid Selection Profile (HASP) developed recently by Jacobson et al (2001). Although most subjective scales are used in postfitting verification, the HASP was developed for administration prior to the hearing-
aid evaluation/consultation to assist in the selection of the most appropriate hearing aids based on a consideration of the patient’s perspectives in areas key to the acceptance of amplification (i.e., motivation, expectations, costs, appearance, attitudes about technology, physical limitations, communications needs, and lifestyle).

Conclusion

Significant technologic advances have been made in providing amplification for patients with NIHL. The use of programmable and digital hearing aids having accessible “comparative” functions should continue to expand as audiologists utilize this flexibility in determining the electroacoustic parameters for each patient. The use of multiple microphones, multichannel compression, adjustable crossover frequencies, fine-tuning of gain in narrow frequency regions, and digital feedback reduction will likely prove to be of great value as superior algorithms and fitting strategies continue to evolve.

Audiologists should continue efforts to develop concise and accurate clinical measures of both psychoacoustic skills and limitations of the individual patient. Among these could be measures of temporal masking, frequency-resolving power, upward spread of masking, loudness growth, and speech detection in noise. It is anticipated that the use of otoacoustic emissions, particularly distortion products, in conjunction with noise-notch masking tests might be of assistance in determining the functional status of hair cells so that the benefits of providing gain in certain frequency regions can be ascertained on an individual basis.

Technological advances notwithstanding, patient counseling regarding realistic expectations is critical. Certainly it is reasonable for the patient to expect enhanced fitting flexibility and efficacy, improved sound quality, better hearing for soft sounds while maintaining comfort for loud sounds, and fewer problems with feedback. It is, however, not realistic to expect a perfect fit, longer hearing aid life, true high-fidelity sound quality (the ear is still impaired), elimination of feedback, perfect word understanding (particularly in noise), and that there will be no need for training or adaptation. It should be explained to the patient that (1) amplification might not provide immediate or obvious benefit, and (2) an overall improvement in word recognition might not easily be realized. The improvement in word recognition might be subtle and limited to fricatives and affricatives. Most important, it is vital that patients with NIHL be told that ease of communication is a quality of life issue and that this might be the most important benefit to be derived from amplification.

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