CHAPTER 5

Principles of High-Fidelity Hearing Aid Amplification

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The 20th century has seen the emergence of many technologies related to the amplification of sound for the listener with hearing impairment. These technologies have ranged from a turn-of-the-century, nonelectric ear trumpet to the fully digital hearing aid of the 1990s. To determine if recent hearing aid technologies (e.g., wide dynamic range compression and digital signal processing) provide more benefit to the listener than older technologies, Bentler and Duve (1997) examined hearing aid processing strategies over the last 50 years, as well as an ear trumpet from the turn of the century. The specific technologies evaluated included (1) an 1800's ear trumpet or speaking tube, (2) a linear hearing aid from the 1930s with peak clipping to limit output (body aid), (3) a hearing aid using wide dynamic range compression only in the high frequencies (K-ampTM), (4) a two-channel wide dynamic range compression hearing aid (ReSoundTM), (5) a digital signal processing (DSP) hearing aid from Oticon (DigifocusTM), and (6) a DSP hearing aid from Widex (SensoTM). Speech recognition was assessed in quiet and in noise at multiple presentation levels. Figure 5–1 shows the speech recognition results in quiet at two presentation levels (58 and 78 dB SPL). The modern technologies, K-amp, ReSound, and the two DSP hearing aids, performed essentially equally at the two levels. In addition, the performance levels were high for these modern technologies with average scores ranging from 76 to 83% for both presentation levels using the CUNY Sentence Test (Levitt & Neuman, 1990). The older technologies, linear processing with peak clipping (body aid) and the speaking tube, did not perform as well. The nonelectric speaking tube did not provide enough gain for the low input level (58 dB SPL), and the linear hearing aid, which distorted due to peak clipping at the 78 dB SPL input level, showed reduced performance at that level.

The good performance of modern instruments comes from the fact that they have eliminated the major deficits of older designs: distortion, very narrow bandwidth, and peaks in the response. From the standpoint of word-recognition scores, it is unlikely that any further improvement can be obtained with further improvements in fidelity.

From the standpoint of sound quality, however, further improvement is possible. The requirements for optimal high fidelity are discussed below. Analog hearing aids can now meet all of these requirements, but none of today's digital hearing aids do, mostly because of a limited dynamic range (effectively 13 to 14 bits equivalent in a 16 to 18 bit high-fidelity world) and limited bandwidth (7–8 kHz in a 16 kHz world). As digital aids continue to improve, such requirements will no doubt be met.

There is one problem that cannot be solved with further increases in fidelity: the difficulty of hearing in noise. Hearing aid users typically
Figure 5-1. Average speech recognition scores in quiet at 58 and 78 dB SPL for the technologies evaluated in the Bentler and Duve (1997) study. (Data are from Progression of hearing aid benefit over the twentieth century, by R. A. Bentler and M. R. Duve, 1997, poster presented at the American Academy of Audiology Convention, Ft. Lauderdale, FL. Used with permission.)
have a 5 dB deficit in their ability to hear in noise (Killion, 1997b). Bentler and Duve (1997) tested 20 study participants with each hearing aid described above in noise at four signal-to-noise ratios (SNRs) and four presentation levels (from 55 dBSPL to 95 dB SPL). The results for the 0 dB SNR are shown in Figure 5-2. No technology provided much assistance at 0 dB SNR, although individuals with normal hearing can typically understand 30% or more of the words. Solutions to the SNR problem involve increasing the SNR directly. Methods for achieving this goal are discussed below.

### THE BASIC REASONING

#### What Is High Fidelity?

Our first problem is to define a high-fidelity hearing aid. An object that reproduces high-fidelity sound should be acoustically and, more important, subjectively transparent. That is, listeners should receive the same auditory sensation with the high-fidelity system interposed between them and the original source of sound as they would receive listening directly to the sound.

![Figure 5-2. Average speech recognition scores in noise for the technologies evaluated in the Bentler and Duve (1997) study. This figure shows the results for 0 dB SNR at four presentation levels. (Data are from Progression of hearing aid benefit over the twentieth century, by R. A. Bentler and M. R. Duve, 1997, poster presented at the American Academy of Audiology Convention, Ft. Lauderdale, FL. Used with permission.)](image-url)
Interestingly enough, if it weren't for the complications of the hearing impairment that creates the need for the hearing aid in the first place, a hearing aid worn on the head would stand a better chance of providing perfect fidelity than any other means of sound reproduction. Near-perfect subjective fidelity is difficult to achieve with recorded sound. Loudspeaker reproduction is heavily influenced by the acoustics of the listening room: Neither the directional cues nor the frequency spectrum experienced in a live concert-hall performance are preserved accurately. Headphone listening to proper binaural recordings offers the best theoretical approach, but because of the lack of head-motion cues to source localization, most listeners experience, for example, the orchestra playing in the middle of their heads. Because the hearing aid is worn in "real time," all head-motion cues to sound localization are preserved. Because the hearing aid is worn in real space, all directional cues can be preserved as well.

But hearing impairment is what creates the need for a hearing aid in the first place, and it must be accommodated. Barford (1972) stated the goal well: "One could say that the ideal hearing prosthesis was an instrument which gave the wearer the same perception of external stimuli as a normal hearing person would have." The signal processing required to achieve this result in the regions of hearing loss may not be known or even possible, but Barford's definition immediately suggests a direct application of the "If it ain't broke don't fix it" principle: The hearing aid should do absolutely nothing (i.e., be acoustically transparent) for sounds that fall within a region of normal hearing. In particular, this principle implies that the hearing aid should provide no amplification within a region of normal hearing.

This "don't-amplify-where-there's-no-loss" principle is well understood in the frequency domain. We know better than to use a closed earmold for individuals who have normal low-frequency hearing, for example, because they would inevitably object to the sound quality, partly because the imperfections in the hearing aid are interposed between them and the natural low-frequency sounds that they can hear quite well without the hearing aid. In addition, the closed-mold "occlusion effect" will make one's own voice sound abnormal.

This same principle has been more recently applied in the amplitude domain. Amplification is certainly necessary for quiet and moderately loud sounds. There is substantial evidence, however, that many individuals with mild or moderate hearing impairment, have normal hearing for high-level sounds. Following our don't-amplify-where-there's-no-loss principle, we conclude that whenever a person has normal or near-normal hearing for high-level sounds, the hearing aid should do absolutely nothing: It should be so acoustically transparent that it subjectively disappears. This idea was subject to much more debate in 1988 than it is today, now that wide-dynamic range compression (WDRC) circuits are common.

The term "unity acoustic gain" or simply "unity gain" (i.e., 0 dB gain) is used to describe the condition in which the hearing aid provides neither amplification nor attenuation of the incoming sound. For quiet sounds, however, a practical hearing aid must provide enough gain to make those sounds audible to an individual with a hearing loss. The amplification compensates for the loss of sensitivity to low-level sounds, or the loss of sensitivity that is measured in a threshold audiogram.

If the hearing loss for low-level sounds is frequency dependent, as in the case of what is commonly called a "sloping hearing loss," more low-level amplification may be required at some frequencies than at others. This requirement, in combination with the requirement for unity acoustic gain at some or all frequencies for high-level sounds, implies that the frequency response of the hearing aid must depend on the sound level. A WDRC circuit meets this requirement. The WDRC circuitry required to provide the variable gain and variable frequency response amplification dictated by the reasoning of the previous paragraphs should operate as unobtrusively as possible.
A Brief History of High-Fidelity Hearing Aids

A unity-gain high-fidelity sound reproduction system (as judged by listeners with normal hearing) is the sensible starting point for a high-fidelity hearing aid design, with appropriate signal processing (e.g., level-dependent gain and frequency response) added to compensate for the hearing loss. The quest for this type of high-fidelity hearing aid began in the late 1970s with the design of hearing-aid-sized microphones that were used in recording and broadcast studios. In 1979, Killion (1979a) demonstrated that hearing aids could be assembled with both objective frequency response accuracy and subjective fidelity ratings comparable to those of highly regarded loudspeaker systems. Hearing-aid earphones are now used in the ER-4 series of 16 kHz bandwidth high-fidelity earphones, which have won high praise from audiophile reviewers.

By the end of the 1970s, only two things were missing from hearing aids: a low-distortion power amplifier small enough and with a low enough battery drain to be practical for use in the smaller hearing aids and a broadcast-quality input amplifier that could handle loud voices and live orchestra concerts without distortion. Both amplifiers became a reality in tiny integrated circuit chips introduced in the late 1980s. The power amplifier is the Class-D chip developed by Etymotic Research for Knowles Electronics. The Class D amplifier will produce 110 to 115 dB maximum undistorted output, yet its internal amplifier idles at about 0.17 mA, a small fraction of the idling current of a typical hearing-aid power amplifier. The broadcast quality input amplifier was developed by Etymotic Research with the help of a $500,000 grant from the National Institute on Aging. This K-Amp™ input amplifier chip amplifies only quiet sounds (Killion, 1993). To this end, an automatic circuit operates an electronic volume control to make quiet sounds audible and a tone control to provide treble boost for quiet sounds. Loud sounds that present a problem for most hearing-aid wearers (e.g., dishes clattering, paper crunching, wind howling, people shouting) pass through without amplification just as if the hearing aid was not there. Amplification for loud sounds is available to the user if he or she chooses to use it, but it is generally not required.

Who Are the Beneficiaries?

High-fidelity hearing aids are not for everyone. There are individuals whose hearing impairment is so great that even amplified speech is not clear, especially in the presence of competing noise. Similarly, individuals with no useful high-frequency hearing will derive little benefit from wide frequency range sound reproduction (Hogan & Turner, 1998; Rankovic, 1991; Skinner, 1980). Their use may be troublesome to others because of the high-frequency feedback squeals that hearing aid users cannot hear. (One of the successful uses of low-pass filter earhooks has been high-power hearing aids and closed earmolds, for individuals with profound loss and a “corner audiogram,” to eliminate the unheard-squeal problem.) Individuals with the most common hearing impairment (a mild sensorineural hearing loss that appears to be restricted to a loss of sensitivity for low-level sounds, with normal hearing for high-level sounds) are the primary beneficiaries of high-fidelity hearing aids.

The psychoacoustic and physiological evidence showing that many individuals with hearing impairment have essentially normal hearing for high-level sounds has been available for some time. Punch (1978) found no statistical difference between hearing aid sound-quality judgments obtained from 10 subjects with normal hearing and 10 subjects with sensorineural hearing loss, even though some subjects had moderate-to-severe losses at high frequencies. The subjects with sensorineural hearing loss studied by Lindblad (1982) were as good at detecting nonlinear distortion in high-level material as normal subjects were. Some of Toole’s (1986) listeners had mild-to-moderate high-frequency hearing losses and still exhibited excellent reli-
bility in their fidelity ratings of high-fidelity loudspeakers.

Steinberg and Gardner discussed such individuals in 1937, when they observed that many of those with impaired hearing had essentially normal hearing for high-intensity sounds, but a loss of normal sensitivity for low-intensity sounds. This phenomenon is commonly referred to as loudness recruitment. Figure 5–3 (Scharf, 1978) illustrates this phenomenon for a “typical listener with a 40 dB hearing loss due to a cochlear impairment.”

A substantial amount of evidence has accumulated since the Steinberg and Gardner study to indicate that many (but not all) individuals with mild-to-moderate cochlear impairments may have essentially normal high-level hearing. In addition to pure-tone loudness per-

![Figure 5-3](image)

**Figure 5-3.** Loudness of a 1000 Hz tone as a function of loudness level for a normal listener, a typical listener with a cochlear impairment (and recruitment), and the typical listener with a conductive impairment. (From "A model of loudness. A summation applied to impaired ears," by B. Scharf, 1978, *Journal of the Acoustical Society of America*, 40, 71–78. Reprinted with permission.)
ception, the following attributes of hearing have been found to be within normal limits at sufficiently high-intensity levels, even in the presence of mild-to-moderate cochlear impairment:

1. Frequency selectivity as determined from the “Fletcher Critical Band” that can be inferred from tone-in-noise masking experiments (Jerger, Tillman, & Peterson, 1960; Palva, Goodman, & Hirsh, 1953)

2. Frequency selectivity as determined from psychophysical tuning curves (McGee, 1978)

3. Frequency discrimination (dF) (Gengel, 1973)

4. Loudness summation for complex sounds (with cochlear impairment below 50 dB) (Scharf and Hellman, 1966)

5. Loudness discrimination (dlI) (Scharf, 1978)

6. Localization abilities (see the summary given by Scharf, 1978)

In addition to the psychophysical data, recent evidence indicates that the high-level electrical potentials in the cochlea can be normal in some cases of mild cochlear impairment. The whole-nerve action potential recorded in laboratory animals (Wang & Dallos, 1972) or recorded from the ear canal in humans (Berlin & Gondra, 1976) is often normal. Indeed, both the wave form and latency of the entire auditory brainstem-evoked response may appear entirely normal at high levels in some individuals. Similarly, data obtained on laboratory animals with drug-induced outer hair cell damage (Dallos et al., 1977) indicate that it is possible to have normal bandwidth for both the psychophysical tuning curve and the classical single-unit tuning curve obtained from fibers of the auditory nerve.

A discussion of the physiology of hearing impairment is beyond the scope of this chapter, but the function of the outer hair cells may be to improve the low-level sensitivity of the ear, while the primary function of audition appears to rest on the inner hair cells. Postmortem hair-cell counts on patients with hearing impairment who had good speech recognition in noise typically show hair-cell loss restricted to outer hair cells. Thus, the loss of outer hair cells may permit normal or near-normal hearing for high-level sounds as long as most of the inner hair cells function properly.

**Adaptation and Good Counseling**

Even today, any observant dispenser realizes that the most important factor in a successful hearing aid fitting may have nothing to do with the aid’s electroacoustic characteristics. Given appropriate counseling and an established trust between the individual with hearing impairment and the dispenser, any hearing aid that makes previously unheard sounds audible can provide a successful fitting. The old advice, “wear it a few weeks and you’ll get used to it,” is as true—and as useful—today as ever.

The process of “getting used to it” can be so successful that a change in the hearing aid creates a problem. Anyone servicing hearing aids is familiar with complaints that may result from restoring the frequency response of a damaged hearing aid to its original configuration, and from the use of “loaner” hearing aids during an extended repair period (Goldberg, 1965). Even when the general frequency responses of the original and loaner aid appear similar, it is not uncommon for a user to complain initially of difficulty understanding speech only to find that after a 2- or 3-week period, the loaner seems much better than the repaired original aid.

Even when change is ostensibly an improvement, it may not be welcome. Smoothing the response (i.e., by adding damping to the earmold tubing) for a long-time hearing aid wearer can result in a negative evaluation. Similarly, a low-distortion hearing aid may not sound loud enough to a user accustomed to hearing the severe distortion products that accompanied overload in an old peak-clipping hearing...
aid. The extraordinary ability of humans to adapt to even badly degraded sensory inputs has been a merciful boon to many hearing aid wearers but can cause unmerciful trouble to the conscientious dispenser. The success of hearing aids decades ago, when the best that could be done in terms of sound quality, intelligibility, and convenience was poor indeed, attests to the ability of the central nervous system to adapt.

High-fidelity hearing aids as described in this chapter are available today to the hearing aid user. Although these devices have been available for 10 years, how these aids differ from traditional linear amplification is still often misunderstood. In the next section, the design guidelines for high-fidelity amplification will be reviewed. We feel this review is important to understand the capability of high-fidelity amplification and the limitations of hearing aids that do not meet these desired design guidelines.

**Design Guidelines for High-Fidelity Hearing Aids**

In this section, we provide engineering guidelines for the design of hearing aids that can be reasonably expected to be judged as high fidelity by those with mild or moderate hearing impairment.

**Frequency Response Guidelines**

Recall that, at high levels, the hearing aid should deliver the same sound pressure to the listener's eardrum—at all frequencies and for all angles of incidence of the original sound—that he or she would receive without the hearing aid interposed. This requires consideration of the relationship between the sound pressure in a sound field and that found at the normal unaided eardrum.

Nature provides each of us with a natural hearing aid in the form of external-ear resonances and "horn" action, which combine to produce substantially greater eardrum sound-pressure levels than those present in an oncoming sound field. At the roughly 2.7 kHz resonance frequency of the outer ear, the gain amounts to 15 to 20 dB (Wiener & Ross, 1946).

The pressure at the microphone inlet of a head-worn hearing aid, on the other hand, will generally be only about 5 dB greater than that in the sound field, depending somewhat on the exact location of the microphone (Kuhn & Burnett, 1977; Madaffari, 1974). Thus, the hearing aid must provide some 10 to 15 dB of acoustic gain in the 2.7 kHz frequency area to compensate for the gain provided by the external ear before it was blocked by the plastic earmold. In contrast, little compensation is required at low frequencies where the eardrum pressure and the pressure available to the hearing aid microphone are essentially the same.

Insertion gain is the ratio of eardrum pressure produced by a hearing aid to the eardrum pressure produced without the hearing aid (Dalsgaard & Jensen, 1974). Expressed in dB, the insertion gain of a hearing aid is the difference between aided and unaided eardrum sound-pressure levels. At high levels, the hearing aid should have 0 dB insertion gain at all frequencies (a flat insertion gain frequency response).

**Coupler Response for Flat Insertion Gain**

For engineering purposes, the required frequency response tailoring is best defined in terms of the coupler response of the hearing aid. In addition to the factors discussed above (loss of external-ear resonance effects and hearing aid microphone location), the COupler Response for a Flat Insertion Gain (CORFIG) curve will be influenced by the choice of couplers, the selection of reference sound-field conditions, and differences among individuals in terms of their external ear resonances, eardrum impedances, and so forth.

A great deal of simplification is provided if we ignore individual differences and choose a
manikin of average anthropometric dimensions and a coupler ("occluded-ear simulator") that approximates the acoustic impedance of an average ear. The KEMAR manikin (Burkhard & Sachs, 1975) meets the first requirement, while the modified Zwischlocki (1970) coupler meets the second requirement up to 7 or 8 kHz (Sachs & Burkhard, 1972). Unpublished experiments comparing eardrum-pressure response on real ears to the response measured in the Zwischlocki coupler were conducted at Eymotic Research during the development of the ER-4 16 kHz bandwidth high-fidelity earphone. These experiments indicated that the Zwischlocki coupler provides a reasonably good representation of the ear in the 8 to 16 kHz region.

Figure 5–4 shows the Zwischlocki-coupler CORFIG curves from Killion and Monser (1980) for three microphone locations and three sound-field reference conditions, based on measurements with a KEMAR manikin. The inset drawings illustrate the three different locations for the microphone inlet, for OTE (behind the ear), ITE (in the ear), and CIC (completely in canal) hearing aids. The curves for the ITE aid were obtained for a microphone inlet approximately flush with the plane of the pinna because at the time the data were obtained, most ITE aids filled the entire concha. (Throughout this chapter, we will use the term OTE to mean a behind-the-ear case in which the microphone is located over, and not behind, the ear.)

The Role of the Pinna

As illustrated in Figure 5–4, the exact shape of the CORFIG curve generally depends on the direction from which the sound is coming. The effect of the pinna flange, for example, is to increase eardrum pressure by a few dB at high frequencies for sounds arriving from the front and to reduce substantially the eardrum pressure for high-frequency sounds arriving from the rear "due to interference between the direct wave and a scattered wave from the edge of the pinna flange" (Shaw, 1974). For microphone locations outside the pinna, such as the BTE, forward-looking-inlet location discussed above, little of this directional dependence is contained in the sound available to the microphone. This loss was graphically illustrated in the data of Berland and Nielsen (1969), who compared the sound pressure available to microphones located behind, over, and in the ear for sound fields at six angles of incidence.

Even when the microphone is in an ITE hearing aid, the presence of the earmold filling the concha substantially reduces the effectiveness of the pinna. Only when the microphone inlet is located directly in front of the blocked ear canal entrance, as with a CIC construction, are the directional effects preserved in their entirety.

With regard to auditory localization, it is clear than any of the microphone locations used in binaural, head-worn hearing aids will preserve the basic interaural time and intensity difference so important to binaural localization (Licklider, 1949) and the changes in interaural phase and intensity caused by head motion that are so important to the externalization of the sound (Wallach, 1940). What is not so clear is the relative importance of the pinna and concha to everyday localization, auditory spatial perception, and the binaural squelching of noise and reverberation.

Informal experiments (such as taping the pinna tightly to the side of the head) have provided some evidence that the pinna and concha provide relatively weak cues to localization, compared to those provided by head movement and interaural time and intensity differences. Such a conclusion is consistent with anecdotal evidence indicating that many individuals who need artificial pinnae wear them only on social occasions, that is, primarily for cosmetic reasons.

Reference Sound-Field Condition

As Figure 5–4 indicates, it is impossible to design a conventional ITE or BTE aid that will have a perfectly flat frequency response for the user (i.e., an insertion gain that is frequency independent) at all angles of incidence. The
Figure 5-4. Required Zwislocki coupler response to produce flat insertion gain (CORFIG) for OTE, ITE, and CIC hearing aids under three sound field conditions. Note that only the CIC CORFIG is essentially independent of where the sound is coming from. (From Acoustical Factors Affecting Hearing and Performance by M. C. Killion and E. L. Munser, 1980, Baltimore, MD: Park Press. Copyright 1980. Reprinted with permission.)
acoustical effects of the unencumbered outer ear are not duplicated in either of the microphone configurations. Where speech discrimination in face-to-face (near-field condition) listening situations is the only consideration, the appropriate reference condition might be a 0°-incidence sound wave.

Where sound quality is the dominant criterion, other considerations apply. In home and concert listening, the reflected energy substantially exceeds the energy arriving directly from the sound source (Olson, 1967). Under those circumstances, the appropriate design compromise for a high-fidelity hearing aid would appear to be a flat insertion gain for random-incidence sound. This conclusion is consistent with the results of the psychoacoustic experiments reported by Schulein (1975).

The CIC aid requires no compromise between conflicting frequency response requirements because the coupler response that provides a flat insertion gain response for 0°-incidence sound provides a flat insertion gain response for all angles (and thus, trivially, for random-incidence sound as shown in Figure 5–4).

**Bandwidth Requirements**

After considering the available information on (a) hearing sensitivity and frequency limits of hearing for typical listeners, (b) measurements of the discomfort level of sound, (c) measurements of room noise in a wide variety of locations, and (d) measurements of the frequency limits in the maximum and minimum levels of speech, orchestral music, and various instruments of the orchestra, Fletcher (1942) concluded that “substantially complete fidelity in the transmission of orchestral music is obtained by use of a system having a volume range of 65 dB and a frequency range from 60 to 8000 cycles per second.” Olson (1957) concluded that “the reproduction of orchestral music with perfect fidelity requires a frequency range of 40 to 14,000 cycles and a volume range of 70 dB.” Both judgments were based on similar data, primarily those of Snow (1931), and the differences reflected the level of fidelity required. Snow’s listeners gave a judged quality rating slightly in excess of 90% to a system in which the frequency range extended from 60 Hz to 8000 Hz, compared to unrestricted range reproduction. With a 40 Hz to 14,000 Hz range, the quality rating was close to 100%. Tests reported by Muraoka, Iwahara, and Yamada (1981) on the audio bandwidth requirements for unlimited digital recordings indicated that only with a 14 kHz cutoff frequency (but not with a 16, 18, or 20 kHz cutoff) could any of their listeners reliably detect the bandwidth restriction.

A frequency response extending from 60 Hz to 8000 Hz thus appears to be a reasonable goal for a high-fidelity hearing aid. This was confirmed in fidelity rating experiments undertaken by Killion in which three panels of listening judges (25 “man-on-the-street” listeners, 5 “golden ear” listeners, and 6 trained listeners) each rated a pair of experimental OTE aids with 8 kHz bandwidth as being comparable in fidelity to a pair of good studio monitor loudspeakers with 20 kHz bandwidth. (Figure 5–5, shows their frequency response curves, labeled OTE and MS, respectively. The irregularity in the frequency response of the loudspeakers was apparently judged to be as important a defect as the 8 kHz bandwidth limitation of the BTE aids.) The 8-kHz-bandwidth hearing aids were rated much higher in fidelity than a pair of “Popular Headphones” (Koss PRO-4AA) in which exaggerated bass and treble response has been widely noted. The good fidelity ratings for the BTE aids held up under four different selections of program material, as shown in Figure 5–6, including a piano trio selection chosen because the drummer’s brush-on-cymbal sound provided substantial energy above 10 kHz.

From these data, it appears reasonable to conclude that restricting the bandwidth of a hearing aid to 8 kHz, or perhaps a bit less, is not likely to be an important limitation to high-fidelity sound reproduction as judged by the hearing aid wearer. Of course, there is no harm in approaching perfection, and a 16 kHz bandwidth is also practical as demonstrated by the experimental ITE aids whose fidelity ratings are also shown in Figure 5–6.
Figure 5–5. Fidelity ratings obtained with different program materials for experimental ITE hearing aids; experimental OTE hearing aids; Electro Voice Sentry V Studio Monitor Speakers (MS); Koss PRO-4AA Popular Choice Phones (PP); Simulated Speech Audiometer with TDH-39 earphones (SA); K-MART special $69.95 "High Fidelity" Discount Stereo (DS); GE $4.95 Pocket Radio in overload (PR).

Even when high-level hearing is not completely normal, informal experiments conducted by Gudmundsen and Killion, using a high-fidelity loudspeaker playing a piano trio recording, indicated that 12 of 14 hearing aid wearers could hear the difference between a 16 kHz bandwidth and the 4 to 5 kHz bandwidth typical of many aids. For these tests, the participants removed their hearing aids and the level was set for 85 dB SPL, which is typical of nightclubs. Of the 12 participants who could tell the difference, 2 preferred the narrower bandwidth. The 10 who preferred the wideband sound placed a dollar value of $20 to $10,000 on the wider bandwidth, with a median value of $100.

Response Smoothness

The question of how irregular the frequency response may be before it has a noticeable effect on fidelity has not been as well studied as the effect of restricting the frequency range.
Buckelin (1962) studied the effect of, among other things, 10 and 20 dB peaks and dips at 3.2 kHz in an otherwise flat-response transmission system. The peaks and dips used appeared fairly sharp, with an apparent 3 dB bandwidth of roughly 10% of the center frequency. He found that 100% of his observers could detect both the 10 and 20 dB peak, but less than half could detect the 20 dB dip; only 10% could recognize the 10 dB dip. Flanagan (1957), in a study of the difference limen for formant amplitude, found that a change of 3 dB in the amplitude of the second formant can be detected approximately 50% of the time.

Toole (1981, 1984, 1986) published the findings from 10 years of investigation into high-fidelity loudspeakers (and high fidelity in general). He observed (personal communication) that the loudspeakers that receive the highest fidelity ratings exhibit frequency responses lying within a ±1.5 dB tolerance from 100 Hz to 1 kHz, widening to about ±3 dB between 1 and 2 kHz, and then +1.5/-3 dB from 2 kHz to 10 kHz.

From all these results, it seems reasonable to infer that a response irregularity in a high-fidelity system of approximately 3 dB can be detected under appropriate conditions when the source material is speech but is not likely to be objectionable. Even much larger response irregularities have been found to be minimally objectionable (Dillon & Macrae, 1984) or even preferable (Cox & Gilmore, 1986) under some experimental conditions.

**Accuracy Scores**

Is there a way to reliably estimate directly from the frequency response curve what the fidelity rating would be if an extensive (and expensive) listening test was performed? With some limitations, the answer appears to be yes. A procedure based on loudness calculations was adopted some time ago by Consumers Union (CU) for rating the frequency response accuracy of high-fidelity loudspeakers (Consumer Reports, 1977). With this procedure, the loudspeakers are driven with a wideband “pink” noise (equal energy in each one-third-octave band) at an overall level adjusted to produce a calculated loudness (Stevens, 1972) of 88 sones, equivalent to a level of 90 PLdB “Perceived Loudness” in the 110 to 14,000 Hz frequency range. In each of the 21 one-third-octave bands, the loudness in sones corresponding to the total sound power
output of the loudspeaker in that band is calculated and compared to the sones calculated for a theoretically perfect loudspeaker producing exactly 74 dB SPL in that band. (A 74 dB SPL in each of those 21 bands produces a calculated loudness of 88 sones.) A percentage accuracy is then calculated from the deviations from perfection, averaged over the 21 bands. For purposes of the accuracy calculations, the individual band deviations from perfection are obtained by subtracting the ratio of measured to ideal sones from 1.0 (perfection), and then subtracting the absolute value of that difference from 1.0. Thus, a measured power equivalent to either .5 or 1.5 times the ideal loudness would produce a 50% accuracy score for that band. For example, a loudspeaker with a perfectly flat “power response” everywhere except for a 7 dB drop-off (half loudness) in the 125-Hz band would have a calculated accuracy score of \((50 + 20 \times 100)/21 = 98\%\). For our purposes, the accuracy score described above will be called the “21-band accuracy score” to distinguish it from a “25-band accuracy score” discussed below.

The 21-band accuracy scores obtained at that time by Consumers Union on 16 models of low-priced ($100 to $200 per pair) high-fidelity loudspeakers ranged between 63% and 93%, with a median value of 80%. Listening tests were said to have borne out the utility of the accuracy scores, although CU stated that “experience has taught us that a group of listeners won’t readily agree on which of two speakers is more accurate when the speakers’ scores differ by eight points or less” (Consumers Reports, 1977).

The general utility of the accuracy score as an indication of the fidelity rating to be expected from a sound system that has no defects other than frequency response inaccuracies (i.e., which doesn’t buzz, crackle, or otherwise audibly distort) was confirmed in listening test experiments (Killion, 1979a). Figure 5–7

![Figure 5-7](image-url). Frequency responses of sound systems used in complete listening tests and compared to calculated accuracy scores in Figure 5–6.
shows a comparison between fidelity rating and calculated 25-band accuracy score (the four one-third-octave bands below 125 Hz were added to the calculations) for a variety of loudspeakers, earphones, and hearing aids that were all rated on the same three selections of program material: orchestra, piano trio, and live voice.

A more detailed look at some examples is useful. The frequency response of six of the tested sound systems is shown in Figure 5-5. The insertion gain frequency response of the ITE aids shows a one-half-octave dip of about 4 dB at 2.7 kHz (due to insufficient compensation for the loss of external-ear resonance “hearing aid”), and a broad peak of about 3 dB at 5 kHz. This resulted in a calculated 21-band accuracy score of 91% and an average fidelity rating (for those three program selections) of 82%. The experimental CIC aids had a dip in their insertion gain response of only 3 dB at 2.7 kHz and a smooth boost of 2 dB at about 1.5 kHz. This resulted in a calculated accuracy score of 92% and an average fidelity rating (same program materials) of 86%. The studio monitor loudspeakers had a dip of 8 dB in their listening-room response at 2 kHz (the crossover frequency between the woofer and horn tweeter), with a calculated accuracy score of 83% and a fidelity rating of 80%. Finally, the BTE aids had an extremely smooth frequency response, with an 8 kHz high-frequency cutoff as their only real defect. This resulted in an accuracy score of 82% and a fidelity rating of 79%.

**Individual Variation in Ears**

If a separate hearing aid were to be designed for each user, it would presumably be possible to take into account individual eccentricities in external-ear (“ear canal”) resonances and eardrum impedances. To be economically practical, however, a high-fidelity hearing aid design must be based on average data. Under those circumstances, individual variations in outer-ear resonance and eardrum impedance may cause the (insertion) gain and the (insertion gain) frequency response of a hearing aid to deviate substantially from the design average in a given individual.

An estimate of the individual differences in outer-ear resonance was provided in the data of Flett, Ross, and Wiener (1945). In that report, individual sound-field-to-eardrum pressure curves for 12 male and 2 female subjects were given, which can be compared to the overall average curves for the same subjects shown by Wiener and Ross (1946). The standard deviation of the individual curves (from the average curve) ranged from 1 to 2 dB below 1800 Hz up to 4 to 7 dB in the 5 to 8 kHz region with peaks at 2.1 and 3.3 kHz. The peak deviations occurred mostly because individual external-ear resonance frequencies were lower or higher than average. No individual eardrum-pressure-level curve deviated more than 7.5 dB from the average curve below 5 kHz, but the majority deviated by at least 5 dB at some frequency below 5 kHz.

As part of a study leading to a validation of the modified Zwislocki coupler, Sachs and Burkhard (1972) reported the probe-tube microphone measurement of the sound pressures developed in 11 ears (6 male and 5 female subjects) by subminiature hearing-aid earphones. The standard deviation of the level of pressure that developed ranged from approximately 1 dB at 1 kHz to 5 dB in the 6 to 8 kHz range. Greater pressure levels were developed in female ears (by 3–5 dB at the higher frequencies).

While the variations in outer-ear resonance and eardrum impedance are only partially independent variables, it is clear that no hearing aid designed for the average ear can be expected, in the majority of individual cases, to produce an insertion gain that does not have at least one deviation of perhaps 7 dB at the worst frequency. Some perspective can be obtained from Killion’s listening-test experiment in which the “large red ears” were substituted for the “standard ears” on the KEMAR manikin. Using the standard ears as the perfect reference, the large ears obtained an average fidelity rating of only 81%. (Perspective is also provided by noting that relocating a “perfect” pair of loudspeakers in a listening room may change the one-third-octave response at the listening position sufficiently to reduce a 21-band accuracy rating from 100% to 85 to 90% area.)
The subjective importance of such response inaccuracies to a long-term hearing-aid wearer has not been studied, although it is known that even larger deviations in unaided frequency response can occur due to the accumulation of ear wax in the canal in which the onset is so gradual that changes often go unnoticed until the canal becomes almost completely blocked. In most cases, it thus seems reasonable to assume that satisfactory adaptation to a slightly inaccurate insertion gain frequency response will make it unnecessary to provide modification for individual eccentricities.

There are now data on the question of individual differences. Palmer (1991) studied the importance of having the hearing aid match the individual ear's response using subjects who had normal ear-canal resonances (REUR, real-ear unaided response) and normal ear-drum impedance (RECD, real-ear-coupler difference); normal REUR and abnormal RECD; and abnormal REUR and normal RECD. Each subject judged quality and intelligibility while listening to a frequency response that matched the ear-canal resonance (REUR) only, matched the ear-drum impedance (RECD) only, matched both, or matched KEMAR's REUR and RECD. She found that people who were different from average, but not too different, preferred the tailored response to their own ear. People who had substantially deviant REUR or RECD preferred the average (KEMAR-based) response. The individuals with dramatically collapsed canals, for example, preferred the average response.

All things considered, a reasonable goal for the smoothness of the frequency response of a high-fidelity hearing aid would thus appear to be an accuracy score of 80% as measured on the KEMAR manikin. Such a hearing aid would have a calculated accuracy score equal to or better than half the $100-$200-per-pair loudspeakers tested in 1977 by Consumers Union. A more stringent goal would be a 21-band accuracy score equal to the median (89%) of the expensive ($600-$1,000 per pair) "state of the art" loudspeakers tested in 1978 by Consumers Reports. In either case, the response accuracy goal would presumably apply only to the idealized "average" hearing-aid wearer.

**Amplitude Response Requirements**

Although the distinctions are somewhat arbitrary, we generally think of the dynamic range of a sound system (between its maximum undistorted output levels and its noise levels), its nonlinear distortion (harmonic and inter-modulation), and any automatic gain control (AGC) functions as taking place in the amplitude or intensity domain. Some amplitude-domain performance requirements imposed on a high-fidelity hearing aid are discussed below.

**Peak Output Levels**

There is no easy answer to the question of what the maximum undistorted output of a wearable unity-gain sound reproduction system should be. If the peak outputs of rock music played at a discotheque are to be reproduced, a 130 dB SPL capability may be required (R.W. Peters, personal communication). A similar capability would be required to reproduce the peak levels produced by some aircraft and industrial noise sources. As summarized by Miller (1974), however, such levels are hazardous to the hearing mechanism.

**Speech and Everyday Sounds**

In everyday conversational settings, the highest levels at the hearing aid microphone will normally be generated by the user's own speech. The data of Dunn and Farnsworth (1939) indicate that the overall speech levels measured at the talker's ear were about equal to the levels 30 cm in front of his or her lips. Thus, the Dunn and White (1940) data on instantaneous peak levels in speech measured at 30 cm may be used directly. For normal conversational speech, instantaneous speech peaks of 90 to 95 dB SPL occur with some regularity (in 1 to 5% of one-eighth-second intervals). (Due to the head-shadow effect, the high frequencies will be attenuated somewhat, but since the majority of the peak energy lies below 1000 Hz, this will have little effect on the overall peak levels.) Unless the conversation becomes agitated, therefore, the instanta-
aneous peak levels at the microphone will be 90 to 95 dB SPL, a range that has often been used as the goal for the level of minimum undistorted input in hearing-aid design. If the conversation becomes agitated, a shout or forceful expletive can easily produce 100 to 105 dB instantaneous peak SPL at the microphone of the speaker’s own hearing aid. The first author, a far-from-professional singer, can still sing a high “A” that has produced a 112 dB instantaneous peak SPL at the microphone of a head-worn hearing aid that was wired to an oscilloscope.

Other commonly encountered sounds, such as the clack of a typewriter key or a finger snap at arm’s length, can also produce a 100 to 110 dB instantaneous peak SPL. A spoon dropped onto a plate can produce a 110 to 115 dB instantaneous peak SPL.

**Live Music**

In live performances of classical music, Marsh (1975) reported that in a good main-floor seat in Chicago’s Orchestra Hall, “a fully scored orchestral passage in a Mendelssohn symphony reaches approximately 95 dB on a decibel meter.” Marsh reported that approximately the same levels are reached during a similar passage in the front benches at outdoor amphitheaters such as Grant Park, or at the edge of the Ravinia stage. The typical instantaneous peak factor for an orchestral passage of this sort is 5 to 10 dB, indicating instantaneous peaks of 100 to 105 dB SPL at these three Chicago-area locations. Similarly, for a typical listening position in a large music hall, Olson (1967) reported the instantaneous peak sound pressure level as 100 dB SPL.

During the last 5 years, one of the authors has regularly carried a sound level meter (SLM) to concerts. The Chicago Symphony Orchestra has produced a peak SLM reading of 104 dB on the A scale during a Mahler symphony, measured in a seventh-row, center balcony seat. (At the other extreme, the quietest piano passage in Strauss’s Burlesque for Piano and Orchestra in D Minor was 35 dB. The same 35 dB level was measured on a solo viola in Elgin and Payne Symphony No. 3. Rapid SLM reading changes of 40 to 50 dB are not uncommon.

Surprisingly high decibel levels are even encountered at audiological parties where bands are playing. Levels of 95 dB are common. At the 1999 opening night party of the American Academy of Audiology Convention, held on the beach, the levels in the dance area ranged from 108 to 112 dBA. At a Chicago blues bar, levels of 110 to 114 dBA are not uncommon.

All things considered, an undistorted input capability of 115 dB instantaneous peak SPL—referred to the sound field—appears to be a reasonable minimum requirement for a high-fidelity hearing aid intended for use at live concerts.

What is also needed for our present purposes, however, is information on the peak output levels required of the hearing aid earphone. This requires consideration not only of the frequency distribution of peak levels in music but also consideration of the increased eardrum pressure levels normally produced by external-ear resonances.

The frequency distribution of the maximum instantaneous peak levels for a 75-piece orchestra was given by Fletcher (1931) and is shown in Figure 5-8. Several more recent measurements have indicated that the high-frequency peak levels do not fall off as fast as Fletcher’s curve indicated (and indeed Fletcher himself intentionally excluded the cymbals; an instrument whose inclusion would have brought the 10 kHz level back up near the level at 250 Hz). It is sometimes argued, therefore, that a no-compromise system would allow no drop in the high-frequency peak capability.

Because occasional high-frequency overload is generally found to be less annoying than a constant hiss, however, a 75 μs, high-frequency preemphasis (a 6 dB per octave upward slope above a 2.1 kHz corner frequency) remains the standard in FM broadcasting. Similarly, current AES standards on prerecorded tape and phonograph records call for a preemphasis ranging from 75 to 150 μs. The effect of the preemphasis, which is corrected in playback, is to improve the signal-to-noise ratio of recorded material at the expense of a reduced high-frequency overload capability.
Figure 5-8. Maximum peaks produced in half-octave bands by large orchestra, referred to audience area (solid line). Instantaneous peak (of sine wave) eardrum pressure capabilities required of hearing aid to reproduce large orchestra (dotted line). (Note: Orchestra peaks correspond to instantaneous wideband peaks of 105 dB SPL or peak readings of 95 dB SPL on sound level meter set to “C, Fast.”)

Not surprisingly, the frequency dependence of the instantaneous peak-pressure requirements is not much different when it is based on the Fletcher curve or on the assumption of a 75 microsecond preemphasis. This comparison is shown in Figure 5-8.

In terms of system requirements, approximately 10 dB of “headroom” is required in any half-octave band over the instantaneous peak levels shown in Figure 5-8. This comes about because the presence of energy in other frequency bands can, when added to that present in a given band, produce instantaneous peaks greater (by approximately 10 dB in the case of half-octave bands) than the peak that would be produced by the in-band energy acting alone. The measured instantaneous peak overload capability of a system at a given frequency (measured as the instantaneous peak of a sinewave signal just before clipping) must therefore exceed the half-octave-band instantaneous peak measurements of music by approximately 10 dB.

The output requirements for a hearing aid must also take into account the increase in eardrum pressure produced by outer-ear resonances. On a random-incidence basis (as in a concert hall where sounds are arriving from all directions), this increase amounts to approximately 15 dB at 2700 Hz (Shaw, 1976). When combined with the considerations of the previous paragraph, this means that a hearing aid that is to provide unity gain for high-level sounds must be capable of producing an instantaneous peak eardrum pressure level at 2700 Hz that is 25 dB greater than the peak level shown in Figure 5-8, or a maximum of 117 dB SPL at 2700 Hz. The dotted curve in Figure 5-8 shows the estimated instantaneous peak (sine-wave) output requirements for a high-fidelity hearing aid operating at unity insertion gain for high-level signals. Fortunately, even a 127 dB eardrum pressure peak corresponds to only 112 dB SPL in a 2cc coupler at 3 kHz because of the normal 15 dB RECD at 3 kHz. This output is easily reached with Class B or Class D receivers.

**Input Noise Level**

The input noise level of a high-fidelity hearing aid should be less than that of the ambient noise levels likely to be encountered by a user. The A-weighted noise level during a quiet listening period in a theater or auditorium may drop to 32 dB (Fletcher, 1942; Olson, 1967)
and sometimes lower (author’s observation). Residential noise levels are generally higher. Seacord (1940) measured noise levels in a large number of residential rooms and found an average level of 43 dB A-weighted; 90% of the levels fell between 33 and 52 dB(A). Seacord’s data (1940) have found common acceptance, although it is generally believed that the greater use of forced-air heating and air-conditioning systems has acted to increase average levels since his data were obtained.

Macrae and Dillon (1986) determined the maximum hearing aid equivalent input noise levels acceptable to their subjects. A hearing aid in which input noise just met their one-third-octave limits at each frequency would have an overall A-weighted equivalent input noise of 38 dB SPL.

The input noise level in a modern hearing aid is determined almost entirely by the microphone noise level. Currently available subminiature microphones have typical A-weighted noise levels equivalent to a 26 dB SPL ambient noise level; a few are 2 to 3 dB quieter. Even in a quiet auditorium, therefore, the microphone noise level would add less than 1 dB to the apparent A-weighted ambient level. The aided threshold determined by typical microphone noise levels is within a few dB of normal threshold and can be better than normal with special design (Killion, 1976a). The internal noise of the hearing aid should thus present no impediment to a high-fidelity design.

**Distortion**

We will use the term distortion to mean nonlinear distortion, in the restricted sense of a distortion that results in the generation of new frequencies appearing in the output but not present in the input stimulus. An example of nonlinear distortion would be the overload of an amplifier so that clipping occurs. Common measures of nonlinear distortion are total harmonic distortion and intermodulation distortion, both of which measure the relative strength of the new frequencies created by the nonlinearity when sine-wave signals of one or more frequencies are presented to the input.

In a study of the amounts of distortion tolerable in a high-fidelity system, Olson (1957) found that total harmonic distortion levels of approximately 1% were just detectable and that total harmonic distortion levels below 3% were not considered objectionable in a system with an 8 kHz upper cutoff frequency. Two systems were tested, one using a single-ended triode amplifier and another using a single-ended triode amplifier and one using a single-ended pentode amplifier. In both cases, the distortion increased with increasing output, and the distortion percentages represented the total harmonic distortion of sine waves with an output level equal to the peak levels of the music used as program material.

Nonlinearities in the system that occur near the upper cutoff frequency are difficult to examine using harmonic distortion measures because the harmonic distortion products (at 2f1, 3f1, etc.) may lie above the cutoff frequency. High-frequency distortion can be easily detected, however, by applying two high-frequency sine waves to a system and measuring the system output at the difference frequency. This has been called the “CCIF method” for measuring intermodulation distortion and is described more completely in the references below.

Tests made by the British postal service and reported by Moir (1958) indicated that CCIF intermodulation distortion on speech and music “is not detectable when the quadratic or cubic difference tones are . . . below one percent.” Termin and Pettit (1952) reported that CCIF intermodulation distortion becomes objectionable at a value of 3 to 4% when the difference frequency lies in the 400 to 5000 Hz range.

Tests employing “golden ears” have produced similar results, as summarized by Milner (1977) and Davis (1978). Intermodulation distortion levels below 2% are generally inaudible on musical material, and even gross distortion levels (6 to 12%) are sometimes inaudible. In general, the barely audible distortion levels for musical material are at least
10 times greater than the barely audible pure-tone distortion levels.

One complication that arises in attempting to apply these results to hearing aids is the necessity of translating the results into eardrum-pressure or coupler-pressure levels. Although eardrum pressure and sound field pressure are roughly equal below 1000 Hz, the combined effect of head diffraction and external-ear resonance results in a 10 to 20 dB boost in eardrum pressure levels in the 2000 to 5000 Hz region (Wiener & Ross, 1946). One consequence of the treble boost provided by the head and outer ear is that the ear is able to detect lower levels of harmonic distortion when the harmonics fall in the 2000 to 5000 Hz region. In addition, small head movements may place one ear in a "null" position for the fundamental tone in the listening room, which can result in incredibly low detection levels for pure tone distortion in real rooms.

A further complication is the level-dependence of the ear's sensitivity to distortion. At low sound pressure levels, the level of any harmonic or intermodulation distortion products may lie below the normal threshold of hearing. At high levels, the increased upward spread of masking (Wegel & Lane, 1924) and the distortion of the ear itself may mask externally generated distortion products. Thus no single-number distortion specification will apply at all listening levels.

The tests reported by Olson (1957) were carried out in a small listening room at a level of "about 70 dB." Assuming that this level was close to the 75 dB levels reported by Olson for other similar listening tests, it corresponds to instantaneous peak levels of approximately 85 dB SPL. This is only slightly greater than the 70 to 80 dB SPL that a study of the masking literature indicates is the region in which the ear should be most sensitive to distortion. Thus, the values for detectable and tolerable distortion obtained by Olson and others reasonably can be applied as a requirement for a high-fidelity hearing aid only for output levels in the range between about 50 and 90 dB SPL (measured at the eardrum or in an ear simulator). Below 50 dB and above 90 dB, a less stringent requirement is clearly in order, and a relaxation to perhaps 10% at 30 dB and 110 dB eardrum SPL appears reasonable.

The last problem in attempting to arrive at reasonable distortion specifications is probably the most important one. Unless the precise distortion mechanism (peak clipping, center clipping, curved transfer characteristic, etc.) is understood, no single distortion measurement can provide reliable information as to how clean the sound will be to a human listener. Thus, 1% of "soft peak clipping" (as employed in the determinations discussed above) may be inaudible, but 1% of center clipping distortion may be intolerable.

Similar difficulties are encountered in attempting to study the effect of distortion on speech discrimination. Thus, Peters and Burkhard (1968) found that the 40% total harmonic distortion produced by one system had negligible effect on speech discrimination, whereas another system in which the total harmonic distortion measured only 20% resulted in a loss of 40 percentage units in speech discrimination score.

In light of the available information, a reasonable initial goal for a high-fidelity hearing aid with an 8000 Hz bandwidth would appear to be a maximum total harmonic distortion or CCIF intermodulation distortion of 2% for output levels between 50 and 90 dB eardrum SPL (assuming the distortion mechanism is a simple one), with a smooth relaxation to 10% between 30 and 110 dB SPL. Figure 5–9 illustrates such a requirement. Listening tests should be employed as a final check.

**Automatic Gain Control Characteristics**

Recall that a unity-gain sound reproduction system is desired for high-level sounds, coupled with sufficient gain for (desirable) low-level sounds to make them audible to the hearing-aid user.

The low-level gain required of a high-fidelity hearing aid can be readily estimated. Although a gain numerically equal to the user's hearing loss would be required to restore his
oration of clinical data and the literature on recruitment indicates that recruitment is typically complete (loudness sensation is essentially normal) for sounds corresponding to a hearing level of 80 dB or greater. More recently, Barford (1978) has shown that in some cases a nearly linear relationship exists between the degree of hearing loss and the hearing level at which recruitment is complete. For Barford's subjects, all of whom had steeply sloping high-frequency losses with nearly normal low-frequency hearing, all hearing losses below 50 dB HL were accompanied by complete recruitment above 75 dB HL. By way of illustration, Figures 5-10A and 5-10B show two hypothetical hearing losses with their corresponding areas of presumably normal hearing, based on Barford's loudness data.

As a practical example, assume a user has a 45 dB HL cochlear impairment with complete recruitment for sounds above 80 dB HL. By our assumptions, he requires a maximum gain of 30 dB (45 minus 15) for sounds at 15 to 20 dB HL, and unity gain for sounds at 80 dB HL and above. (For speech sounds, 80 dB HL corresponds to a 95 dB SPL in a 0°-incidence sound field, or 100 dB SPL under headphones. For pure tones, 80 dB HL corresponds to approximately 80 to 90 dB SPL in the sound field in the frequency range important for speech perception.)

**Optimum Amplitude Input-Output Characteristics**

To avoid constant adjustments of the volume control, an Automatic Gain Control (AGC) system is required. To introduce the minimum degradation in perceived sound quality, the operation of the AGC system must be unobtrusive. Compression amplification has found wide acceptance in the broadcast and recording industry for such purposes (Blesser & Ives, 1972). As originally defined at Bell Telephone Laboratories (Mathes & Wright, 1934), compression amplification meant what is now sometimes called logarithmic compression (to distinguish it from some of the misuses of the term), that is, a constant ratio between the logarithms of the input and output signal.
amplitudes. When input and output levels are expressed in dB, for example, a compression ratio of 2:1 corresponds to a 5 dB increase in output level for each 10 dB increase in input level.

The idea of applying compression amplification to hearing aids is not new, dating back at least to Steinberg and Gardner (1937). Wide dynamic range compression amplification was apparently first reduced to commercial practice in wearable hearing-aid designs by Goldberg (1960, 1966).

Figure 5-11 illustrates input-output characteristic (Killion, 1979b) for a hearing aid intended to meet the requirements of the example discussed above. There are four stages of amplification illustrated in Figure 5-11: a low-level, constant-gain stage; a mid-level, constant-compression ratio (2:1) stage; a high-level, unity-gain stage; and a very high-level, compression-limiting stage.

The final compression-limiting stage requires further comment. Fast-acting, low-distortion compression limiting applied to the microphone output for sounds above roughly 100 dB hearing levels (equal to 110–115 dB SPL) was originally planned to prevent audible distortion or bias shifts in the amplifier associated with overload. Output limiting was not expected to be needed to prevent discomfort if the hearing aid was set for unity gain for loud sounds. In practice, the K-AMP circuit has never had its output limiting circuit activated, and complaints of excessive loudness have been almost nonexistent.

Note that the input-output characteristic illustrated in Figure 5-11 is quite different from any of the three other types that were commonly employed in hearing aids at the time the first edition of this chapter was written and shown for comparison in Figure 5-12: Output limiting (peak clipping or low-distortion compression limiting), AVC (automatic volume control), or wide dynamic range compression. Properly adjusted, output limiting can prevent amplified sounds from ever becoming too loud, but linear amplification combined with output limiting often makes a large proportion of sounds almost too loud. An automatic volume control can insure that all sounds are amplified to the most comfortable listening level, but the resulting restriction of the dynamic range lends a sameness to all sounds that is highly unnatural. Wide dynamic range compression comes closest to the “ideal” characteristic of Figure 5-11 but continues to perform signal processing even for loud sounds, when the hearing aid should be simply “getting out of the way” of the wearer’s presumed normal hearing.

**Figure 5-10.** Threshold audiograms for two hypothetical subjects, with areas of presumed normal hearing based on Barford’s (1978) data.
Level Dependent Frequency Response

Few threshold audiograms are flat. A greater loss at the higher frequencies is commonly found in those with hearing loss. For low-level sounds, a high-frequency emphasis in the hearing aid's frequency response is required to make the entire range of quiet speech sounds audible without overamplifying the low frequency sounds. For high-levels sounds, on the other hand, unity gain (and thus a flat frequency response) is required.

This need for a level-dependent frequency response can be met in several ways. The technique most often described in the literature is the use of multiple-channel compression amplification, with the compression ratio in each channel chosen to compensate for the degree of hearing loss in that frequency band (Vilchur, 1973).

A simpler method that is adequate in the case of mild-to-moderate hearing impairments is to use a single-channel amplifier with a suitable capacitor in series with the gain-determining element. As an illustration of what can be accomplished with relative ease in a practical hearing aid amplifier, Figure 5-13 shows the level-dependent frequency response characteristic of the K-AMP (Killion, 1979b). A greater high-frequency emphasis at
Figure 5–12. Input-output characteristics for three conventional approaches to dynamic range reduction in hearing aids.

Figure 5–13. Relative amplifier frequency response versus input level in “speech mode” (compression ratio set for 2:1) of first author’s experimental amplifier.
low levels would be required to meet the needs of the hypothetical individual whose audiogram was shown in Figure 5–10, but that also should be practical.

The need for a level-dependent frequency response has been discussed by Barford (1976), Goldberg (1960, 1972), Skinner (1976, 1980), and Villchur (1973, 1978). Whatever the optimum characteristic, properly executed compression amplification should, in the ideal, not significantly degrade the fidelity as judged by someone without a hearing impairment, and this can be achieved. BTE aids with the K-amp (Killion, 1979b), intentionally operated in their worst-case operating range (from the standpoint of measured distortion) but in their flat frequency response mode, received average fidelity ratings across subject groups that were insignificantly different (2.5% higher) than the ratings given the BTE hearing aids themselves (Killion, 1979a), verifying not only that high-fidelity compression amplification circuitry operating on a 1.5 V hearing aid battery was practical, but that the tested compression characteristic was acceptable from the standpoint of fidelity.

**Time Response Requirements**

In this section, the transient response of a hearing aid is considered.

**Transient Wave Form Response**

There are two common interpretations of the term “poor transient response.” One is the difference between the output and input wave forms viewed on an oscilloscope screen (transient wave form response) when a transient is applied to a sound reproduction system. In general, this difference is an inevitable consequence of any frequency response shaping in the system under test. By the above definition, for example, the ear itself has poor transient response because of the resonances in the external ear.

To a reasonable first approximation, a hearing aid system can be represented as a minimum-phase network. Under those circumstances, the transient response can be predicted directly from the frequency response. The frequency response tailoring of a particular hearing aid may or may not be useful, but its effect on “transient response” is inevitable. Indeed, the inverse procedure—obtaining the frequency response of a hearing aid by analyzing its transient wave form response to short clicks—is now routinely used in one commercial probe microphone system for obtaining the real-ear response of hearing aids.

Two papers should be read by anyone interested in poor transient response. A comprehensive set of frequency response curves and their corresponding transient response wave form resultants was given in the classic paper by Mott (1944). The surprisingly large changes in wave form due to phase shifts that are nonetheless completely inaudible (or audible only under what F. V. Hunt used to call “carefully contrived listening tests”) have often been discussed. Bauer (1974) described some of the more rigorous experiments along those lines, which indicated that introducing even 90° per octave phase shifts does not produce audible effects. (Many contaminated experiments exist. Loudspeaker manufacturers selling “time aligned” loudspeakers have been some of the worst offenders, demonstrating readily audible changes that are claimed to be caused by phase changes alone when in fact large, readily audible changes in the amplitude frequency response accompanied the manipulations.)

**Transient Amplifier Overload**

The other type of “poor transient response” generally involves an amplifier that exhibits a slow recovery from overload. As mentioned above, instantaneous peak sound pressure levels of 110 to 115 dB at the hearing aid input are not uncommon. Such peaks can easily cause sufficient amplifier overload to upset the bias levels on the internal coupling capacitors, causing a condition of high amplifier distortion lasting much longer than the transient itself (Ingelström, Johanson, Pettersson,
and Sjogren, 1971). This “blocking distortion” was more often a problem with older amplifier designs and is much less of a problem now that the majority of hearing aids use some form of fast-acting compression limiting or AGC system.

**AGC Time Constants**

The attack and release time constants of the AGC system used to obtain the desired input-output characteristics throughout the operating range are an important consideration in any sound processing system. As extensively discussed by Lippman (1978), the proper choice of time constants depends greatly on the goal set for the AGC system of the hearing aid. When the goal is to maximize speech discrimination, for example, the results of Ahren and colleagues (1977) and Schweitzer and Causey (1977) indicate the attack time should be as short as possible and the release time between 30 and 100 ms, but conclusive findings on this subject have not yet been made. When the goal is to maximize sound quality, on the other hand, the situation is much less clear. Even under ideal conditions, such as those found in professional recording studios, the optimum choice of attack and release time for minimum perceived distortion is highly dependent on the program material. Thus, any choice will be “wrong” at least part of the time. These issues were discussed at some length by Blesser and Ives (1972), who reported that values of 10 ms and 150 ms for attack and release times, respectively, have found common acceptance in equipment designed for the broadcast industry. In the absence of reliable research findings on the optimum values for hearing aids (with sound quality as the goal), these values would presumably represent a reasonable first choice for the AGC system of a high-fidelity hearing aid.

A way out of this dilemma has been recently implemented following an earlier RCA development (Singer, 1950) that has been used for some time in the broadcast industry (Smriga, 1986). This “Adaptive Compression” or “Variable Recovery Time” technique automatically varies the release time dependent on the duration of the high-level signal that engaged the compression action.

In the case of compression-limiting circuits that employ high compression ratios to prevent high-level inputs from producing outputs exceeding the loudness discomfort of the wearer, the use of variable recovery time can increase intelligibility in noise. Fikret-Pasa (1993) compared two 8:1 low-distortion compressors (using basic K-AMP circuitry) with eight subjects having moderate to severe hearing loss. Two of her subjects, Harry Teder and Larry Revit, are well-known engineers in the industry. Listening to a female speak in four-talker babble (the forerunner of the SIN test), each subject obtained the worst SNR with peak clipping and the best with variable recovery time compression. The industry-typical 50 ms recovery time compression was in between. The average SNRs that required for 50% correct in noise were:

- Peak clipping: 20 dB SNR
- 50 ms compression: 14 dB SNR
- Variable recovery time compression: 9 dB SNR

The presumed explanation (Teder, 1991) is that a 50 ms recovery time brings the background noise quickly up to the level of the desired speech, making it more difficult for the listener with hearing impairment to use envelope cues to help separate the two.

In the case of wide dynamic range compression circuits using relatively low compression ratios (and increasing gain for quiet sounds rather than limiting gain for loud sounds), the known benefit of variable recovery time (VRT) is better subjective fidelity. VRT increases the steady-state recovery time to 500 ms or so, substantially reducing the “pumping” sound of typical 50 ms recovery, while reducing the recovery time after short transients (hand clap, knife on plate, etc.) to only 20 ms. Intelligibility may or may not be improved (it has not been studied to the author’s knowledge), but sound quality certainly is.
IMPLEMENTATION OF THE GUIDELINES IN PRACTICAL HEARING AIDS

Success of Wide Dynamic Range Hearing Aids

In the first article on the new K-AMP amplifier, the final sentence read “The ultimate test of this amplifier design will be the degree of its acceptance by hearing-impaired wearers.” The acceptance has been good: There have now been some 3 million of these devices sold, and several times that many Class D receivers. In 1990, wide dynamic range compression (WDRC) processing was found mostly in ReSound (two-channel) and K-AMP (one-channel) circuits. It is now found in nearly all digital hearing aids and in many modern analog circuits, such as the DynamEQ2.

CIC Hearing Aids

Two developments made high-fidelity hearing aids more practical: soft-tip, deeply sealed earmolds and completely-in-the-canal (CIC) fittings.

The practicality of deeply sealed eartips has been demonstrated in the tens of thousands of Musicians Earplugs® sold around the world, which typically use a deeply sealed eartip to reduce the occlusion effect to, in the best cases, less than zero. Zwislocki (1953) first observed the elimination of the occlusion effect with a deep seal; Killion, Wilber and Gudmundsen (1988) demonstrated its effectiveness and practicality. By sealing in or near the bony portion of the ear canal, the canal-wall vibration caused by speaking, chewing, or blowing on a horn can be isolated from the eardrum. The evidence from Musicians Earplugs indicates that deeply sealed BTE earmolds are entirely practical. (They also reduce the likelihood of feedback by improving the seal and reducing the sound-producing vibration of the earmold.) Soft-tip or soft-shell canal aids were first introduced by Voroba and often eliminated the occlusion effect as discussed in the Killion et al. paper above. Since that time, a steady search for practical, durable materials and construction methods has brought us close to the time when soft-shell or soft-tip ITE, canal, and CIC aids may become routine.

The CIC aid is a special case, because many wearers report little occlusion effect. That was the first author's experience with a pair made for him years ago by Randolph Giller. A deeply sealed hard-shell ITE aid has the disadvantage that any bump to the aid or wiggling of the ear can cause intense pain in the bony portion of the ear canal, where the skin is extremely thin and the bone extremely hard. The CIC construction has the advantage that nothing sticks out into the ear canal to be bumped or wiggled so that a snug fit is practical even with hard materials. It may also be that part of the occlusion effect with shallow-tip ITE hearing aids comes from concha vibration transmitted to the shell, causing the shell to pump in the ear canal. Whatever the cause, it is usual to obtain, initially or after a remake, a CIC aid with low occlusion effect and, perhaps for similar reasons, surprisingly low problems with feedback.

Until recently, one problem in smoothing out the real-ear frequency response of hearing aids was the fact that available hearing aid receivers (earphones) had their primary peak at about 3.5 to 4 kHz. Knowles has recently introduced two Class D receivers, the EP-series and the even smaller ES-series, whose primary peaks are near the 2.8 kHz resonance of the external ear.

Musicians as Judges

One of the sternest tests of a hearing aid is whether or not a musician can wear it while performing. For one thing, the circuit of a hearing aid must be capable of handling the peak
SPLs created by the musicians themselves. In the case of virtuoso violinists, this amounts to 115 dB at the left ear (closest to the violin body). As an amateur violinist, it has been gratifying to the first author that several violinists in the Chicago Symphony Orchestra now wear K-AMP hearing aids during rehearsal and performance. (The old stigma is finally gone; it is now well understood that a loss of sensitivity for quiet sounds does not have to interfere with a musician's ability to play; one of the best violinists in the symphony has been wearing a K-AMP hearing aid for years.)

In contrast, one author as choir director has tried a wide variety of popular all-digital hearing aids with disappointing results. Most overload (distort) so badly with the live choir and piano levels that it is impossible to listen for choral balance and intonation: The errors introduced by the hearing aids overwhelm those of the choir!

The problem of handling performance-level SPLs without distortion may explain the comment made by Chasin (personal communication, June 7, 1999): “Out of the last 200 hearing aid fittings on musicians, 180 of them were K-AMPS. Musicians seem to like the low distortion levels of the K-AMP in high-volume environments—perhaps it has to do with the high input limiting level of the K-AMP that distinguishes it from other hearing aids, including digital products.”

A Persistent Wax Problem

There is one remaining problem in CIC applications. The damping mesh that must be placed in the receiver outlet to provide a smooth frequency response tends to clog fairly quickly because it is, of necessity, exposed to ear wax. An electronic damping chip is being developed as this chapter is being completed. The use of electronic damping—inverse filtering to remove peaks in the frequency response—should permit wax-free operation of CIC hearing aids with full fidelity.

FIDELITY AS A RETURN TO "HOW WELL I USED TO HEAR IN NOISE"

Performance in Noise

Signal-to-Noise Ratio Loss

As discussed in the introduction of this chapter, "high-fidelity" hearing aids have enabled listeners with hearing impairment to hear speech in quiet at all input levels without adding distortion (Benter & Duve, 1997). The same study, however, showed that today's circuits still do not improve the ability to understand speech in high-level noise. Recently tests that quantify a listener's ability to hear in noise have become an important part of evaluating a patient for amplification (Killion, 1997a, 1997b). These measures (the Speech-in-Noise [SIN] test and the Hearing-in-Noise Test [HINT]) will determine the SNR required for a listener to understand 50% of words-in-sentences. Listeners with normal hearing typically need a 2 dB SNR on the SIN test to understand 50% of words-in-sentences. In other words, the listener with normal hearing requires the signal to be 2 dB louder than the noise to understand 50%. Listeners with hearing impairment usually require a greater SNR to understand 50%. The SNR required is poorly correlated with the audiogram, however, and therefore needs to be measured on all listeners with hearing impairment. Once the SNR required for 50% correct identification of words-in-sentences is determined, the patient's SNR loss is calculated by subtracting 2 dB (the SNR required by a listener with normal hearing). Therefore, a listener with a measured SNR of 8 dB has a 6 dB SNR loss.

Figure 5–14 shows a plot of hearing loss (pure-tone average) by the SNR loss. Each dot represents an individual listener with hearing impairment. Note that the regression line does not fit the data well, except showing that in general the more hearing loss, the greater the SNR loss. There is a wide range of SNRs
for individual listeners with the same average hearing loss, for example, look at the average loss of 60 dB HL. Data from four subjects are plotted on the graph. For these four subjects, the SNR required for 50% correct ranges from 4 dB to 16 dB.

Figure 5–15 shows another way to look at these results. Plotted in this figure is the SNR required for 50% correct for a subject with normal hearing and a subject with a 40 dB average hearing loss. For the listener with normal hearing, the SNR required for 50% correct is approximately 2 dB when listening at a comfortable suprathreshold level. Below 30 dB HL, the SNR required increases because many speech sounds will be missed at this low level. The curve for the individual with a 40 dB average hearing loss is shifted by the amount of the hearing loss, but in addition, this individual will require a greater SNR than normal even at a comfortable suprathreshold level. For this hypothetical listener, a 7 dB SNR is required to produce 50% correct. Thus, this patient with a 40 dB average loss has a 5 dB SNR loss. Figure 5–16 illustrates what happens with a 1960’s peak clipping hearing aid and a modern hearing aid. The curve for the 1960’s hearing aid shows that while the aid does provide somewhat better audibility (represented by the shift in the curve to the left), the SNR requirements substantially increase with this aid. In other words, this individual will have greater trouble understanding in noise with the hearing aid than without it due to distortion in the hearing aid. The high-fidelity aid of the 1990s does not increase the SNR requirement for 50% correct, but it does not improve the SNR requirement either. The patient with the 40 dB average loss still has a 5 dB SNR loss with this modern hearing aid. Figure 5–17 shows Bentler and Duve’s (1997) SNR results for each hearing aid used in their study. With those subjects, all of the modern hearing aids required an SNR of approximately 6 dB for 50% correct.

**Directional Microphones**

Given that hearing aid circuitry has not solved the hearing-in-noise deficit experienced by lis-
Figure 5-15. SNR required for 50% correct identification of words-in-sentences plotted by presentation level. Results for a normal-hearing listener and a listener with a 40 dB average hearing loss are shown.

Figure 5-16. SNR required for 50% correct identification of words-in-sentences plotted by presentation level. Results for a normal-hearing listener and a listener with a 40 dB average hearing loss are shown. The performance of two hearing aids (1960s peak clipper and a 1990s modern high-fidelity aid) are given for the listener with the 40 dB average hearing loss.

teners with hearing impairment, engineers have looked for solutions to improve the signal-to-noise ratio for these listeners. The oldest such solution was the speaking tube. Because the talker speaks directly into the mouthpiece and the level at the lips is typically 105 to 115 dB
SPL, the SNR with a speaking tube can easily reach 20 to 40 dB. (The losses in the tube bring the listening level down to comfortable levels.) FM systems can provide similar SNR improvements with the advantage of wireless operation.

Less dramatic but more convenient SNR improvements can result from using directional microphones in hearing aids. Directional microphones improve the SNR for sounds coming from the front of a listener by attenuating arriving sounds from the sides and the back. Directional microphones in hearing aids are not a new development, although today’s microphones provide substantially more benefit than the pre-1990s directional microphones (Killion et al., 1998).

A full chapter in this book is dedicated to directional microphones so this discussion on directional microphones will be limited to the benefits that these microphones can provide listeners.

Today’s directional microphones are designed in two ways: (1) electrically subtracting the outputs of two omnidirectional microphones and (2) using a single directional microphone with two ports and acoustically subtracting the signals at these two ports following a delay (called a single cartridge approach). Both methods will provide good directivity if designed correctly.

Killion and colleagues (1998) evaluated the performance of a single-cartridge directional microphone (D-MICTM) in real-world environments. In this study, recordings were made from hearing aids equipped to record from the outputs of both the omnidirectional and the directional microphones simultaneously. Recordings in a variety of environments were accomplished, and results indicated SNR ratio improvements in the directional condition ranging from 4 dB in reverberant environments such as restaurants and cocktail parties to 11 dB in outdoor (nonreverberant) environments. These improvements in SNR ratio translated to significant improvements in speech recognition scores in noise ranging from 30 to 60%.

 Appropriately designed directional microphones finally provide listeners help when listening in noisy environments. Thus, coupling a directional microphone with a high-
fidelity hearing aid amplifier will provide listeners with mild-to-moderate hearing loss substantial benefit when listening in both quiet and noisy environments.

**SUMMARY AND CONCLUSIONS**

This chapter has attempted to present the design characteristics required for high-fidelity amplification. An understanding of these characteristics is essential for making decisions about proper amplification for listeners with hearing impairment. Modern hearing aids have eliminated the major problems with older technologies. These problems included distortion, narrow bandwidths, and peaks in the response. High-fidelity hearing aids are available that provide the listener who has hearing impairment with excellent speech understanding in quiet environments for both soft and loud input levels.

Unfortunately, modern high-fidelity hearing aids have not solved the speech understanding problems experienced by listeners with hearing impairment in noisy environments. The problem of hearing in noise has shifted to attention on improving the SNR for the listener. This can be done in a variety of ways but most conveniently by using a directional microphone coupled with a high-fidelity hearing aid.

Future developments in amplification will likely include microphone options that provide greater SNR improvements for listeners with hearing impairments. These improvements will allow hearing aids to provide excellent speech understanding in both quiet and noisy environments. A goal that has taken many decades to achieve.

**REVIEW QUESTIONS**

**Principles of High-Fidelity Hearing Aid Amplification**

1. In the Bentler and Duve study (1997), hearing aid technologies over the last 50 years were evaluated. Which of these technologies performed the best?

2. What are the deficits mentioned in the chapter that were characteristic of older hearing aid technologies that have now been overcome by modern hearing instruments (circle all that apply)?
   a. Narrow bandwidth
   b. Background noise
   c. Distortion
   d. Low hearing aid gain
   e. Peaks in the frequency response
   f. Low SSPL90s

3. A WDRC circuit is one in which:
   a. Amplification is only provided in frequency regions where hearing loss is present
   b. Compression is applied at high input levels so that the signal does not get clipped
   c. Amplification is given for soft- and mid-level inputs but is transparent for high-level inputs
   d. Digital signal processing is used to mathematically manipulate the output of the hearing aid to just below the patient’s uncomfortable loudness level (ULL)

4. The principal beneficiaries of high-fidelity amplification include:
   a. Individuals with profound hearing impairment
   b. Individuals with severe-to-profound hearing impairment in the high frequencies
   c. Individuals with a loss of sensitivity for low-level sounds and normal hearing for high-level sounds
   d. Individuals with poor discrimination abilities

5. One of the most important factors that determines the success of a hearing aid fitting is the:
a. Hearing aid  
b. Motivation of the patient  
c. Patient's type and degree of hearing loss  
d. Counseling given to the patient by the hearing aid dispenser  
e. Patient's age

6. What should the minimum bandwidth of a high-fidelity hearing aid be?  
a. 250–4000 Hz  
b. 125–6000 Hz  
c. 60–8000 Hz  
d. 500–6000 Hz

7. The input noise level of a modern hearing aid is determined almost entirely by the:  
a. Receiver  
b. Type of signal processing  
c. Size of the hearing aid  
d. Style of the hearing aid

8. What should be the maximum total harmonic distortion of CCIF intermodulation distortion for a high-fidelity hearing aid for output levels between 50 and 90 dB eardrum SPL?  
a. 1%  
b. 2%  
c. 5%  
d. 7%  
e. 10%

9. Normal-hearing listeners typically need a ____ dB SNR to understand 50% of words-in-sentences in noise.  
a. +1 dB  
b. +2 dB  
c. 0 dB  
d. +5 dB  
e. −2 dB

10. Today's directional microphones can be expected to provide a ____ dB SNR improvement in a typical room:  
a. 2  
b. 4  
c. 5  
d. 10  
e. 11

REFERENCES


