
■ PRINCIPLES OF HIGH FIDELITY HEARING AID AMPLIFICATION

MEAD KILLION ■

Why high fidelity hearing aids? The question might better be: What possible excuse is there for *low* fidelity hearing aids? Four possible excuses come to mind:

1. Most hearing aid wearers have become used to their low-fidelity hearing aids and won't want to change.
2. The hardware required to make sub-miniature high fidelity devices is not yet available.
3. They would be too expensive for any but the rich to afford.
4. Even if the hardware were available, we wouldn't know how to make high fidelity hearing aids, because we don't know what high fidelity means for someone with a hearing impairment.

This chapter is dedicated to removing those excuses.

The first excuse is well founded. Evidence that hearing aid wearers become so used to the defects of their aids that they

sometimes reject any change, even for the better, will be discussed later in this chapter. The excuse is, however, hardly relevant to the individual in whom we are primarily interested: the mild-to-moderate hearing-impaired individual who has so far rejected a hearing aid.

It is easy to show that the second excuse is invalid. The author has a prized cassette copy of a WFMT broadcast master tape in which the entire Chicago Brass Quartet concert was recorded from the output of flat-response, wideband hearing aid microphones he helped design in the early 1970s. The widespread (and mistaken) belief that there were no suitable transducers available to make truly high fidelity hearing aids was a prime motivator behind the author's doctoral research in the late 1970s. That research will be reviewed briefly in this chapter, accompanied by a discussion of what remains to be done in hardware development.

The third excuse is not consistent with the history of high quality audio devices. A successful high fidelity hearing aid that was too expensive for all but the rich would, in all probability, be quickly followed by a lower-cost version of almost equal or superior quality. All that is required is one successful high fidelity hearing aid design, regardless of cost.

The most difficult excuse to dismiss is the last one; an attempt to answer its implicit question will occupy most of this chapter.

THE BASIC REASONING

The first problem is to define a high fidelity hearing aid. Except for the *hearing aid* part, the definition would be easy: A high fidelity sound reproducer would be acoustically — and more importantly, subjectively — transparent. That is, the listener should receive the auditory sensation with the high fidelity system interposed between him or her and the original source of sound as he or she would have received listening directly.

This goal is difficult to achieve with prerecorded sound. Loudspeaker reproduction is heavily influenced by the acoustics of the listening room, so that neither the directional cues nor the frequency spectrum experienced in the 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 the orchestra in the middle of their heads.

Interestingly enough, if it weren't for the complications of the hearing impairment that creates the need for the hearing aid, a headworn hearing aid would stand a better chance of providing perfect fidelity than any other means of sound reproduction. 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. These are not just theoretical speculations. Experimental in-the-ear hearing aids whose accuracy of frequency response exceeded that of many expensive high-fidelity loudspeakers have provided, for the author as critical listener, nearly transparent "I forgot they were on" listening at concerts.

But the hearing impairment is what creates the need for a hearing aid in the first place, and it must be accommodated. Barfod (1972, p. 1) 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 Barfod'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 (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 an individual who has normal low-frequency hearing, for example. He or she would inevitably object to the sound quality, partly because the imperfections in the hearing aid are interposed between him or her and the natural low-frequency sounds that he or she can hear quite well *without* the hearing aid, and partly because the closed-mold "occlusion effect" (discussed later) will make his or her own voice sound abnormal. More recently, open-canal "K-BASS" fittings (Killion, Berlin, & Hood, 1984) have been used successfully for individuals with low frequency hearing loss, and normal high frequency hearing. In this case, a low-pass filter earhook removes the high-frequency

output from the hearing aid, leaving amplification in the 300 to 1500 Hz region without interfering with the wearer's normal hearing for high-frequency sounds.

This same principle has generally *not* been applied in the amplitude domain. Amplification is generally provided for both quiet and moderately loud sounds, even though there is substantial evidence that many mild and moderately hearing-impaired individuals have normal hearing for high-level sounds. (Although hardly scientific evidence, the author's observations convinced him that his mother had such a hearing loss. She showed absolutely no indication of any abnormality in hearing for high-level sounds, even though a mild-to-moderate loss at threshold was measurable at the speech frequencies and noticeable when someone talked to her too quietly.) Following the "don't amplify where there's no loss" principle in the presence of normal hearing for high-level sounds, it can be concluded that the hearing aid should do absolutely nothing for such sounds: it should be so acoustically transparent that it *subjectively disappears*.

This concept is so often misunderstood that it may bear one more restatement: If an individual has normal hearing for loud sounds, the hearing aid should, for loud sounds, neither stand in the way of his normal hearing nor give him or her unnecessary amplification.

The term *unity acoustic gain* or simply *unity gain* (i.e., 0 dB gain) is used to describe that condition in which the hearing aid provides neither amplification nor attenuation of the incoming sound. For example, a "hearing aid" consisting of a few brightly colored wisps of thread inserted into the ear canal would provide unity acoustic gain for *all* incoming sounds, regardless of level or frequency. (It might even work, indirectly, as a hearing aid if it induced the talker to speak slowly and enunciate more clearly!)

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; 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 be dependent on the sound level. Needless to say, the automatic-gain-control 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*.

Using the definition of the last paragraph, 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. We will return to consider the specific frequency responses, amplitude response, and time response required of such a hearing aid, but note in summary that we have been assuming that the candidate for a high fidelity hearing aid has some region of normal hearing.

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, an individual with no useful high-frequency hearing would presumably derive little benefit from wide-frequency-range sound reproduction. (Indeed, the person would

risk irritating nearby companions with a high-frequency feedback squeal that would be inaudible to himself or herself.) One of the successful uses of low-pass filter earhooks has been with high-power hearing aids and closed earmolds, for individuals with profound loss and a "corner audiogram," to eliminate the unheard-squeal problem.)

But there are presently no hearing aids theoretically well suited to those with the most common hearing impairment, a mild sensorineural hearing loss that appears restricted to a loss of sensitivity for low-level sounds, with normal hearing for high-level sounds.

Some individuals may accept *only* a high fidelity hearing aid. They reject present hearing aids because they aren't good enough to overcome all the other objections to wearing a hearing aid. Amplified sounds falling within their region of normal hearing are immediately recognized as unnatural and distorted. In the process of editing this section, a college classmate, a banker, called the author out of the blue about his hearing problem: "I don't have any problem at all except around a conference table when some jerk thinks it is cute to talk in low tones. Then I can't hear him. But the in-the-ear hearing aid I tried drove me bananas because it magnified everything."

The psychoacoustic and physiological evidence that many hearing-impaired individuals 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. Some of Toole's (1986) listeners had mild-to-moderate high fre-

quency hearing losses and still exhibited excellent reliability in their fidelity rating of high fidelity loudspeakers.

Such individuals were discussed by Steinberg and Gardner 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 is commonly referred to as *loudness recruitment*. Figure 3-1 (from Scharf, 1978) illustrates this phenomenon for a "typical listener with a 40 dB hearing loss due to a cochlear impairment." The findings of nearly complete recruitment (normal loudness sensation for high-level sound) in subjects with cochlear impairment is so universal, in fact, that some form of loudness-recruitment test is routinely included in diagnostic test batteries, aimed at excluding the possibility of hearing loss caused by a tumor along the eighth cranial (auditory) nerve. (As a philosophical aside: Most of us were taught to think of recruitment as "an abnormal growth of loudness." Although that definition can be justified as technically correct, it may have been unfortunate in its influence on hearing aid design: It directs our attention to the region of defective hearing rather than the region of normal hearing. Thus, we amplify *everything*. A more useful description of recruitment for someone with mild hearing loss might be "a loss of sensitivity only for quiet sounds," which would, perhaps, have led to a totally different design philosophy.)

A substantial amount of evidence has accumulated since the 1937 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 perception, 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:

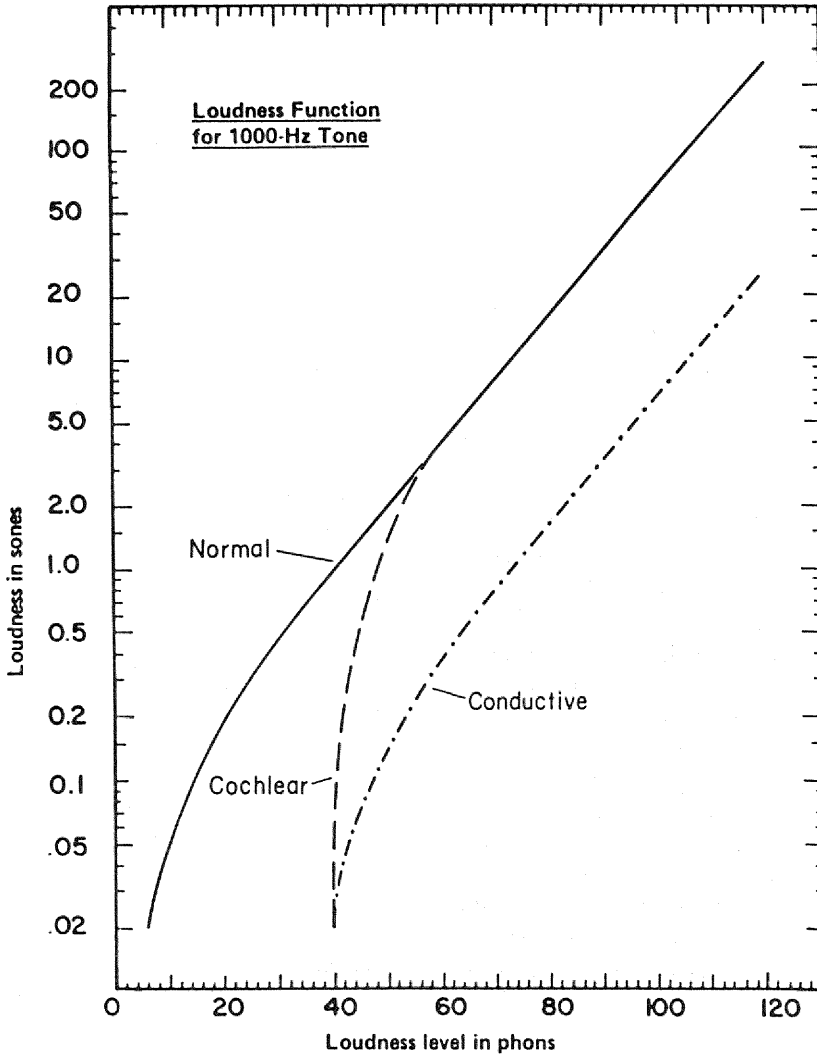


FIGURE 3-1. Loudness of a 1000-Hz tone as a function of loudness level for the normal listener, the typical listener with a cochlear impairment (and recruitment), and the typical listener with a conductive impairment. From Sharf, 1978. Reprinted with permission.

1. Frequency selectivity as determined from the "Fletcher Critical Band" which 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, 1978a)
3. Frequency discrimination (d1F) (Gengel, 1973)
4. Loudness summation for complex sounds (with cochlear impairment below 50 dB) (Scharf & Hellman, 1966)
5. Loudness discrimination (dII) (Scharf, 1978)

6. Localization abilities (see the summary given by Schafer, 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, 1982) or recorded from the ear canal in humans (Berlin & Gondra, 1976) is often normal. Indeed, both the waveform and latency of the entire auditory brainstem evoked response may appear entirely normal at high levels in some individuals (McGee, 1978b). Similarly, data obtained on laboratory animals with drug-induced outer hair cell damage (Dallos, Ryan, Harris, McGee, & Ozdamar, 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, whereas the primary function of audition appears to rest on the inner hair cells. There are about 20 auditory neurons for each of the inner hair cells, for example, compared to only 1 neuron gathering information from 10 or 20 outer hair cells. Post mortem hair cell counts on hearing-impaired patients with 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 hearing for high-level sounds as long as most of the inner hair cells function properly.

All evidence points to the existence of a large number of mildly hearing-impaired individuals who appear to have essentially normal hearing for high-level sounds. These would be the likely beneficiaries of high fidelity hearing aids.

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 hearing-impaired individual and the dispenser, any hearing aid that makes previously unheard sounds audible can provide a successful fitting. The old saw "wear it a few weeks and you'll get used to it" is as true — and as useful — today as ever.

The "getting used to it" process can be so successful that a change in the hearing aid creates a problem. Anyone servicing hearing aids is familiar with the 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 in understanding speech . . . and then find after a two or three week period that the loaner seems much better than his or her repaired original aid.

Even when the change is ostensibly an improvement, the change may not be welcome. The author has had two experiences where smoothing the response (by adding damping to the earmold tubing) for a long-time hearing aid wearer resulted 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 his or her old peak-clipping hearing aid. The extraordinary ability of the human being 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 to a badly degraded sensory input.

In summary, it is possible that the primary beneficiaries of high fidelity hearing aids would be those who do not now wear hearing aids. But it also seems plausible that a fair number of those current hearing aid wearers who find that their aids represents the best of a bad situation, those, for example, who complain that "I hear better in noise with my hearing aids removed," would also welcome a truly unobtrusive (i.e., high fidelity) hearing aid.

The important but as yet unanswered question appears to be: What would happen if wearing a hearing aid carried no *audible* disadvantages to the first-time wearer with a mild to moderate hearing impairment?

DESIGN GUIDELINES FOR HIGH FIDELITY HEARING AIDS

This section provides engineering guidelines for the design of hearing aids that can be reasonably expected to be judged high fidelity by those with mildly impaired hearing.

The guidelines for frequency response bandwidth and accuracy, and some of the guidelines for amplitude response accuracy (dealing with maximum output, allowable noise level, and allowable nonlinear distortion), can be offered with some confidence. Experimental hearing aids designed in accordance with those guidelines have been judged by those with normal hearing to have high fidelity; a fidelity comparable to that of a common (and expensive) recording-studio-monitor loudspeaker system.

The amplitude response and time response guideline dealing with AGC (automatic gain control) characteristics lie at the heart of the signal processing required to amplify quiet sounds up to audibility while

leaving loud sounds alone (unchanged by attenuation, amplification, or distortion). These guidelines can be offered only as reasoned arguments supported as yet by minimal experimental verification in the real world. If hearing aids constructed in accordance with these (or improved) guideline prove successful in providing transparent hearing aids for those with mild hearing impairment, better solutions for those with greater impairment may ultimately result.

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. Thus, the relationship between the sound pressure in a sound field and that found at the normal unaided eardrum must be considered.

Nature provides each person 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 headworn 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 region, in order to compensate for the gain the external ear itself provided before it was blocked by the plastic earmold. Little compensation is required at low frequencies, on the other hand, where the eardrum pressure

and the pressure available to the hearing aid microphone are essentially the same.

It is useful at this point to formally introduce the term *insertion gain*, which 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.

Restating the goal in terms of insertion gain, 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 previously discussed (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, etc.

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, and the modified Zwislocki (1970) coupler meets the second requirement up to 7 or 8 kHz (Sachs & Burkhard, 1972).

Figure 3-2 shows the Zwislocki coupler CORFIG curve 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 (over-the-ear), ITE (in-the-ear), and ITC (in-the-canal) hearing aids. The curves for the ITE and were obtained for a microphone inlet approximately flush with the plane of the pinna, since at the time the data were obtained most ITE aids filled the entire concha. (Throughout this chapter, the term OTE is used instead of BTE to emphasize that the data are for a microphone located over, and not behind, the ear.)

ROLE OF THE PINNA AND CONCHA. As illustrated in Figure 3-2, 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, p. 467). For microphone locations outside the pinna, such as the OTE, forward-looking-inlet location discussed, very 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 ear-mold 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 are the directional effects preserved in their entirety.

With regard to auditory localization, it is clear that any of the microphone locations used in binaural, headworn hearing aids will preserve the basic interaural time and intensity difference so important to

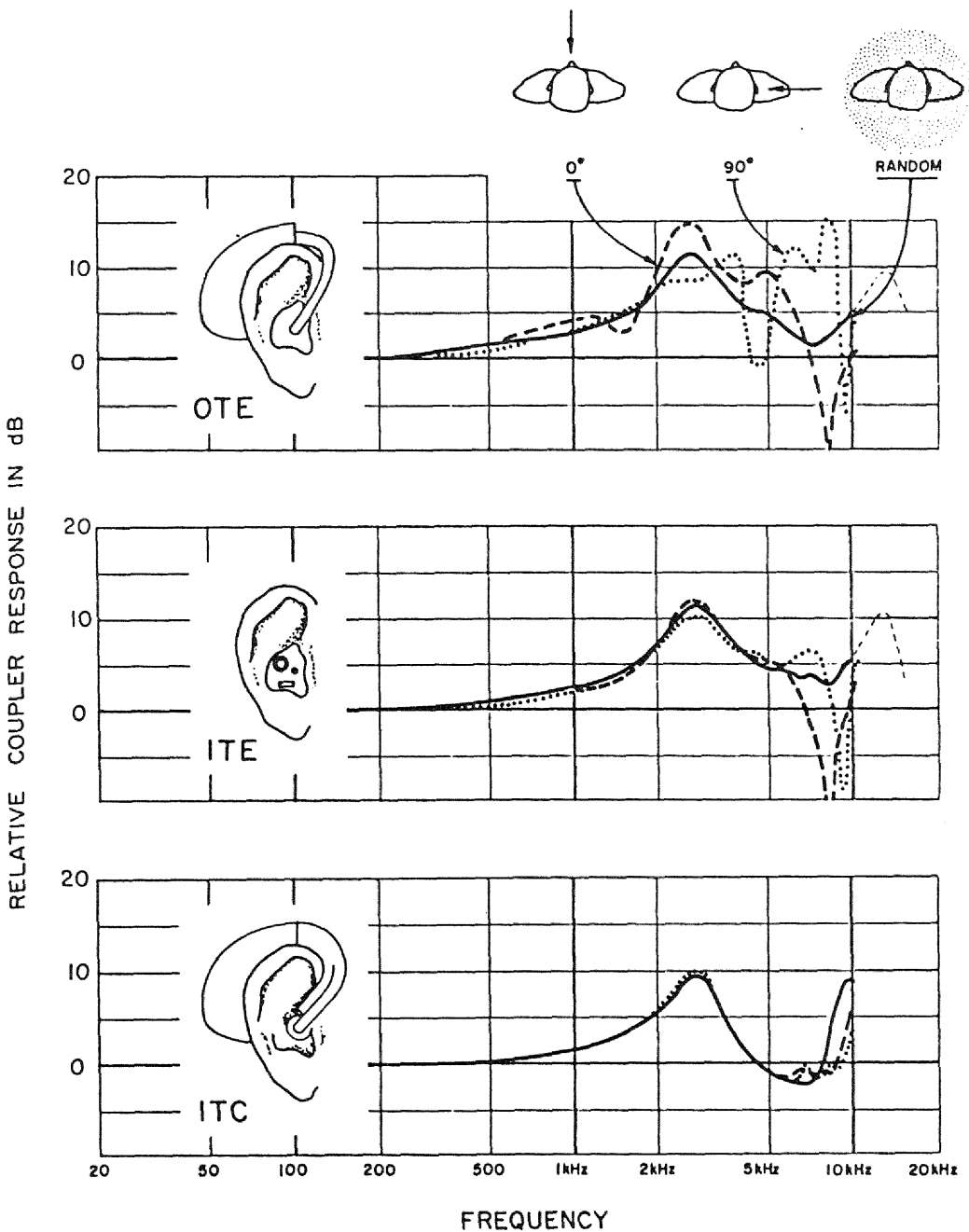


FIGURE 3-2. Required Zwislocki coupler response to produce flat insertion gain (CORFIG) for OTE, ITE, and ITC hearing aids under three sound field conditions. Note that only the ITC CORFIG is essentially independent of where the sound is coming from. From Killion & Monser, 1980. Reprinted with permission.

binaural location (Licklider, 1949), and the change in interaural phase and intensity caused by head motion which 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. It is clear from Figure 3-2 that it is impossible to design a conventional ITE or OTE 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 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 dominant consideration, the appropriate reference condition is probably a 0° incidence sound wave.

Where sound quality is the dominant criterion, other considerations apply. Most home and concert listening is done where 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 Schuelein (1975).

The ITC aid requires no compromise between conflicting frequency response requirements, since the coupler response that provides a flat insertion gain response for 0° incident sound provides a flat insertion gain response for *all* angles (and thus, trivially, for random-incidence sound as shown in Figure 3-2).

Bandwidth Requirements

After considering the available information on (1) hearing sensitivity and frequency limits of hearing of typical listeners, (2) measurements of the discomfort level of sound, (3) measurements of room noise in a wide variety of locations, and (4) measurements of the frequency limits in the maximum and minimum levels of speech, orchestral music, and various instruments of the orchestra, Fletcher (1942, p. 266) 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, p. 580) concluded that "the reproduction of orchestral music with *perfect* (author's emphasis) fidelity requires a frequency range of from 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 nearness to perfection required. Snow's listeners gave a judged quality rating slightly in excess of 90 percent to a system whose frequency range extended from 60 Hz to 14,000 Hz range, compared to unrestricted-range reproduction. With a 40 Hz to 14,000 Hz range, the quality rating was close to 100 percent. Recent technological advances have not changed these conclusions. Tests reported by Muraoka, Iwahra, 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 the author, where 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 3-5 shows their frequency response curves, labled OTE and MS, respectively. The irregularity in the frequency response of the loudspeakers was apparently judged as important a defect as the 8 kHz bandwidth limitation of the OTE aids.) The 8 kHz bandwidth hearing aids were rated *much* higher in fidelity than a pair of "Popular headPhones" (Koss PRO-4AA) whose exaggerated bass and treble response has been widely noted. The good fidelity ratings for the OTE aids held up under four different selections of program material, as shown in Figure 3-3, 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 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 3-3.

Response Smoothness

The question of how irregular the frequency response may be before it has a noticeable effect on the fidelity has not been as well studied as the effect of restricting the frequency range.

Bucklein (1982) 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 that he used appeared fairly sharp, with an apparent 3 dB bandwidth of

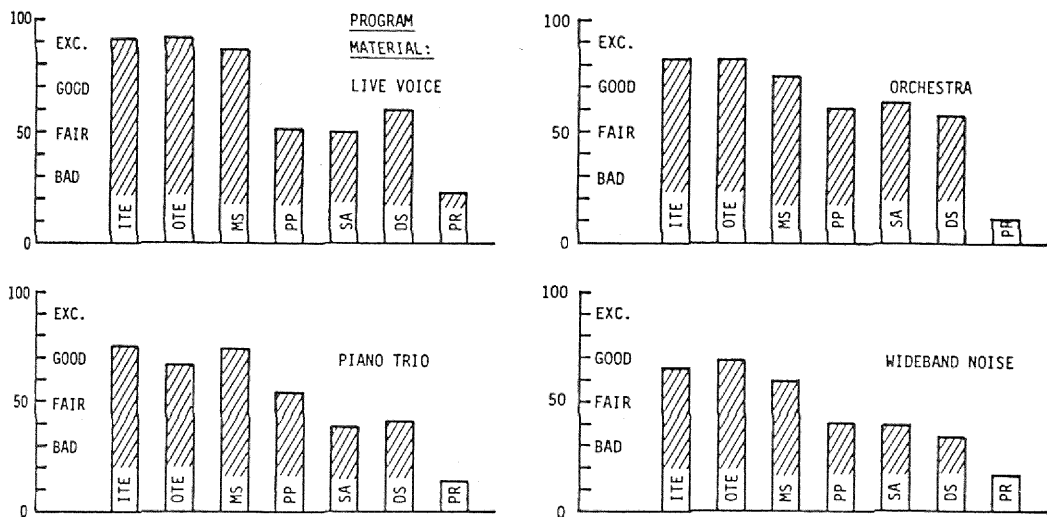


FIGURE 3-3. Fidelity ratings obtained with different program materials for experimental ITE hearing aids (ITE); experimental OTE hearing aids (OTE); Electro Voice Sentry V Studio Monitor Speakers (MS); Koss PRO-4AA Popular Choice Phones (PP); Simulated Speech Audiometer with TDH-39 earphones (SA); K-MART special \$69.95 "High Fidelity" Discount Stereo (DS); GE \$4.95 Pocket Radio in overload (PR).

roughly 10 percent of the center frequency. He found that 100 percent of his observers could detect both the 10 and 20 dB peak, but less than half could detect the 20 dB dip and only 10 percent 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 percent of the time.

Toole (1981, 1984, 1986) is in the process of publishing the findings from 10 years of investigation into high fidelity loudspeakers (and high fidelity in general). He has observed (personal communication) that the loudspeakers that receive the highest fidelity ratings exhibit frequency responses lying within a ± 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 (Dillion & 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 for rating the frequency response accuracy of high fidelity loudspeakers ("How CU's audio lab tests loudspeaker accuracy," 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 in that range, 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 taking the absolute value of that difference. Thus, a measured power equivalent to either 0.5 or 1.5 times the ideal loudness would produce a 50 percent accuracy score for that band. For example, a loudspeaker that had a perfectly flat "power response" everywhere except for a 7 dB dropoff (half loudness) in the 125 Hz band would have a calculated accuracy score of $(50 + 20 \times 100) / 21 = 98$ percent. For these purposes, this accuracy score will be called the "21-band accuracy score" to distinguish it from a "25-band accuracy score" discussed later.

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 percent and 93 percent, with a median value of 80 percent. Listening tests were said to have borne out the utility of the accuracy scores, although Consumers Union 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' score differ by eight points or less" (*Consumer Reports*, 1977 p. 406).

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., that doesn't buzz, crackle, or otherwise audibly distort) was confirmed in the author's listening test experiments (Killion, 1979a). Figure 3-4 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 program-material selections: 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 3-5. The insertion-gain frequency response on 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 nature's own 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 percent and an average fidelity rating (for those three program selections) of 82 percent. The experimental ITC 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 percent and an average fidelity rating (same program materials) of 88 percent. The studio-monitor loudspeakers had a dip of 8 dB in their listening-room response at 2 kHz (the cross-over frequency between the woofer and horn tweeter), with a calculated accu-

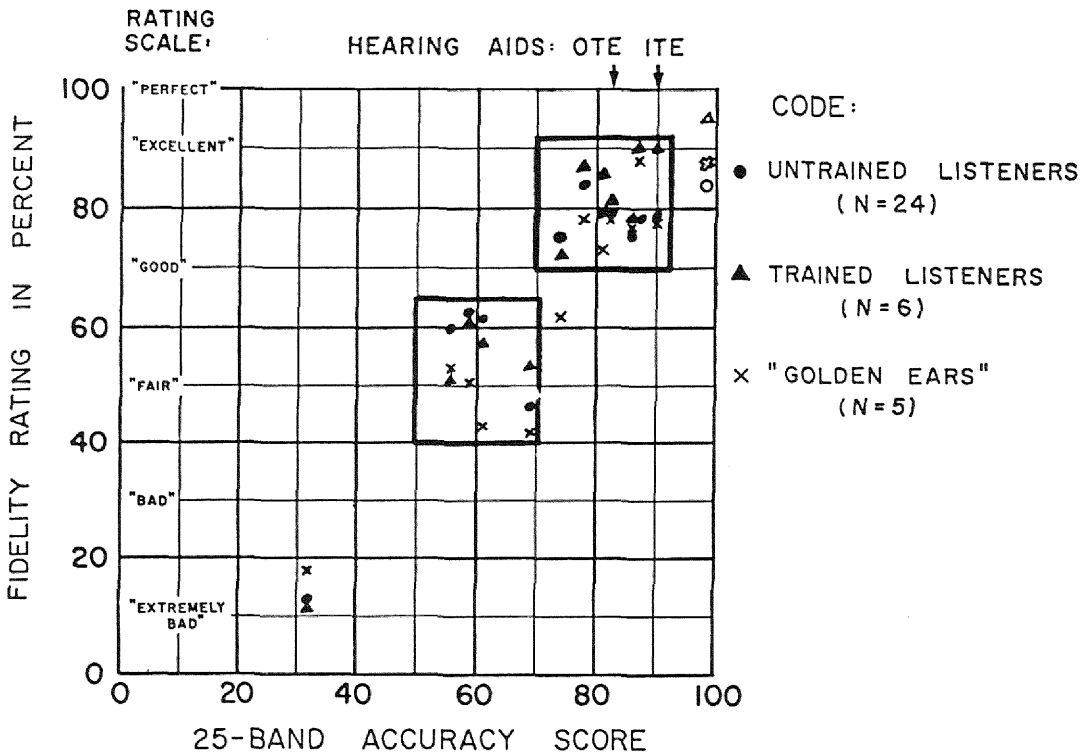


FIGURE 3-4. Comparison between average subjective fidelity (similarity) ratings and calculated 25-band accuracy scores for 12 systems rated on three identical program selections (orchestra, piano trio, male voice). Fidelity ratings obtained for a +3 dB shift in level as the only difference between the comparison and the reference are plotted as open symbols.

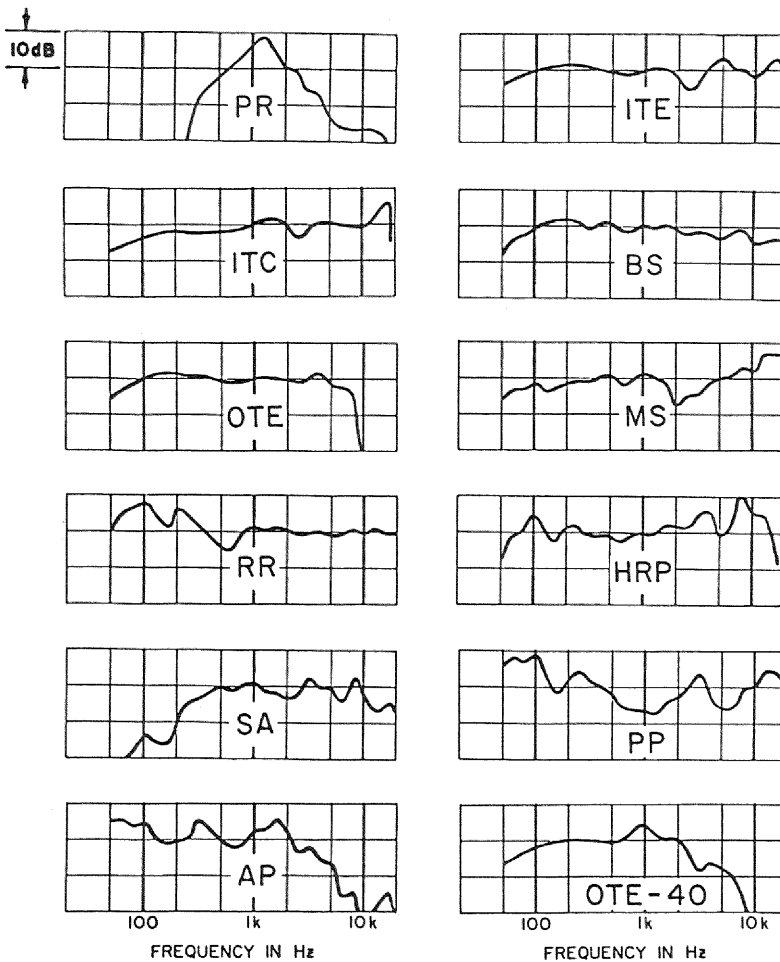


FIGURE 3-5. Frequency responses of sound systems used in complete listening tests and compared to calculated accuracy scores in Figure 3-4.

racy score of 83 percent and a fidelity rating of 80 percent. Finally, the OTE aids had an extremely smooth frequency response, with an 8 kHz high-frequency cut-off as their only real defect. This resulted in an accuracy score of 82 percent and a fidelity rating of 79 percent.

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") resonance 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, for a given individual, from the design average.

An estimate of the individual differences in outer-ear resonance has provided in the data of Filler, 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 1-2 dB below 1800 Hz to 4-7 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) by subminiature hearing-aid earphones. The standard deviation of the level of pressure developed ranged from approximately 1 dB at 1 kHz to 5 dB in the 6-8 kHz region. Greater pressure levels were developed in female ears (by 3 to 5 dB at the higher frequencies).

Although 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 the author's listening-test experiment where 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 percent. (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 percent to the 85 to 90 percent area.)

The subjective importance of such response inaccuracies to a long-term hear-

ing aid wearer has not been studied, although it is known that even longer deviations in unaided frequency response can occur due to the accumulation of earwax in the canal; deviations whose onset is so gradual that they often go unnoticed by the sufferer 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 necessary to provide modification for individual eccentricities.

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 percent as measured on the KEMAR manikin. Such a hearing aid would have a calculated accuracy score equal to or better than half the \$100 to \$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 percent) of the expensive (\$600 to \$1,000 per pair) "state of the art" loudspeakers tested in 1978 by *Consumer Reports*. In either case, the response accuracy goal would presumably apply 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 intermodulation), and any automatic gain control (AGC) functions as taking place in the amplitude or intensity domain. Some amplitude-domain performance requirements imposed on a hearing aid if it is to provide high fidelity sound reproduction will be discussed.

Peak Output Levels

There is no easy answer to the question of what the maximum undistorted out-

put 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 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 percent to 5 percent of $\frac{1}{8}$ -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 instantaneous peak levels at the microphone will be 90 to 95 dB SPL; a range that has often been used as a minimum undistorted input level goal 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; as can a child's enthusiastic greeting. These numbers are actually quite conservative: At 3 to 4 years old, the author's daughter could produce an enthusiastic 114 dB instantaneous peak SPL "Hi, Dad" at $2\frac{1}{2}$ feet. The author, a far-from-professional singer, can still sing a high "A" that has produced a 112 dB instantane-

ous peak SPL at the microphone of a headworn hearing aid he had 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, p. 76) 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 Grant Park or at the edge of the Ravinia stage, and the author has recently measured similar levels in a first balcony seat at Orchestra Hall. 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 four 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.

These more recent data are consistent with earlier measurements made of the Philadelphia Orchestra playing a wide variety of selections during a three hour recording session (Fletcher, 1942). At a measurement location 20 feet from the center of the orchestra, the instantaneous peak levels were estimated at 112 dB SPL.

All things considered, an undistorted input capability of 105 dB instantaneous peak SPL — referred to the sound field — appears to be a reasonable minimum requirement for a high fidelity hearing aid.

What is also needed for the 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

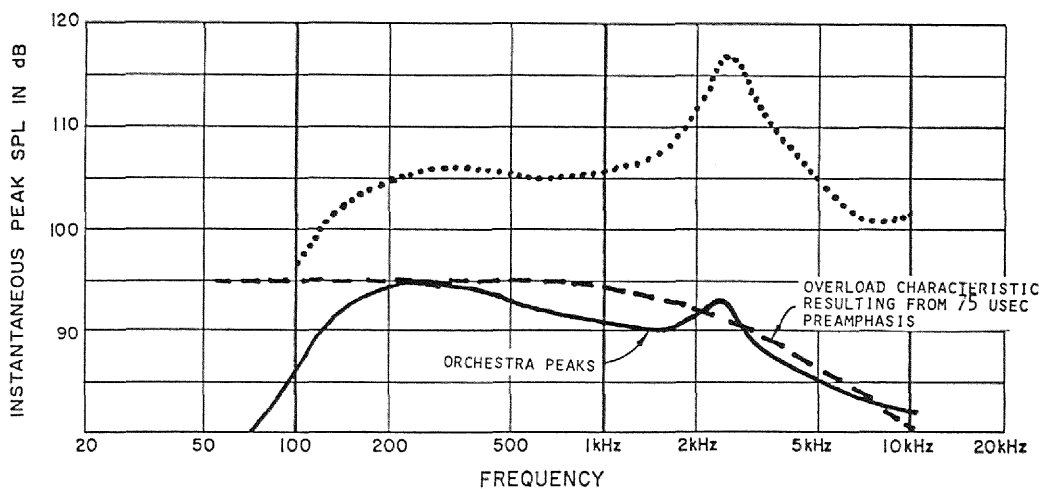


FIGURE 3-6. Maximum peaks produced in half-octave bands by large orchestra, referred to audience area (solid line). Instantaneous peak (of sine wave) eardrum pressure capabilities required of hearing aid to reproduce large orchestra (dotted line). (Note: Orchestra peaks correspond to instantaneous wide-band peaks of 105 dB SPL or peak readings of 95 dB SPL on sound level meter set to "C, Fast.")

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 3-6. 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 bought the 10kHz 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.

Since occasional high-frequency overload is generally found to be less annoying than a constant hiss, however, a "75 microsecond" high-frequency preemphasis (a 6 dB/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 microseconds. 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 recorded high-frequency overload capability.

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 3-6.

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 3-6. 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 which is 25 dB greater than the peak level shown in Figure 3-6, or a maximum of 117 dB SPL at 2700 Hz. The upper (dotted) curve in Figure 3-6 shows the estimated instantaneous peak (sinewave) output requirements for a high fidelity hearing aid operating at unity insertion gain for high-level signals. (Entirely different considerations apply when output limitation is used in *high-gain* hearing aids to prevent user discomfort. In that application, the maximum output of the hearing aid should probably *not* have a peak at 2.7 kHz).

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). 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 percent of the levels fell between 33 and 52 dB(A). Seacord's data 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 whose input noise level 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. 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, 1976). The internal noise of the hearing aid should thus present no impediment to a high fidelity design.

Distortion

The term *distortion* is used 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 one percent were just detectable — and total harmonic distortion levels below three percent 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 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 whose output level was 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 $2f_1$, $3f_1$, 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.

Tests made by the British Post Office and reported by Moir (1958, p.51) 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 three percent to four percent when the difference frequency lies in the 400 to 5000 Hz range.

More recent tests employing "golden ears" have produced similar results, as summarized by Milner (1977) and Davis (1978). Intermodulation distortion levels below two percent are generally inaudible on musical material, and even gross distortion levels (6 to 12 percent) are sometimes inaudible. In general, the just-audible distortion levels for musical material are at least 10 times greater than the just-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 re-

sults 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 value for detectable and tolerable distortion obtained by Olson and others can reasonably 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 percent 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 trans-

fer 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, one percent of soft peak clipping (as employed in the determinations discussed previously may be inaudible, but one percent 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 percent total harmonic distortion produced by one system had negligible effect on speech discrimination, whereas another system whose

total harmonic distortion measured only 20 percent 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 two percent for output levels between 50 and 90 dB eardrum SPL (assuming the distortion mechanism is a simple one), with a smooth relaxation to 10 percent at 30 and 110 dB SPL. Figure 3-7 illustrates such a requirement. Listening tests should be employed as a final check.

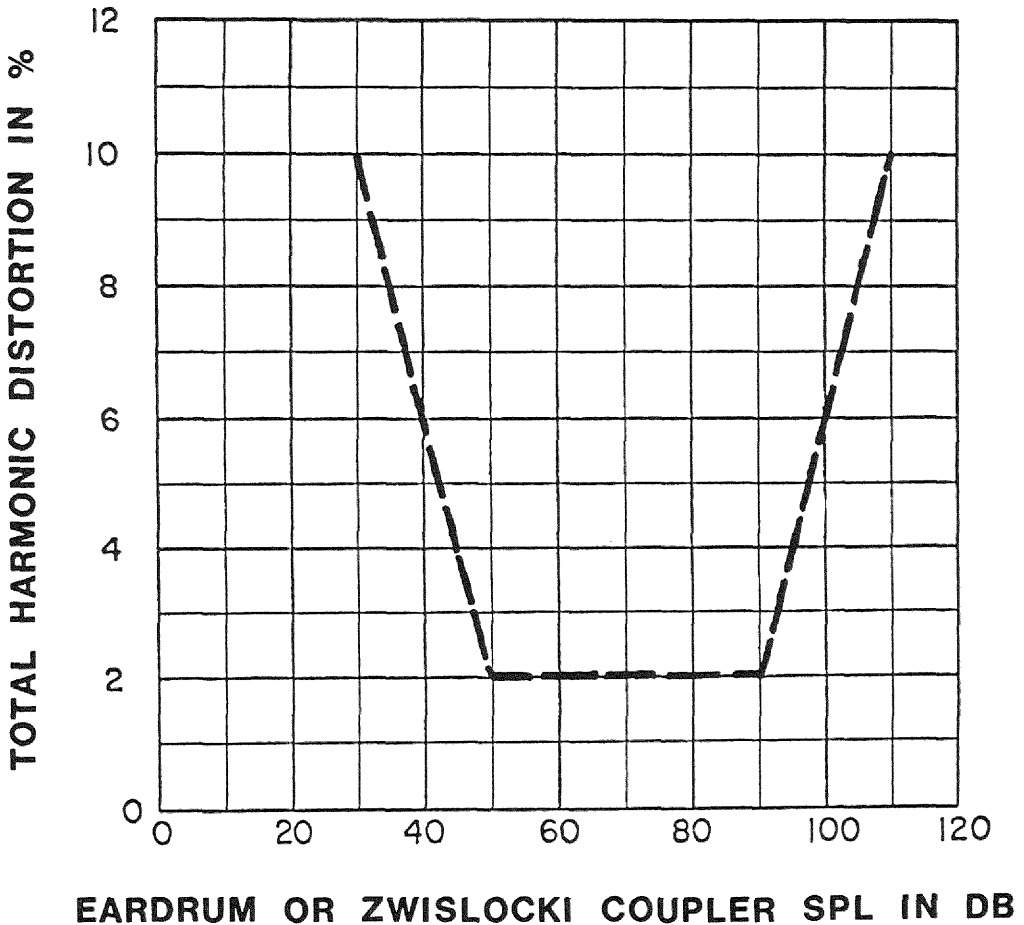


FIGURE 3-7. Estimated maximum hearing aid distortion that will be inaudible on speech or music.

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 or her threshold down to audiometric zero levels, such a large amount of gain is commonly found to be unacceptable (Lybarger, 1944; Martin, 1973).

Under most circumstances, the masking produced by the background noise level commonly encountered in residences, offices, etc., render even those with unusually acute hearing incapable of detecting sounds which are less than 15 to 30 dB above commonly accepted audiometric zero levels. Based on the average room noise and spectra data of Seacord (1940) and Hoth (1941), Killion and Studebaker (1978) estimated that the masking effect of typical residential room noise produces a nearly uniform 23 dB hearing loss across the 250 to 4000 Hz speech frequencies. (Similar estimates had been made by others. Thus, Olson (1957) calculated a 20 to 22 dB loss using

slightly different assumptions.) Gain more than that required to make background noises audible will be "empty gain," which makes everything louder but does not improve the detection of quiet sounds. Thus, a maximum gain sufficient to improve the aided threshold to 15 or 20 dB hearing level has been generally found appropriate.

The input level at which the gain should be reduced to unity can also be estimated. Examination 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, Barfod (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 Barfod'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. Figure 3-8 shows two hypothetical hearing losses with their corresponding areas of presumably normal hearing, based on Barfod's loudness data.

As a practical example, assume a user has a 45 dB HL cochlear impairment with

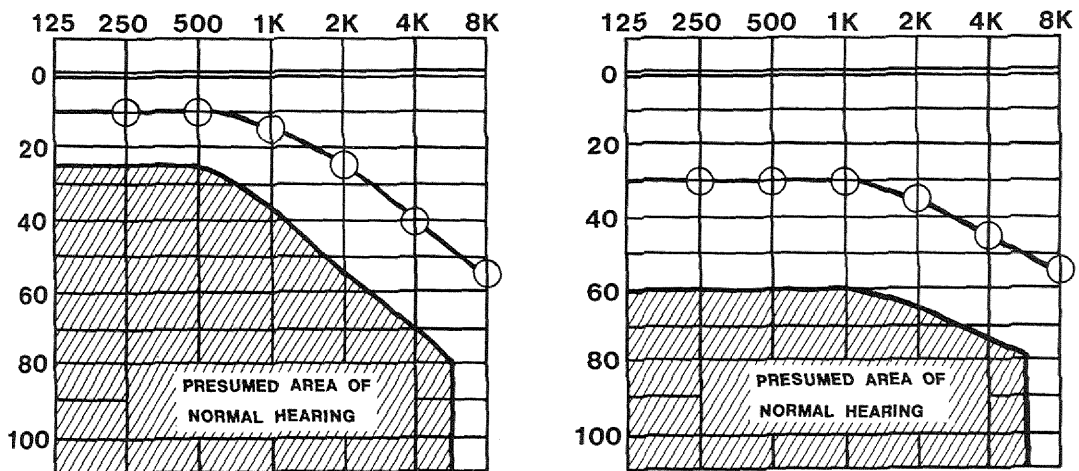


FIGURE 3-8. Threshold audiograms for two hypothetical subjects, with areas of presumed normal hearing based on Barfod's (1978) data.

complete recruitment for sounds above 80 dB HL. By these 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 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. In order to avoid constant adjustments of the volume control, an automatic gain control (AGC) system is required. In order to introduce the minimum degradation in perceived sound quality, the operation of the AGC system must be unobtrusive. Compression amplification had found wide acceptance in the broadcast and recording industry for such purposes (Blessner & 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 an old one, 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 3-9 illustrates one possible input-output characteristic (Killion, 1979b) for a hearing aid intended to meet the requirements of the previous example. There are four stages of amplification illustrated in Figure 3-9: a low-level constant-gain stage, a mid-level constant-compres-

sion-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 to 115 dB SPL) serves to prevent audible distortion or bias shifts in the amplifier associated with overload. Output limiting should not be required to prevent sounds from becoming uncomfortably loud if the hearing aid has unity acoustical gain for high-level sounds; the user would be exposed to uncomfortably loud sounds no more often with his or her hearing aid than without it. Even with unity gain in a hearing aid, however, the author has observed that the wideband "spectral splatter" accompanying amplifier overload can make an otherwise innocuous sound uncomfortably loud.

Note that the input-output characteristic illustrated in Figure 3-9 is quite different from any of the three types commonly employed in hearing aids and shown for comparison in Figure 3-10. 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 leaves a large proportion of sounds *almost* too loud. An automatic volume control can ensure 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 3-9, 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.

LEVEL DEPENDENT FREQUENCY RESPONSE.

Few threshold audiograms are flat. A greater

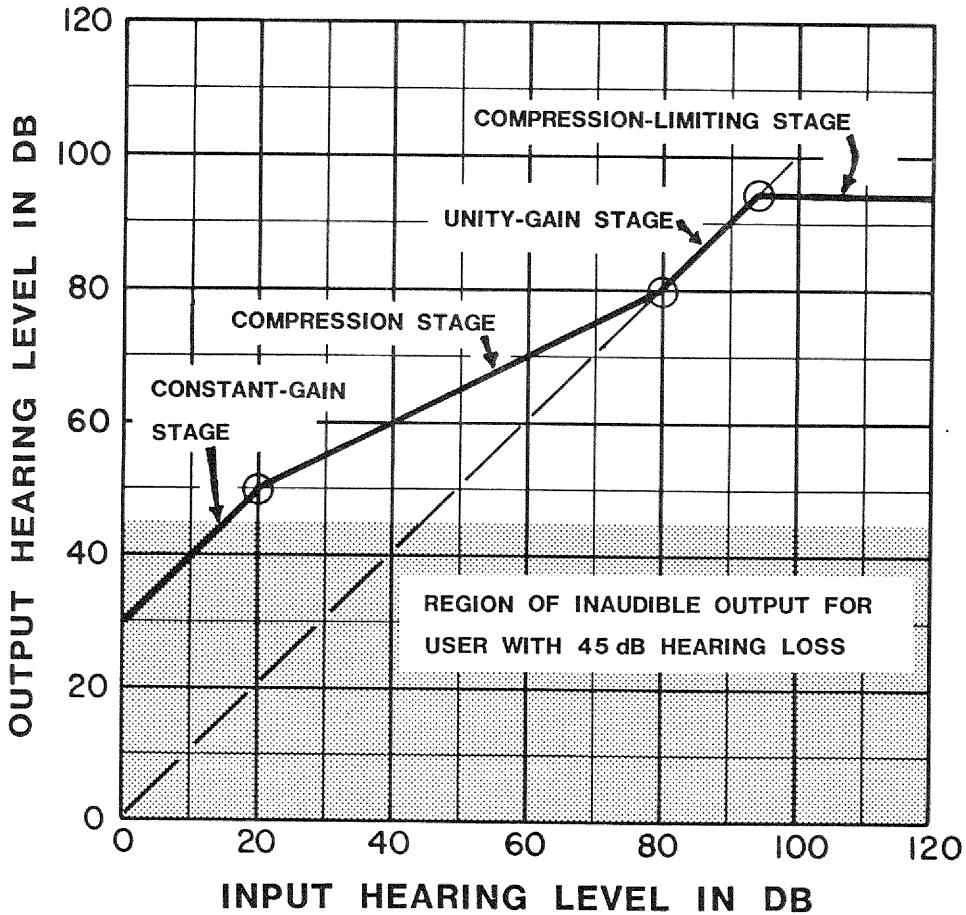


FIGURE 3-9. Presumed ideal hearing aid input-output characteristics for listener with 45 dB hearing loss and complete recruitment above 80 dB HL.

loss at the higher frequencies is commonly found in those with hearing loss. For low-level sounds, in such cases, a high-frequency emphasis in the hearing aid's frequency response is required to make the entire range of quiet speech sounds available without overamplifying the low frequency sounds. For high level 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 described most often 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 (Villchur, 1978).

A simpler method that may well be 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 fairly easily accomplished in a practical hearing aid amplifier, Figure 3-11 shows the level-dependent frequency response characteristic of the experimental four-stage compression amplifier developed for the author's experiments (Killion, 1979b). A greater high-frequency emphasis at low levels would be required to meet the needs

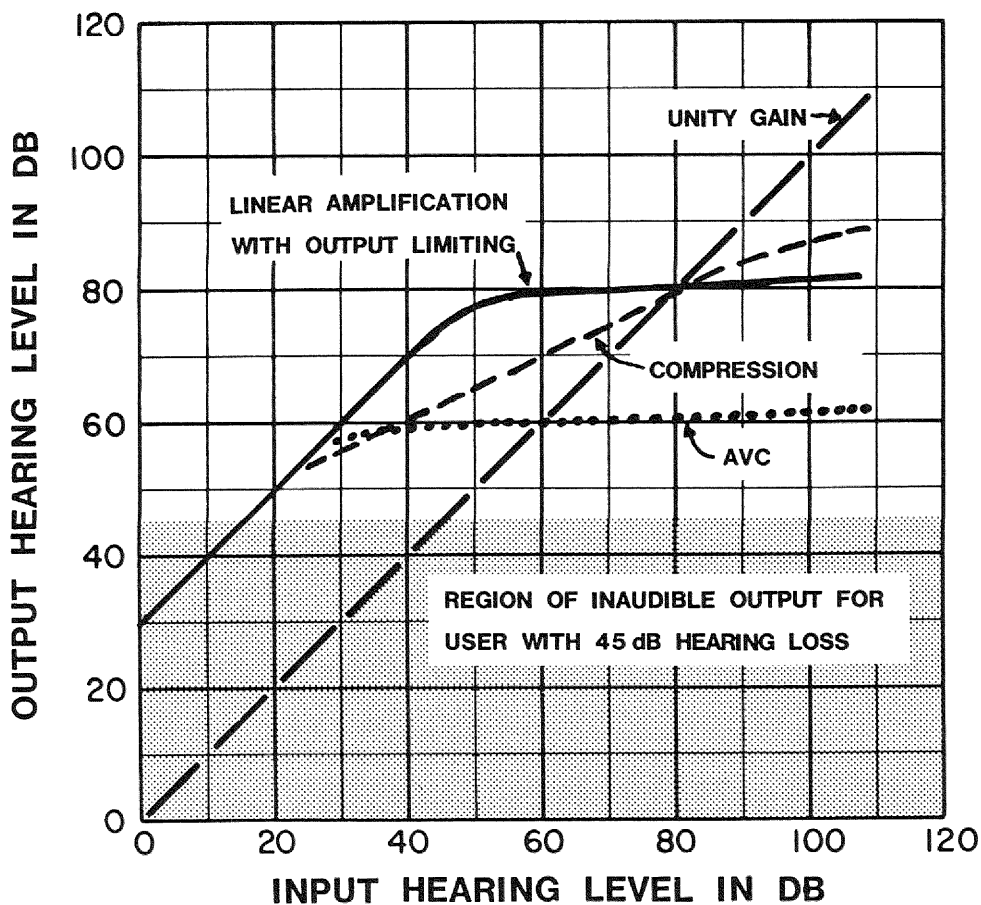


FIGURE 3-10. Input-output characteristics for three conventional approaches to dynamic range reduction in hearing aids.

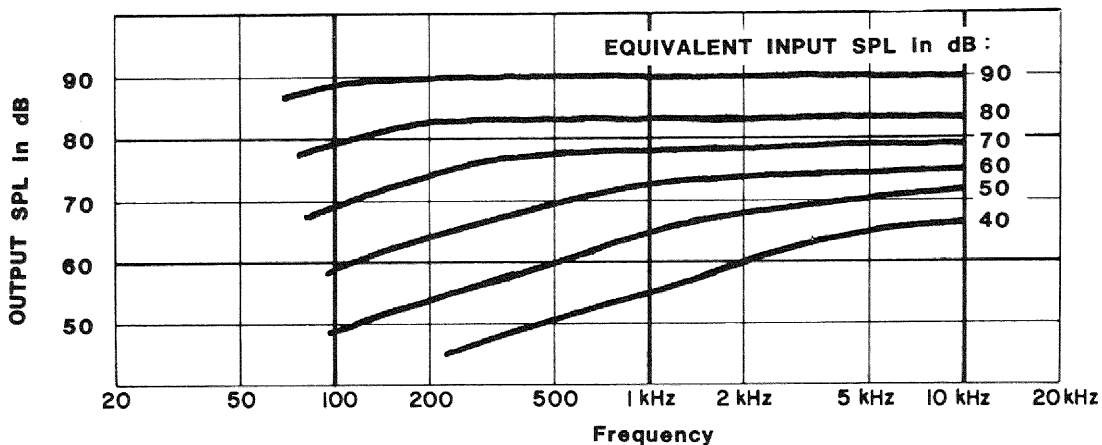


FIGURE 3-11. Relative amplifier frequency response versus input level in "speech mode" (compression ratio set for 2:1) of author's experimental amplifier.

of the hypothetical individual whose audiogram was shown in Figure 3-8, but that, too, should be practical.

Just as the idea of applying compression amplification to hearing aids is an old one, the need for a level-dependent frequency response has been discussed by Goldberg (1960, 1972), Villchur (1973, 1978), Barford (1976), and Skinner (1976, 1980).

MAINTAINING FIDELITY. 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. This can be achieved. Experimental high fidelity OTE aids with prototype four-stage compression amplifiers (Killion, 1979b) intentionally operated in their worst-case operating range (from the standpoint of measured distortion) but in their flat-frequency-response mode, received across-subject-group average fidelity ratings insignificantly different (2.5 percent higher) than the ratings given the OTE hearing aids themselves (Killion, 1979a, p. 212), 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 of Hearing Aids

Transient Waveform Response

There are two common interpretations for the term *poor transient response*. One is the difference between the output and input waveforms viewed on an oscilloscope screen (transient waveform 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 this definition, for example, the ear itself has very 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 waveform 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-waveform resultants was given by Mott (1944). The surprisingly large changes in waveform 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, experiments indicating that introducing even 90-degree-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 which are claimed to be caused by phase changes along when in fact larger (and 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 previously, 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

high-amplifier-distortion condition lasting much longer than the transient (Ingelstram, Johanson, Pettersson, & Sjögren, 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 characteristic throughout the operating range are an important consideration in any sound processing system.

As discussed by Lippman (1978), the proper choice of time constraints depends a great deal 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 msec, 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 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 by Blesser and Ives (1972), who reported that values of 10 msec and 150 msec 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 (Hotvet, 1988) fol-

lowing an earlier RCA development (Singer, 1950) that has been used for some time in the broadcast industry. This "adaptive compression" technique automatically varies the release time as a function of the duration of the high-level signal that engaged the compression action (Smirga, 1986). Informal listening test demonstrations indicate that a substantial improvement is obtained in some listening environments compared to traditional fixed-release-time hearing aid compression circuits.

Generally speaking, a separate compression circuit will be required to handle input sounds that are so intense that compression is needed to prevent audible distortion due to amplifier overload. For this latter purpose, a simpler circuit may suffice. The attack time should be as short as possible. Since the compression limiting is needed only for occasional transient peaks, the release time can also be made quite short — perhaps 50 msec or so — because the resulting distortion of low-frequency signals will be so brief as to be unnoticed.

IMPLEMENTATION OF THE GUIDELINES IN PRACTICAL HEARING AIDS

Many of today's hearing aids have a fidelity so much better than that of hearing aids two decades ago that there is already no excuse for fitting a low fidelity hearing aid. Nonetheless, there are not, to the author's knowledge, any no-holds-barred high fidelity hearing aids yet available. The problems in producing such aids will be discussed.

The Easy Part

The transducers, amplifiers, frequency response shaping, and coupling required for high fidelity hearing aids are all available now or potentially available. This has been clearly demonstrated both in objective engineering data and in the results of the author's fidelity rating experiments. The

only piece of hardware missing is a commercial version of the low-distortion four-stage compression amplifier described previously.

The Harder Part

There are still some difficulties in producing hearing aids that will be judged transparent in the real world, however, as discussed in the following sections.

Closed Ear Effects

ENHANCED BONE CONDUCTION. The increase in the level of bone-conducted sound when the ear canal is occluded is well known, and has been used for nearly 90 years in the Bing tuning fork test for differential diagnosis of conductive versus "nerve" types of hearing loss. Data obtained by Watson and Gales (1943) and Tillman (1962) indicated roughly 20 dB of low-frequency enhancement of bone-conducted hearing can occur when the ear canal is occluded with a well-sealed earmold. The more recent data of Berger and Kerivan (1983) indicate that the occlusion effect depends strongly on the depth of insertion of an earplug. He obtained 20 to 25 dB at 125 Hz and 20 dB at 250 Hz with a V-51R ear defender or a shallow-insertion EAR plug, but only 5 to 10 dB with a deeply inserted EAR plug. This latter result is consistent with the 5 to 10 dB occlusion effect obtained by Kruger and Mazor (1988) on eight subjects using deeply inserted ER-3A insert earphone eartips.

The experiments of Queller and Tonndorf (Khanna, Tonndorf, & Queller, 1976), Berger and Kerivan (1983), and Killion, Wilber, and Gudmundsen (1985) all indicate that the occlusion effect is dominated by the air conduction pathway, and is caused by increased sound pressure developed at the eardrum. With the ear canal open, the normal dilational vibration of

the canal wall develops little sound pressure at the eardrum because it all "spills out" the open ear canal. When the ear canal entrance is blocked, however, the sound generated by canal-wall vibration is trapped.

Someone with normal hearing wearing a high fidelity hearing aid set for unity gain may well find that the hearing aid subjectively "disappears" for external sounds. But if it employs a shallow, closed earmold, the enhanced level of autogenously generated sound (chewing, swallowing, etc.) may make such normally unnoticed sounds quite noticeable. In some cases, interference with normal communication is possible. The author has found the crunch of "Wheaties" so magnified that he occasionally missed parts of breakfast-table conversation while wearing unity-gain ITE aids. (The problem generally disappeared near the end of the bowl as the cereal became sufficiently soggy.)

SPEAKING IN A BARREL. Probe-tube measurements of the SPL developed in the ear canal behind relatively shallow, probably typical BTE earmolds were reported by Killion, Wilber, and Gudmundsen (1988). Their data showed:

1. The closed vowels "EE" (/i/) and "OO" (/u/) showed an occlusion effect of 20 to 30 dB at low frequencies, presumably because the pathway from the throat to the outside was relatively closed off, and the resulting high SPLs (roughly 140 dB) developed in the mouth cavity caused vigorous cheek and jaw vibration, which is coupled to the ear canal wall.
2. The open vowel "AAAH" (/a/) showed little occlusion effect, presumably because the pathway from the throat to the outside was quite open and a relatively low SPL (roughly 115 dB) developed in the mouth cavity.
3. Surprisingly high SPLs were developed in the closed ear canal: 80 to 95 dB for the closed vowels.

The vowel dependence (and thus, apparent frequency dependence) of the occlusion effect is the likely basis for the complaint of unvented-earmold hearing aid wearers that their voices sound as though they are "talking in a barrel." To explain: The first formant of the closed vowels occurs at a low frequency, and the occlusion effect is greatest for the closed vowels. The first formant of the open vowels occurs at a higher frequency, and the occlusion effect for open vowels is minimal. The subjective result is much like talking into a low-frequency resonator. Since eardrum SPLs of 80 to 95 dB at 250 Hz correspond to hearing levels of 61 to 76 dB (Killion, 1978), even someone with a 50 dB low-frequency hearing loss should find the occlusion effect quite noticeable.

Earmold Venting

Venting the earmold is the time-honored solution to this problem. Where the individual has normal low-frequency hearing, venting may also provide nearly perfect fidelity hearing for low frequencies: Venting allows low frequency sound to enter the ear canal unattenuated, and under the right circumstances reduces the low frequency output of the hearing aid sufficiently to prevent it from intruding. Substantial venting is required to eliminate the audible occlusion effect, however. A vent with a 500 Hz cutoff frequency provides only 12 dB reduction in the occlusion effect at 250 Hz, whereas a 20 to 30 dB reduction appears to be required to eliminate the occlusion effect completely with a shallow earmold. Figure 3-12 shows the effect of various amounts of venting on the output of an idealized flat-response hearing aid and on the occlusion effect for one subject saying "EE."

An open-canal fitting of an OTE aid solves the problem, of course, but at best permits only 25 to 30 dB of high-frequency gain before squealback sets in. (As Cox [1982] correctly points out, feedback can be

oscillatory [squealback], not oscillatory but still audible [as when the frequency response of the hearing aid exhibits a sharp peak caused by incipient feedback], and inaudible).

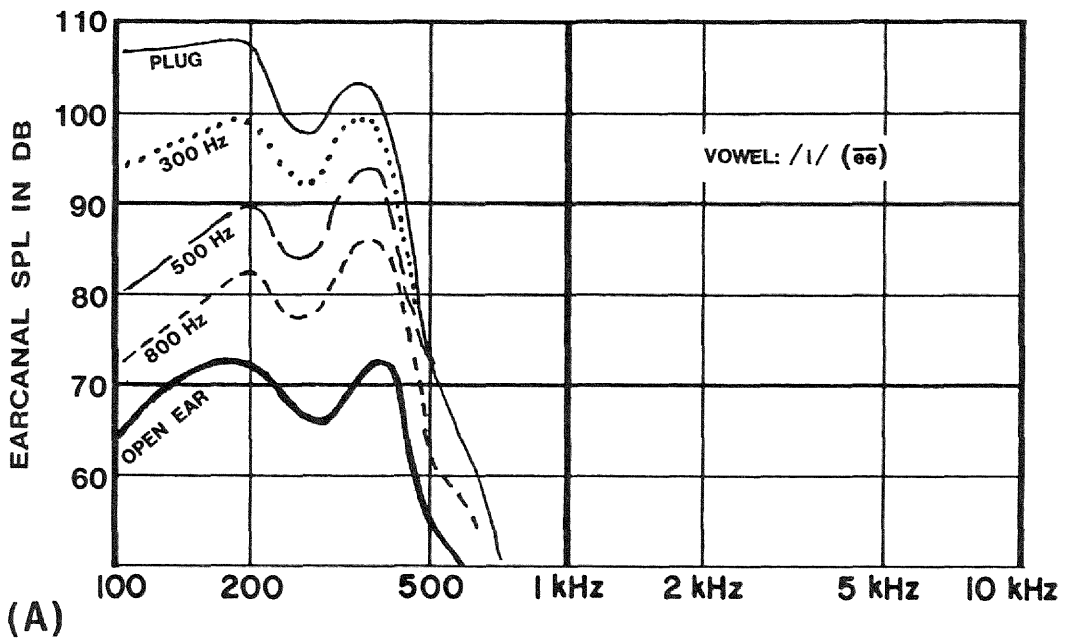
Open-Canal Fittings

Where the maximum loss of sensitivity for quiet sounds is less than 40 dB a gain of 15 dB will often be adequate, and a gain of 25 dB will almost always be adequate: A gain of 25 dB will reduce the hearing loss to 15 dB, less than the loss due to background noise levels that someone with *normal* hearing experiences in a typical living room or office (Killion & Studebaker, 1978).

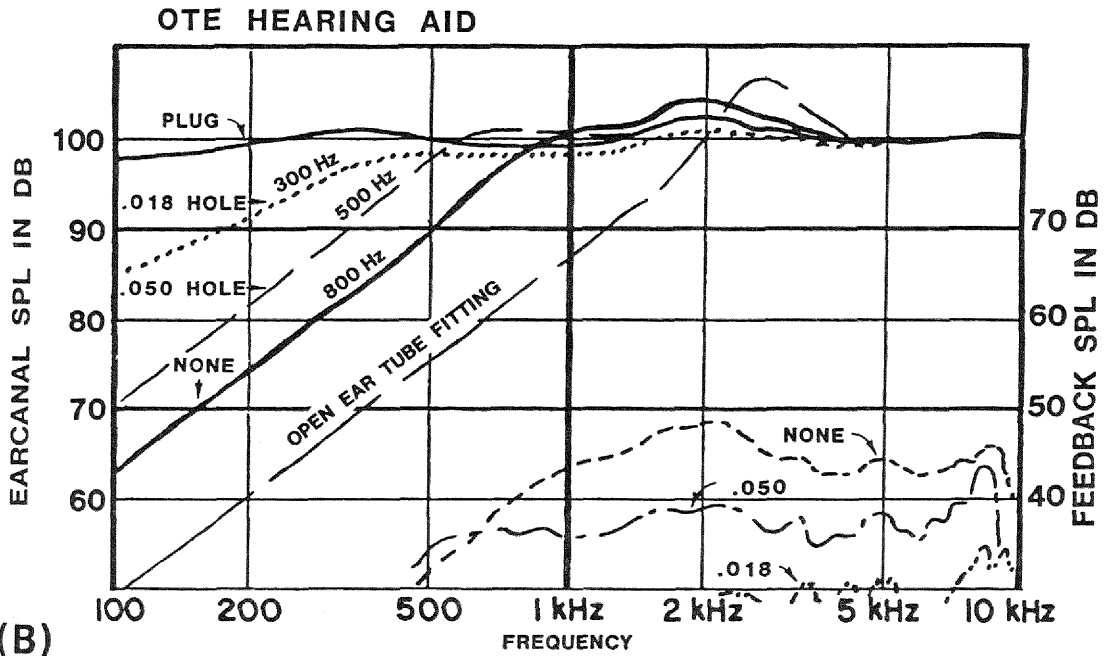
If only 25 dB of maximum gain is required, an open-canal fitting becomes practical as long as the frequency response is free of peaks: The KEMAR manikin eardrum-pressure and relative-feedback-SPL data reported by Killion and Wilson (1985) indicate that at least 27 dB of gain, with a 5 dB margin of safety before squealback (oscillatory feedback), can be obtained over the 100 to 10 kHz range as long as the eartube is inserted less than 12 mm into the ear canal.

The acoustical effect of the eartube, or of a properly constructed "free field" earmold in the ear canal, is minimal in an open canal fitting: no effect at low frequencies and perhaps a 3 dB loss in eardrum SPL above 3 kHz. This suggests that the simplest high-fidelity hearing aid for someone with normal high-level hearing would be based on an OTE aid coupled to a free field earmold. A variation of the K-BASS low-frequency response coupled with the right eartube would permit a relatively smooth insertion gain from perhaps 300 Hz to 6 to 8 kHz while leaving the ear canal completely open.

Nearly perfect fidelity can be obtained at high levels in an open-canal fitting, because the AGC circuit would reduce the hearing aid gain well below unity gain,



(A)



(B)

FIGURE 3-12. Illustration of the similar effect of venting on the low frequency SPL in the ear canal for sound produced by (A) the occlusion effect or (B) the output of a hearing aid, with the vent cut-off frequency as parameter. In both cases, the venting was accomplished with a SAV insert in a short, large vent (8 mm long by 3 mm diameter). The subject vocalizing "EEE" was a 51 year old male (A). The hearing aid was a laboratory aid designed with flat frequency response at the eardrum (B).

leaving only the unaltered natural sound arriving at the eardrum.

At low levels, on the other hand, an appropriate AGC circuit will increase the gain to 20 to 25 dB, providing the required amplification for low-level sounds. (The appropriate *amplifier* input-output function for an open-canal fitting will differ from that shown in Figure 3-9. As the unity-gain condition approaches, the amplifier gain should quickly drop to a very low value so that the hearing aid can get out of the way.)

An open-canal aid will not be required to provide amplification above, for example, 90 dB SPL, because no gain is required for such high-level sound. Thus, the power requirements for the hearing aid will be modest. A hearing aid that produced 115 dB maximum SPL in a closed-canal fitting would still produce 90 to 95 dB SPL at 500 Hz in an open-canal fitting (Killion & Wilson, 1985), allowing even the inefficiencies of an open-canal fitting without undue battery drain for amplification down to 400 to 500 Hz.

Note that there will be no occlusion effect with an open-canal fitting.

Preliminary experiments in the author's laboratory indicate that only two problems can be expected in attempting to produce truly high fidelity OTE hearing aids for use with free-field earmolds: providing a low-distortion compression amplifier with the proper input-output function, and maintaining an appropriate phase response in the hearing aid so that severe cancellations cannot occur.

As an unlikely example for the sake of illustrating the cancellation problem, assume the hearing aid had been designed to produce a flat insertion gain response between 500 and 6000 Hz, and its output was exactly 180 degrees out of phase with the natural sound in the ear canal. As the sound level dropped and the hearing aid gain increased, at some point the output from the hearing aid would exactly equal — and thus cancel — the natural sound in the ear canal: The wearer would suddenly hear

nothing between 500 and 6000 Hz! In practice, no one would design such an amplifier (or would at least reverse the polarity of the earphone once the problem became obvious), but real-life engineering difficulties occur at the upper and lower cut-off frequencies of the hearing aid, where a substantial amount of phase shift is likely to occur. Figure 3-13 shows the cancellation effect for a special purpose open canal bandpass hearing aid providing amplification only between 500 and 2500 Hz. Note that at about 400 and 3000 Hz the sound entering the open ear canal partially cancels the output from the hearing aid, providing a 5 to 10 dB insertion loss.

Closed-Canal Fittings

Perhaps unfortunately, as the amplifier and transducer technology reached the point where high fidelity OTE hearing aids became relatively easy to produce using free-field earmolds, canal aids became popular. Unfortunately because a canal aid will come closer to a vented closed-canal fitting than an open-canal fitting. In particular, the physical presence of even the smallest canal aid will cause some occlusion effect and an attenuation of 15 to 20 dB in the high frequency sound reaching the eardrum when it is turned off, so the aid itself must provide transparent, unity-gain amplification for high level sounds at all but the lowest frequencies.

Although a greater proportion of potential hearing aid purchasers than it appears from the sales statistics might be better satisfied with an ITE or OTE aid than a canal aid, there is still some merit in considering canal aids for someone with only a mild hearing loss, if for no other reason than their ready acceptance. All other things being equal, most people would rather have a hearing aid the size of an eraser tip than one the size of a typical OTE or ITE aid. The observation that the location of the microphone in a canal aid allows it to partake in virtually all of the

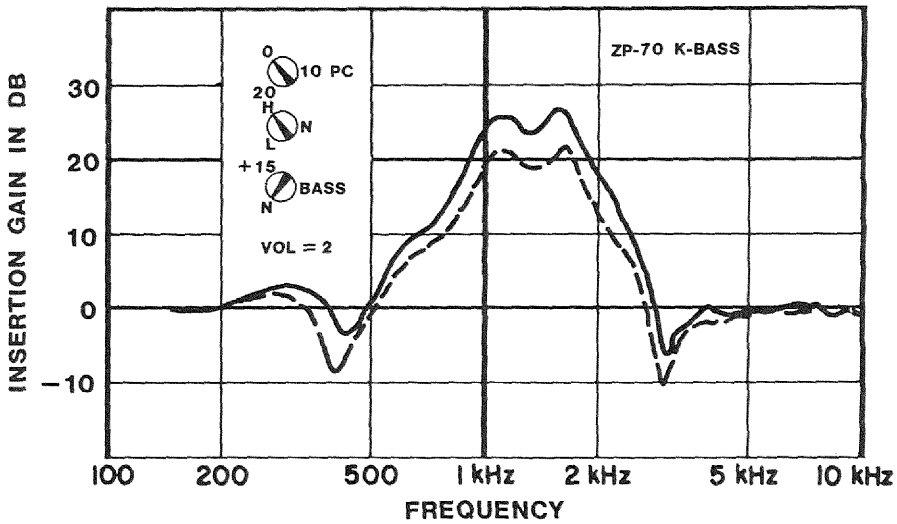


FIGURE 3-13. Illustration of insertion gain minima for two different volume control settings of special purpose bandpass hearing aid in an open canal fitting.

directional properties of the ear may be overplayed relative to its perceptual importance, but it is still true, and gives a theoretical advantage to the canal aid as potentially the highest-fidelity aid in terms of auditory spatial perception.

So a reasonable question becomes: How equal can we make a canal aid? The author's experience indicates that canal aids whose frequency response and ear-phone distortion levels — even for high-level sounds — can meet the highest fidelity standards can be built.

This leaves the occlusion-effect problem. Fortunately, canal aids virtually free of the occlusion effect problem appear practical for some individuals in light of two recent developments. In both cases, the secret was to affect a seal deep within the earcanal: with a soft rubber “bulb” in the case of some of the Voroba Technology Associates modular canal aids (Voroba, 1987), and with surgical reconstruction of the earcanal in the case of the RESOUND deep-canal aid (Perkins & Goode, 1987). Killion, Wilber, & Gudmundsen (1988) reported examples of both, where ear canal SPLs measured while the subjects vocal-

ized were no greater behind deeply sealed hearing aids than in the open ear.

An additional advantage of deeply sealed canal aids is that the gain and maximum output of the hearing aid itself is increased because of the reduction in residual ear canal volume. This increase can amount to 3 to 4 dB at low frequencies and 6 to 8 dB at high frequencies. Figure 3-14 shows the increase in real ear insertion gain over 2 cc coupler gain for OTE, ITE, and both conventional and deeply sealed canal (ITC) hearing aids. These curves have been labeled “GIFROC” curves because they are the inverse of the “CORFIG” curves of Figure 3-2 (GIFROC is CORFIG spelled backwards). The first three curves are based on recent probe-tube measurements of 16 ears in a diffuse sound field facility (Killion, Berger, & Nuss, 1987); the last curve is estimated from computer calculations using an analog model of the ear. In addition, since most hearing aids are measured in a 2 cc coupler and not a Zwislocki coupler, the Zwislocki-coupler-to-2-cc-coupler corrections of Sachs & Burkhard (1972) have been included in Figure 3-14.

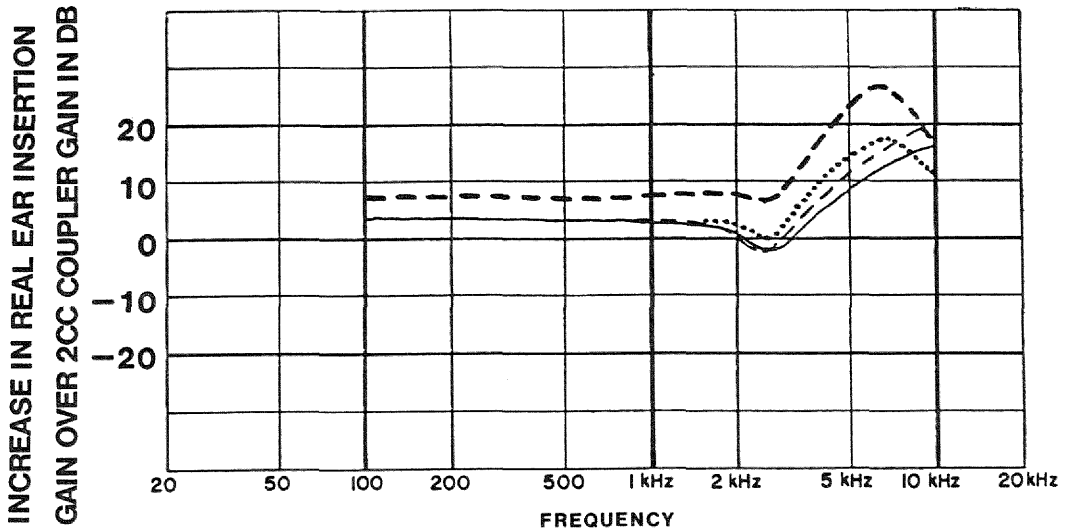


FIGURE 3-14. 2 cc coupler "GIFROC" curves for diffuse sound field and four types of hearing aids: (—) OTE (BTE); (----) ITE; (....) small canal aid; (-.-.-) very deeply sealed (long) small canal aid.

Based on these considerations, the author believes that true high fidelity hearing aids of both constructions (OTE aids with open-canal fittings or deeply sealed canal aids) should be practical in the not-too-distant future. Etymotic Research is developing an integrated circuit version of a low-distortion, four-stage compression amplifier to help bring this about.

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REFERENCES

- Ahren, T. Arlinger, S., Holmgren, C., Jerlvall, L., Johansson, B., Lindblad, A. C., Persson, L., Pettersson, A., & Sjögren, G. (1977). *Automatic gain control and hearing aids* (Report TA No. 84). Stockholm: Karolinska Institutet, Tech. Audiol.
- Barfod, J. (1972). *Investigations on the optimum corrective frequency response for high tone hearing loss* (Report No. 4). Technical University of Denmark, Acoustics Laboratory.
- Barfod, J. (1976). *Multichannel compression hearing aids* (Report No. 11). Technical University of Denmark, Acoustics Laboratory.
- Barfod, J. (1978). Multichannel compression hearing aids: Experiments and consideration on clinical applicability. Sensorineural hearing impairment and hearing aids. C. Ludvigsen & J. Barfod (Eds.), *Scand. Audiol. (Suppl. 6)*, 315-340.
- Bauer, B. B. (1974, March). Audibility of phase distortion. *Wireless World*, 27-28.
- Berger, E. H., & Kerivan, J. E. (1983). Influence of physiological noise and the occlusion effect on the measurement of real ear attenuation at threshold. *J. Acoust. Soc. Am.*, 74, 81-94.

- Berland, O., & Nielsen, E. (1969, Summer). Sound pressure generated in the human external ear by a free sound field. *Audicibel*, 103-109.
- Berlin, C. I., & Gondra, M. I. (1976). Extratympanic clinical cochleography with clicks. In Rubin, Eberling & Salomon (Eds.), *Electrocochleography*. Baltimore: University Park Press.
- Blessner, B. A., & Ives, F. (1972). A reexamination of the S/N question for systems with time-varying gain or frequency response. *J. Audio Eng. Soc.*, 20, 638-641.
- Bucklein, R. (1962). Horbarkeit von Unregelmäßigkeiten in Frequenzgängen bei akustischer Übertragung. *Frequenz*, 16, 103-108.
- Burkhard, M. D., & Sachs, R. M. (1975). Anthropometric manikin for acoustic research. *J. Acoust. Soc. Am.*, 58, 214-222.
- Consumer Reports*. (1977). Low-priced loudspeakers. *Consumer Reports*, 42, 406-409.
- Cox, R. M. (1982). Combined effects of earmold vents and suboscillatory feedback on hearing aid frequency response. *Ear and Hearing*, 3, 12-17.
- Cox, R. M., & Gilmore, C. (1986). Damping the hearing aid frequency response: Effects on speech clarity and preferred listening level. *J. Speech Hear. Res.*, 29, 357-365.
- Dallos, P., Ryan, A., Harris, D., McGee, T., & Ozdamar, O. (1977). Cochlear frequency selectivity in the presence of hair cell damage. E. F. Evans & J. P. Wilson, (Eds.), *Psychophysics and physiology of hearing*. London: Academic Press.
- Dalsgaard, S. C. & Jensen, O. D. (1974). Measurement of insertion gain of hearing aids. *Eighth Int. Cong. on Acoust.*, 1, (p. 205). London.
- Davis, M. (1978, June). What's really important in loudspeaker performance? *High Fidelity*, 53-58.
- Dillon, H., & Macrae, J. (1984). *Derivation of design specifications for hearing aids* (NAL Report No. 102). Australia: National Acoustical Laboratories.
- Dunn, H. K., & Farnsworth, D. W. (1939). Exploration of pressure field around the human head during speech. *J. Acoust. Soc. Am.*, 10, 184-199.
- Dunn, H. D., & White, S. D. (1940). Statistical measurements on conversational speech. *J. Acoust. Soc. Am.*, 11, 278-288.
- Filler, A. S., Ross, D. A., & Wiener, F. M. (1945). *The pressure distribution in the auditory canal in a progressive sound field*. Harvard University Report PNR-5 (Contract N5ori-76) to Office of Research and Inventions, U. S. Navy.
- Flanagan, J. (1957). Difference limen for formant amplitude. *J. Speech Hear. Dis.*, 22, 205-212.
- Fletcher, H. (1931). Some physical characteristics of speech and music. *Bell Syst. Tech. J.*, 10, 349-373.
- Fletcher, H. (1942). Hearing, the determining factor for high-fidelity transmission. *Proc. Inst. Rad. Eng.*, 30, 266-277.
- Gengel, R. W. (1973). Temporal effects in frequency discrimination by hearing impaired listeners. *J. Acoust. Soc. Am.*, 54, 11-15.
- Goldberg, H. (1960). *A new concept in hearing aids*. (Dynauro, Inc. Report). Flushing, NY.
- Goldberg, H. (1965). Psychological adaptation to hearing correction. *Analog of MPLS Dealer Bulletin*.
- Goldberg, H. (1966). *Hearing aid*. U.S. Patent No. 3229049. Filed Aug. 4, 1960.
- Goldberg, H. (1972). The utopian hearing aid: Current state of the art. *J. Aud. Res.*, 12, 331-335.
- Hoth, D. F. (1941). Room noise spectra at subscriber's telephone locations. *J. Acoust. Soc. Am.*, 12, 499-504.
- Hotvet, D. (1988). Automatic gain control for hearing aid. U. S. Patent No. 4,718,099.
- How CU's auditory lab tests loudspeaker accuracy. (1977). *Consumer Reports TNG-3*. Mount Vernon, NY.
- Ingelstram, R., Johansson, B., Pettersson, A., & Sjögren, H. (1971). *The effect of non-linear amplitude distortion, an investigation by variation of the quadratic and the cubic components*, (Report TA No. 64). Stockholm: Karolinska Institutet, Tech. Audiology.
- Jerger, J. F., Tillman, T. W., & Peterson, J. L. (1960). Masking by octave bands of noise in normal and impaired ears. *J. Acoust. Soc. Am.*, 32, 385-390.
- Khanna, S. M., Tonndorf, J., & Queller, J. E. (1976). Mechanical parameters of hearing by bone conduction. *J. Acoust. Soc. Am.*, 60, 139-154.
- Killion, M. C. (1976). Noise of ears and microphones. *J. Acoust. Soc. Am.*, 59, 424-433.

- Killion, M. C., (1978). Revised estimate of minimum audible pressure: Where is the "missing 6 dB?" *J. Acoust. Soc. Am.*, 63, 1501-1508.
- Killion, M. C., (1979a). *Design and evaluation of high fidelity hearing aids*. Doctoral dissertation, Northwestern University. (University Microfilms, Ann Arbor).
- Killion, M. C., (1979b). *AGC circuit particularly for a hearing aid*. U.S. Patent No. 4,170, 720.
- Killion, M. C., Berger, E. H., & Nuss, R. A. (1987). Diffuse field response of the ear. *J. Acoust. Soc. Am.*, 81 (Suppl. 1), S75.
- Killion, M. C., & Monser, E. L. (1980). CORFIG: Coupler response for flat insertion gain. G. A. Studebaker & I. Hockberg, (Eds.), *Acoustical Factors affecting hearing and performance*. Baltimore: Park Press.
- Killion, M. C., & Studebaker, G. A. (1978). A-weighted equivalents of permissible ambient noise during audiometric testing. *J. Acoust. Soc. Am.*, 63, 1633-1635.
- Killion, M. C., Wilber, L. A., & Gudmundsen, G. I. (1985). Insert earphones for more interaural attenuation. *Hear. Instr.*, 36, 34-36.
- Killion, M. C., Wilber, L. A., & Gudmundsen, G. I. (1988). Zwislocki was right . . . a potential solution to the "hollow voice" problem. *Hear. Instr.*, 39(1), 14-17.
- Killion, M. C., & Wilson, D. L. (1985). Response-modifying earhooks for special fitting problems. *Audicibel*, 34, 28-30.
- Kruger, B., & Mazor, R. M. (1988). The occlusion effect: Supraaural and insert audiometric earphones. *Ear and Hearing*, (in review).
- Kuhn, G. F., & Burnett, E. D. (1977). Acoustic pressure field alongside a manikin's head with a view towards in situ hearing-aid tests. *J. Acoust. Soc. Am.*, 62, 416-423.
- Licklider, J. C. R. (1949). Basic correlates of the auditory stimulus. S. S. Stevens (Ed.), *Handbook of experimental psychology*. NJ: Wiley.
- Lindblad, A. C. (1982). *Detection of nonlinear distortion on speech signals by hearing impaired listeners* (Report TA105). Stockholm: Karolinska Institutet Technical Audiology.
- Lippman, R. P. (1978). *The effect of amplitude compression on the intelligibility of speech for persons with sensorineural hearing loss*. Doctoral dissertation, Mass. Inst. Tech. (University Microfilms, Ann Arbor).
- Lybarger, S. F. (1944). U.S. Patent Applications S.N. 543,278. July 3, 1944.
- Macrae, J., & Dillon, H. (1986). *Updated performance requirements for hearing aids* (NAL Report No. 109). Australia: National Acoustical Laboratories.
- Madaffari, P. L. (1974). Pressure variation about the ear. *J. Acoust. Soc. Am.*, 56, S3(A). (Available from Knowles Electronics, Franklin Park, IL).
- Marsh, R. C. (1975, August). Tweeters in the grass, alas. *Chicago*, 76-78.
- Martin, M. C. (1973). Hearing aid gain requirements in sensorineural hearing loss. *British J. Audiol.*, 7, 21-24.
- Mathes, R. C., & Wright, S. B. (1934). The compandor — an aid against static in radio telephony. *Bell Syst. Tech. J.*, 13, 315-322.
- McGee, T. (1978a). *Psychophysical tuning curves from hearing impaired listeners*. Doctoral dissertation, Northwestern University. (University Microfilms, Ann Arbor).
- Miller, J. D. (1974). Effects of noise on people. *J. Acoust. Soc. Am.*, 56, 729-764.
- Milner, P. (1977, June). How much distortion can you hear? *Stereo Rev.*, 64-68.
- Moir, J. (1958). *High quality sound reproduction*. (p. 51). New York: Macmillan.
- Mott, E. E. (1944). Idical response of telephone receivers. *Bell System Tech. Jour.*, 23, 135-150.
- Muraoka, T., Iwahara, M., & Yamada, Y. (1981). Examination of audio-bandwidth requirements for optimum sound signal transmission. *J. Audio Eng. Soc.*, 29, 2-9.
- Olson, H. F. (1957). *Acoustic engineering*. New York: Van Nostrand.
- Olson, H. F. (1967) *Music, physics, and engineering*. New York: Dover.
- Palva, T., Goodman, A., & Hirsh, I. J. (1953). Critical evaluation of noise audiometry. *Laryngoscope*, 63, 842-860.
- Perkins, R., & Goode, R. (1987, September). Surgical modification of the external auditory canal for an invisible aid. Paper given at the Annual Meeting of the American Academy of Otolaryngology-Head and Neck Surgery, Chicago, IL.
- Peters, R. W., & Burkhard, M. D. (1968). *On noise distortion and harmonic distortion measurements*. (Industrial Res. Prod. Report No. 10350-1) Knowles Electronics, Franklin Park, IL.
- Punch, J. D. (1978). Quality judgments of hearing aid-processed speech and music by normal and otopathologic listeners. *J. Amer.*

- Audio. Soc.*, 3, 179-188.
- Rood, G. M. (1986). An active noise reduction system for flying helmets and ear protectors. *J. Acoust. Soc. Am.*, 79(Suppl. 1), S18(A).
- Sachs, R. M., & Burkhard, M. D. (1972). Ear-phone pressure response in ears and couplers. *J. Acoust. Soc. Am.*, 51, 140(A). (Available as Industrial Research Products Report No. 20021-2 to Knowles Electronics).
- Scharf, B. (1978). Comparison of normal and impaired hearing. I. Loudness, localization: II. Frequency analysis, speech perception. C. Ludvigsen & J. Barfod (Eds.), *Sensorineural hearing impairment and hearing aids. Scand. Audiol.* (Suppl. 6), 4-106.
- Scharf, B., & Hellman, R. P. (1966). A model of loudness summation applied to impaired ears. *J. Acoust. Soc. Am.*, 40, 71-78.
- Schulein, R. B. (1975). In situ measurement and equalization of sound reproduction systems. *J. Audio Eng. Soc.*, 23, 178-186.
- Schweitzer, H. D., & Causey, G. D. (1977). The relative importance of recovery time in compression hearing aids. *Audiology*, 16, 61-72.
- Seacord, D. F. (1940). Room noise at subscriber's telephone locations. *J. Acoust. Soc. Am.*, 12, 183-187.
- Shaw, E. A. G. (1974). The external ear. W. D. Keidel & W. D. Neff (Eds.), *Handbook of sensory physiology* (Vol. VII, Chapter 14). Berlin: Springer-Verlag.
- Shaw, E. A. G. (1976). Diffuse field sensitivity of the external ear based on the reciprocity principle. *J. Acoust. Soc. Am.*, 60, S102(A).
- Singer, G. A. (1950, November 18-19). New limiting amplifiers. *Audio Eng.*, 69-70.
- Skinner, M. W. (1976). *Speech intelligibility in noise-induced hearing loss: Effects of high-frequency compensation*. Doctoral dissertation, Washington University. (University Microfilms, Ann-Arbor).
- Skinner, M. W. (1980). Speech intelligibility in noise-induced hearing loss: Effects of high-frequency compensation. *J. Acoust. Soc. Am.*, 67, 306-317.
- Smirga, D. (1986, July). Modern compression technology, developments & applications Part II. *Hearing J.*, 13-16.
- Snow, W. B. (1931). Audible frequency ranges of music, speech, and noise, *Bell Sys. Tech. J.*, 10, 616-627.
- Steinberg, J. C., & Gardner, M. B. (1937). The dependence of hearing impairment on sound intensity. *J. Acoust. Soc. Am.*, 9, 11-23.
- Stevens, S. S. (1972). Perceived level of noise by Mark VII and decibels (E). *J. Acoust. Soc. Am.*, 51, 575-601.
- Termin, F. E., & Pettit, J. M. (1952). *Electronic measurements* (p. 341). NY: McGraw-Hill.
- Tillman, T. W. (1962). *A critical view of the SAL test*. (Tech. Docum. Rep. No. SAM-TDR-62-96). School of Aerospace Medicine, Aerospace Medical Div. (AFSC), U.S. Air Force. TX: Brooks Air Force Base.
- Toole, F. E. (1981). Listening tests — turning opinion into fact. *J. Audio Eng. Soc.*, 30, 431-445.
- Toole, F. E. (1984). Subjective measurements of loudspeaker sound quality and listener performance. *J. Audio Eng. Soc.*, 33, 2-32.
- Toole, F. E. (1986). Loudspeaker measurements and their relationship to listener preferences: Part 2. *J. Audio Eng. Soc.*, 34, 323-348.
- Villchur, E. (1973). Signal processing to improve speech intelligibility in perceptual deafness. *J. Acoust. Soc. Am.*, 53, 1646-1657.
- Villchur, E. (1978). A critical survey of research on amplitude compression. C. Ludvigsen & J. Barfod (Eds.), *Sensorineural hearing impairment and hearing aids. Scand. Audiol.*, (Suppl. 6), 305-314.
- Voroba, B. (1987). Patient-selected soft canal hearing instruments. *Hear. Instr.*, 38(4), 39-40.
- Wallach, H. (1940). The role of head movements and vestibular and visual cues in sound localization. *J. Exp. Psychol.*, 27, 339-368.
- Wang, C. Y., & Dallos, P. (1972). Latency of whole nerve action potentials: Influence of hair cell normalcy. *J. Acoust. Soc. Am.*, 52, 1678-1686.
- Watson, N. A., & Gales, R. S. (1943). Bone-conduction threshold measurements: Effects of occlusion, enclosures, and masking devices. *J. Acoust. Soc. Am.*, 14, 207-215.
- Wegel, R. L., & Lane, C. E. (1924). The auditory masking of one pure tone by another and its probable relation to the dynamics of the inner ear. *Rev.*, 23, 266-285.
- Wiener, F. M., & Ross, D. A. (1946). The pressure distribution in the auditory canal in a progressive sound field. *J. Acoust. Soc. Am.*, 18, 401-408.
- Zwislocki, J. J. (1970). *An acoustic coupler for ear-phone calibration* (Report LSC-S-7). Syracuse, NY: Syracuse University, Lab. of Sensory Comm.