CHAPTER 3

TRANSDUCERS AND ACOUSTIC COUPLINGS

THE HEARING AID PROBLEM THAT IS (MOSTLY) SOLVED

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This chapter proceeds from a summary of the state of the art in hearing aid transducers directly to a discussion of many of the traditional problems involving transducers. Problems of size, response of smoothness, bandwidth limitations, and noise level have now been completely solved.

TRANSDUCERS AND COUPLINGS

Transducer Miniaturization

Figure 3.1 shows a series of hearing aid transducers of decreasing size. The largest (3 cc) is the Knowles AJ-series microphone introduced in 1954; the smallest (.03 cc) is the EM-series microphone introduced in 1988. At each stage in this miniaturization process, it has been possible to make receivers (the hearing-aid and telephone-industry jargon for earphone) and microphones of similar size. This size reduction has permitted hearing aids to progress from body worn aids (the writer's high school chemistry teacher used a body aid with an AJ disguised as a treble clef tie tack), to headworn aids (eyeglass and behind-the-ear), to in-the-ear (ITE) aids, to in-the-canal (ITC) aids.

In addition to the obvious cosmetic advantages, this same size reduction has brought significant acoustic benefits for the hearing aid wearer. With head-worn hearing aids, head diffraction and head motion cues become available to the extraordinarily powerful binaural signal processing computer located in the mammalian brainstem, providing several dB of direct "binaural squelch" for interfering noise and reverberation, as well as a much more pleasant sense of auditory space. With ITE hearing aids, some of the directional properties of the pinna are preserved in the hearing aid output; with the smallest canal aids, which pick up sound directly in front of the blocked ear canal leaving the concha unfilled, essentially *all* of the directional properties of the external ear are preserved in the hearing aid output. The source-direction-dependent spectral "signature" impressed on sounds by the head, pinna, and concha, in combination with the head-motion information from the vestibular system, permits some individuals with

Note: The microphone and receiver model designations used throughout this chapter are those of Knowles Electronics, where the writer studied transducer design under Elmer Carlson from 1962 until 1983.

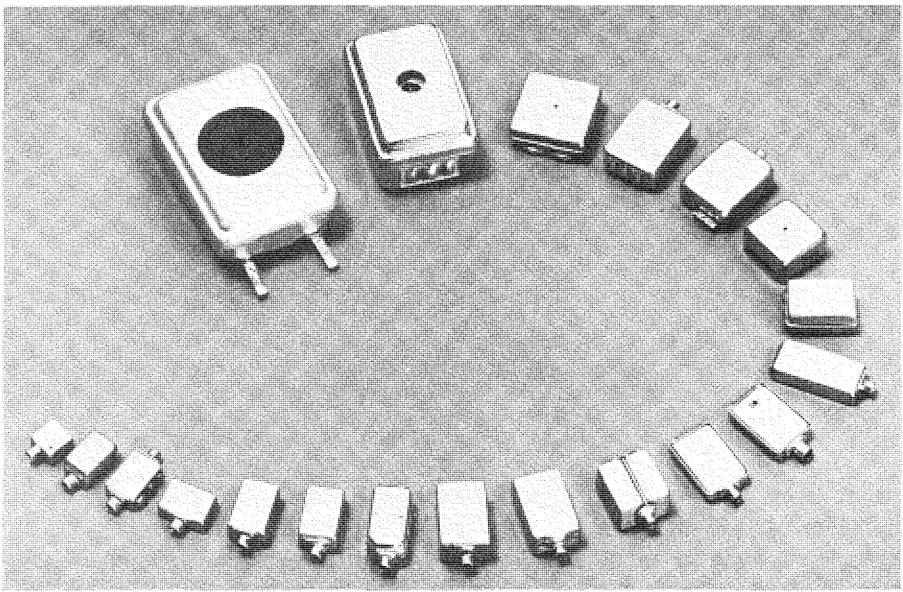

FIGURE 3.1. Hearing aid microphones and receivers from 1954 to 1989. (Courtesy Knowles Electronics.)

only one working ear to experience a surprisingly good sense of auditory space and ability to localize sounds. The auditory system appears to have been designed to use a rich set of cues, cues that can be preserved in their entirety only with sufficiently small transducers.

Transducer Types

The only energy-efficient receiver designs produced to date are magnetic, typically using a push-pull type of "balanced armature" magnetic construction such as illustrated in Figure 3.2 (Carlson, Cross, and Killion, 1971). Other transducer mechanisms have been proposed, but to the writer's knowledge none come close to providing the same electromechanical coupling efficiency; the next nearest competitor appears to be 10 dB less efficient in converting electrical energy to acoustic output. The dominance of the magnetic receiver is thus readily understood. The hearing aid wearer must carry his power source around with him, and the receiver consumes 50 percent to 95 percent of the power in a well-designed hearing aid. Thus, receiver efficiency determines almost directly

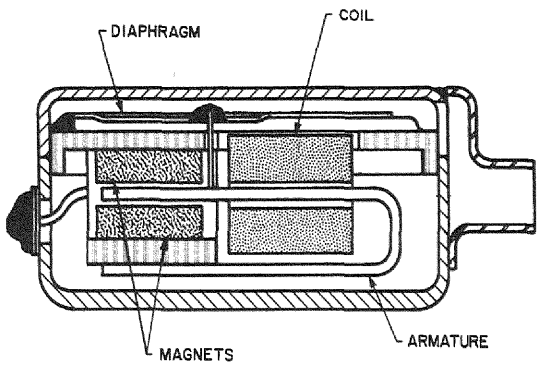

FIGURE 3.2. Balanced armature magnetic receiver (BK-series).

how often the battery must be changed: A 10 dB reduction in efficiency would require purchasing 5 to 9 times as many batteries each year.

The first head-worn hearing aid microphones were also magnetic, because a good *energy* efficiency was required to provide adequate sensitivity when the microphone was connected to the relatively low input impedance of the "bipolar" transistor hearing aid amplifiers that were common in hearing aids. The few picoFarads capacitance of a typical electret-condenser microphone would have been effectively shorted out by a low input impedance. Once low-noise Junction Field Effect Transistor (JFET) preamplifiers became available, however, *energy* efficiency in the microphone ceased to be important, and the lower-efficiency but higher *sensitivity* piezoelectric ceramic and electret microphone elements became practical because the JFET could be driven by a low-current, low-energy signal. Figure 3.3 illustrates the dramatically different rank ordering that results when magnetic, ceramic, and electret microphones are compared using a low-input-impedance amplifier (Figure 3.3a) and a high-input-impedance amplifier (Figure 3.3b).

Figure 3.4 shows the construction of the BT-series electret microphone the writer helped design in the early 1970s (Killion and Carlson, 1974). Note that the JFET preamplifier is shielded by the metal case of the microphone, without which the hum and electrical noise pickup at the high-impedance input would be unacceptable. (The input resistor typically used in these preamplifiers is in the 1000-megohms range.) In a sense, the introduction of the JFET preamplifier in a size small enough to fit inside the microphone case brought us back full circle to the days of the high-input-impedance subminiature-vacuum-tube hearing aid amplifier, when piezoelectric microphones were common. In today's hearing aid designs, where the use of monolithic-integrated-circuit amplifiers is common, separating the preamplifier function from the amplifier function means that the integrated circuit can be readily optimized independently of the special requirements of the JFET preamplifier.

Performance Versus Size

At one time it was popular to excuse poor hearing aid sound quality by pointing to the "inevitable degradation in performance with decreasing size." That was plausible armchair reasoning, but just the opposite has happened in practice. The bandwidth and response smoothness of both microphones and earphones have steadily improved with their decreasing size, as has nearly every other property: resistance to shock damage, magnetic shielding, and microphone insensitivity to vibration. Figure 3.5 shows the improved (reduced) vibration sensitivity of microphones with succeeding generations of designs.

The acoustic gains routinely obtained in modern ITE and canal hearing aids—with the microphone and receiver almost touching each other—would have been out of the question with such close spacing even 15 years ago: If the magnetic coupling didn't cause feedback, the 40 to 60 dB greater vibration sensitivity of the microphone would have. Similarly, the minimal shock mounting routinely used in modern ITE and canal hearing aids would have been impractical 15 years ago.

It is perhaps not surprising that the bandwidth of microphones and receivers has generally improved with smaller size. Miniature magnetic receivers have a real-ear frequency response that is intrinsically flat from very low frequencies (limited only by the quality of the earmold seal and the choice of barometric-release vent inside the receiver) up to the frequency where the combined acoustic-mechanical masses and compliances resonate. Smaller mechanical devices have a natural ten-

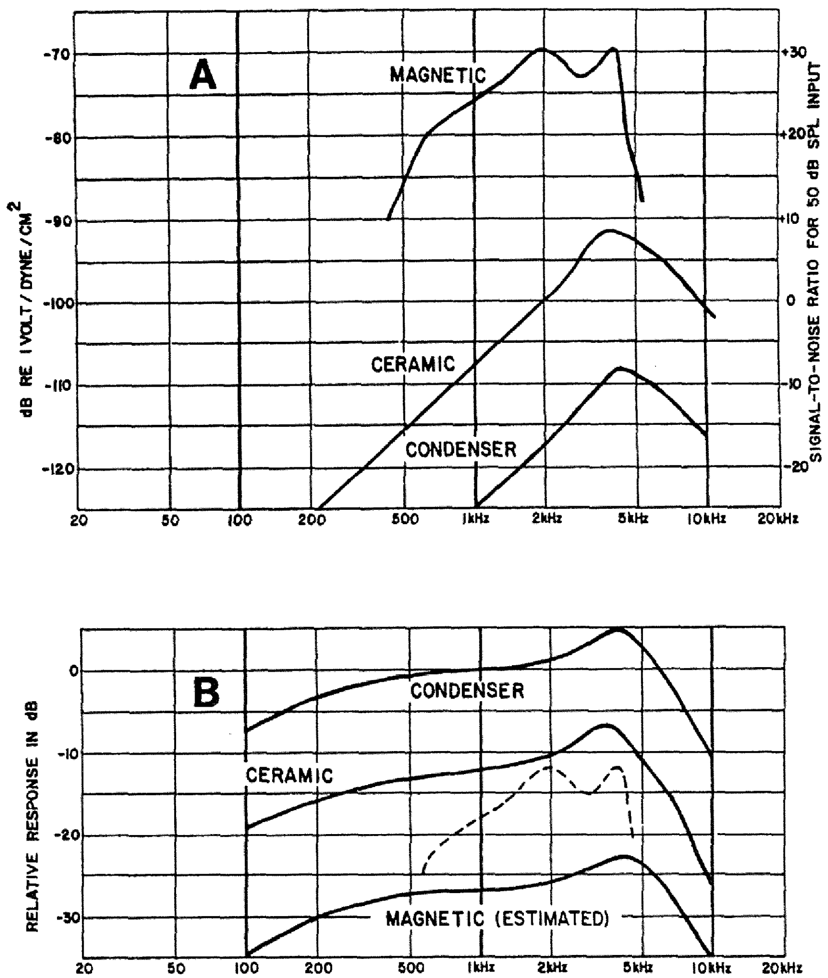

FIGURE 3.3. Rank ordering of the sensitivity of magnetic, piezoelectric ceramic, and electret condenser microphones operating into: (a) Low input impedance (5000 ohm) amplifier, and (b) High input impedance (100 megohms) amplifier.

dency toward higher resonance frequencies, so it is natural that the receiver bandwidth should increase with smaller size. The problem in the smallest of the recent designs has been to keep the principal resonance frequency *down* to approximately 2800 Hz, the resonance frequency of the average external ear.

What *is* surprising is that the signal-to-noise level of the microphone has not diminished markedly along with its size. Even the sensitivity of the receivers has held up *better than one might expect.*

Microphone Noise. The signal-to-noise level of any microphone that is small compared to a wavelength should be proportional to the

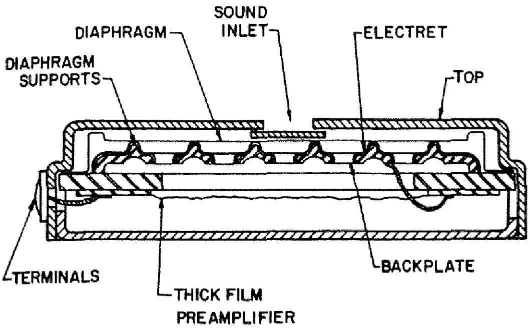

FIGURE 3.4. Electret-condenser microphone with built-in preamplifier (BT-series). (Reprinted with permission from Killion & Carlson, 1974.)

square root of the case volume, as can be seen by considering the case where two microphones are T-connected to the same sound inlet tube and their electrical outputs connected in series. The resulting *signal* sensitivity will be 6 dB greater, because the two microphones' signal voltage outputs are equal and coherent (in phase) and thus add linearly. Their combined *noise* output will be only 3 dB greater, however, because their individual noise voltage outputs will be generally completely uncorrelated (random) and thus add in a root-mean-square fashion.

Following this reasoning, the EK-series microphone might be expected to be 9 dB noisier than the BA-series microphone that occupies 8 times the volume. Because of steady improvements in materials and design, however, the EK-series microphone is just as quiet as the BA-series. The equivalent SPL of the noise in a 1 Hz bandwidth at 1 kHz is about –14 dB for both microphones. Neither microphone produces noise that is any limitation to hearing, as discussed in the section on noise.

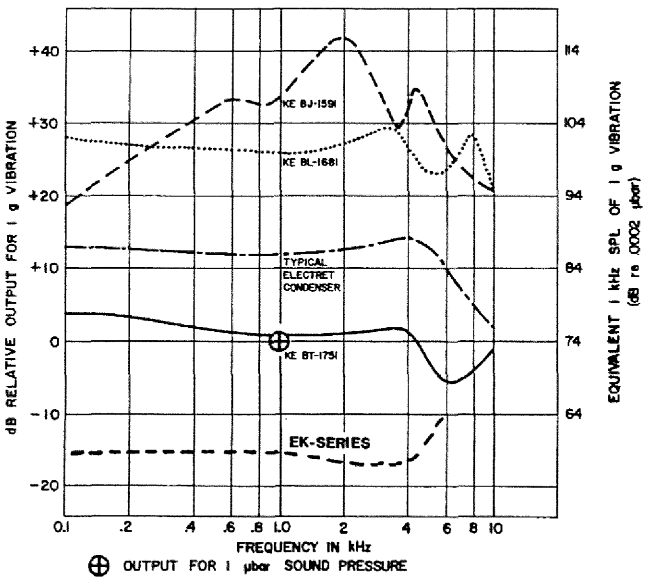

FIGURE 3.5. Vibration sensitivity of magnetic (BJ-series), piezoelectric ceramic (BL-series), and three generations of electret condenser microphones (CA-series, BT-series, and EK-series).

Receiver Sensitivity. A similar theoretical expectation holds for hearing aid receivers, whose sensitivity and maximum undistorted output should be proportional to the square root of the case volume. Again this is easily seen by simple reasoning once the following observation is made: The important load on a subminiature receiver diaphragm is not the 0.6 cc or so of air in the occluded ear canal (or the total 1.3 cc equivalent volume when the compliance of the eardrum is included), but the much smaller volume of air trapped *behind* the diaphragm. This amounts to less than .06 cc in the ED-series receiver; therefore, producing 110 dB SPL in the ear canal at low frequencies requires the production of 137 dB SPL in the "back volume" behind the receiver diaphragm.

Thus, if the outputs of two equal receivers are combined into a single sound tube, while their electrical terminals are connected in series across a low-impedance amplifier (constant voltage source), the output SPL in the ear will remain the same but the amplifier output current will drop in half. This can be seen as follows: With the two receiver coils in series, their combined impedance Z will be doubled and, by Ohm's law ($i = e/Z$), each will experience half the former current. Because each receiver's diaphragm motion is determined almost completely by the pressure behind the diaphragm and thus is unaffected by the presence of the other receiver, each receiver will now produce half the volume displacement. The *total* volume displacement of air, and consequently the SPL developed in the ear canal, will remain constant even though the pair of receivers uses one-half (3 dB less) the power. Clearly the pair of receivers (which occupy twice the volume of a single receiver) will produce 3 dB more output on the *same* power.

Despite this theoretical expectation, the EC-series receiver (.12 cc) has a sensitivity only 1 dB less than that of the BK-series receiver (.18 cc) that occupies 1.5 times the volume. Here again, steady improvements in materials and design have partially held off the presumed automatic loss of performance with decreasing size.

It is interesting to consider, however, what would happen if the improvements in materials and design were applied to the redesign of an older, larger receiver. Stuart Ewens did just that with the CI-series receiver, which occupies the same case volume as the older BI-series receiver (.29 cc). The result is a receiver that requires only half the power to generate the same SPL, yet will handle *twice* the maximum power input without overload, for a net increase of 6 dB in maximum undistorted output. The CI is capable of producing greater than 130 dB SPL at the eardrum without significant distortion.

Even with improvements in receiver efficiency, it is still true that the smallest receivers require the greatest battery power to produce a given output SPL. Until recently, the smallest canal aids presented a seemingly unsolvable dilemma: They needed the smallest (lowest-sensitivity) receiver, which required the highest battery drain, *and* the smallest (lowest mA-hr capacity) battery. In order to provide the one-week battery life that is generally considered the minimum acceptable, the manufacturer was limited to approximately .3 mA receiver bias before battery life became unacceptably short. Worse, space often did not permit anything but a "class A" (constant-power-drain) amplifier. The result was a limit on undistorted receiver output, especially at high frequencies, that left it marginal at best (see the section on distortion).

Class D Receivers. Just as the availability of JFET preamplifiers allowed a breakthrough in microphone design, the availability of low-voltage silicon gate CMOS circuitry made it possible to include a subminiature "class D" (high-efficiency switching) drive amplifier inside the receiver case (Killion, 1986, 1987; Carlson, 1988). Not only is the idling current low in the new class D amplified receivers, but their battery consumption increases much more slowly with increasing output than class B amplifiers. At an output of 100 dB SPL, for example, the battery drain of the EP-3074 class D receiver has increased only .02 mA over its idling value of .15 mA. In cooperation with appropriate amplifier design, the result can be a canal aid that averages only .32 mA of battery drain, yet has an undistorted output capability of 110 to 115 dB SPL. Since the instantaneous peaks in speech and music are typically 10 to 15 dB above the rms level, such a hearing aid *operating below clipping* will, using a class D receiver, be operating near idle most of the time. One of life's excitements for the writer, who was group leader of the team that designed the class D driver used in these receivers, was to wear a pair of ITE hearing aids for 26 and 28 days before their batteries needed changing: almost exactly the calculated battery life for their *idling* current drain.

With the CMOS class D driver built into the receiver case, it becomes a "zero space consuming" output amplifier as far as the hearing aid designer is concerned, permitting its use even in canal aids. In addition, with both the JFET microphone preamplifier and the CMOS receiver driver taken care of, their special processing requirements will not burden the standard bipolar transistor processing typically used for the main hearing aid amplifier. This is fortunate, since the bipolar transistor integrated circuit still appears far superior to other technologies wherever low noise levels and sophisticated signal processing must be accomplished in minimum size and with minimum battery drain. (A CMOS input device, comparable in noise performance to that routinely achieved in only .02 mm^2 of bipolar circuit area, would exceed the size of some complete hearing aid amplifier chips. For high-speed digital signal processing, on the other hand, CMOS devices hold enormous promise for low-voltage, low-current operation. The writer and colleagues have produced analog-plus-digital CMOS integrated circuits with all portions operating properly on less than 0.1 volts dc supply.)

Receiver Vibration and Feedback. There is one important characteristic of subminiature receivers that *has* degraded with decreasing size. The smaller receivers produce greater mechanical vibration for the same acoustic SPL delivered to the ear canal. This is the principal limitation to the maximum available acoustic gain before feedback (the term feedback is used somewhat imprecisely in this chapter to mean "the hearing aid whistles"), but it comes about indirectly. The *direct* mechanical coupling to the microphone produces little feedback problem, because the mechanical vibration sensitivity of most recent electret microphone designs (the EK-series and EM-series "AVM" microphones, for example) is so low it is nearly impossible to measure. (This was almost true even on previous generations of microphones whose vibration sensitivity was 20 dB greater. See Killion, 1975.) These microphones can be safely cemented directly to the hearing aid housing with either no increase in feedback problems or, in the case where previously a rubber tube allowed the microphone to "pump" on the tube, a decrease in feedback problems.

In the writer's experience, the feedback problem can normally be traced to the SPL developed at the microphone inlet due either to a slit leak along the ear canal exiting near the microphone inlet or, more commonly, the SPL developed by the entire hearing aid case acting like a miniature speaker diaphragm in response to mechanical vibrations from the receiver. One of the strongest bits of evidence that the latter is the most important problem derives from the 20 dB or so increase in maximum gain before feedback that Harada (1989) routinely obtained from two receivers mounted "belly to belly" so their mechanical vibrations canceled but their acoustic outputs added. This arrangement is now commercially available as the Knowles EJ-series of dual receivers. Deeply sealed eartips appear to be another possible solution: The greater mechanical shear impedance in or near the bony part of the ear canal may be sufficient to partially restrain the shell from vibrating. (Used diagnostically, a finger lightly touching the faceplate of an ITE will often stop the oscillation well before enough pressure is applied to improve the earmold seal.) For the moment, the manufacturer of ITE and canal aids must apply his finely tuned art of compliant mounting and careful orientation of the receiver in order to provide high-gain instruments.

THE TRADITIONAL PROBLEM AREAS

Bandwidth

At one time hearing aid transducers were intentionally designed with a bandwidth restricted to the region between 500 and 3000 or 4000 Hz (Killion and Carlson, 1970). The common belief at that time was that such a restriction would allow nearly all of the important speech information to pass while rejecting noise in those frequency regions where the residual speech cues were less important. This reasoning appeared to be supported by data from the articulation-index experiments performed at Bell Laboratories (French and Steinberg, 1947), which showed that the frequency bands below 400 Hz contained only 5 percent of the speech information and those bands above 4000 Hz contained only 15 percent of the speech information. Similar and equally plausible reasoning more recently has led to "noise blocker" and "ASP" hearing aid circuits that, in an attempt to improve the user's ability to understand speech in noise, reduce the response of the hearing aid in frequency regions presumed to contain predominately noise.

Most recent experiments, however, have indicated that (a) filtering out low- and high-frequency noise simultaneously filters out low- and high-frequency speech cues, and (b) it is precisely in the presence of competing speech and noise that the residual speech cues in these "less important" frequency regions become most important. Most individuals, including those with hearing impairment, have little trouble understanding sufficiently intense speech in quiet. Indeed, even the old-fashioned telephone with its 500 to 3000 Hz passband and its 30 percent total harmonic distortion typically yielded nearly perfect sentence intelligibility unless the talker was in a noisy room. The problem comes in the presence of noise, where a 10 percentage-point increase in articulation index may result in a 30 to 50 percentage-point increase in sentence intelligibility. As Villchur (1989) has observed, much research and development effort has focused on the wrong target—noise reduction—instead of the problem itself—improved clarity of speech in noise. Virtually all noise reduction schemes to date have thrown out the vital speech-clarification cues along with the noise. The resulting sound is indeed less noisy, but also less clear.

Fortunately, the transducers themselves have not been a limiting factor in recent years. The available bandwidth from hearing aid microphones and receivers has been, from the point of view of hearing, unlimited (i.e., equal to the roughly 20 Hz to 16 kHz of the ear) for nearly 15 years. The flat-frequency-response BT-1759 version of the Knowles BT-series electret hearing aid microphone has been regularly used in broadcast and recording studios since the early 1970s, and a standard-response Knowles ED-series receiver is used in what is arguably one of the most accurate high-fidelity earphones available (the 16 kHz-bandwidth Etymotic Research ER-1 insert earphone).

More direct proof that hearing aid transducers were not a limiting factor in hearing aid design was obtained in a series of simulated live versus recorded fidelity rating experiments conducted at Northwestern University by the writer in the late 1970s (Killion and Tillman, 1982). Figure 3.6 shows the real-ear frequency response of a variety of "high fidelity" sound systems evaluated by three groups of listening-test juries, one of which was made up of "golden ear" listeners such as Julian Hirsch, the regular stereo system reviewer for *Stereo Review*. All three juries gave roughly similar fidelity rating scores to the various sound systems. The major difference among the three groups was in the reliability of their ratings: On the average, it took 8 times as many "man-on-the-street" subjects to give the same reliability as could be obtained from a single golden-ear subject. The grand-average fidelity rating, averaged across juries and the four program selections (live voice, symphony orchestra, piano trio, and speech-spectrum noise), is shown beside the frequency response of each system. Note that the experimental over-the-ear, ITE, and canal-aid designs (labeled OTE, ITE, and ITC, respectively) were judged to have fidelity comparable to that of the ElectroVoice Sentry V stereo loudspeaker system (MS), the most popular recording-studio monitor system in the Chicago area in the 1970s. The experimental hearing aids were rated much higher in fidelity than the KOSS Pro 4AA headphones (PP), reputed to be the most popular stereo headphones ever produced, a speech audiometer (SA), airline headphones (AP), or a low-distortion but otherwise conventionally undamped BTE hearing aid (OTE-40).

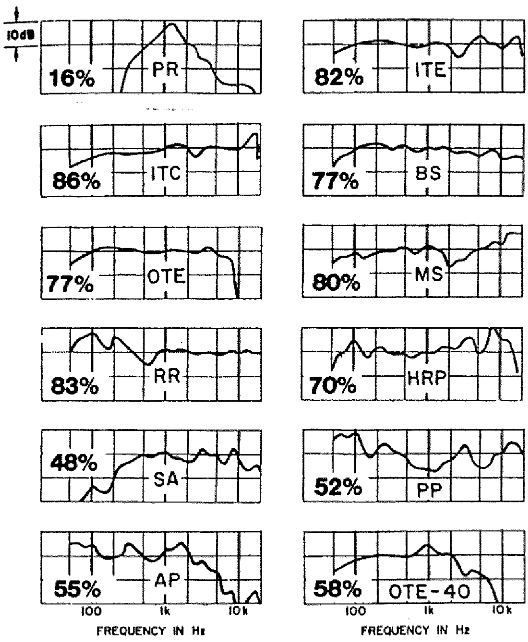

FIGURE 3.6. Frequency responses of sound systems used in listening tests, with their average fidelity rating shown in percent: $4.95 pocket radio (PR), bookshelf speakers (BS), monitor speakers (MS), relocated reference loudspeakers (RR), Senheisser HD-414 earphones (HRP), Speech audiometer (SA), Koss Pro-4AA earphones (PP), airline earphones (AP). ITE, ITC, OTE, and OTE-40 are hearing aids discussed in the text.

Frequency Response Shaping

Most recent experiments have also confirmed that a real-ear frequency response that has been tailored to the hearing loss provides improved speech intelligibility—especially in noise—for the hearing aid user. One of the most versatile hearing aids in that regard permits individual adjustment of the gain in each of thirteen frequency bands, using a digitally-programmed switched-capacitor-filter integrated circuit, so that the hearing aid may be tailored on the spot to the individual client.

The more traditional approach is for the manufacturer to choose from among the wide variety of available microphone and receiver frequency responses, combine those with frequency-response shaping in the hearing aid amplifier itself, and deliver the finished product to the dispenser who, in recent years, would test the insertion response of the delivered aid on the intended user. In the case of behind-the-ear aids, the dispenser might modify the earmold to obtain additional response control.

Microphone Responses. A sampling of the available electret microphone responses is shown in Figure 3.7. (These are plotted to the standard 30 dB/decade engineering scale rather than the standard 50 dB/decade hearing aid scale, a transducer tradition that goes back nearly as far as the use of the term "re-

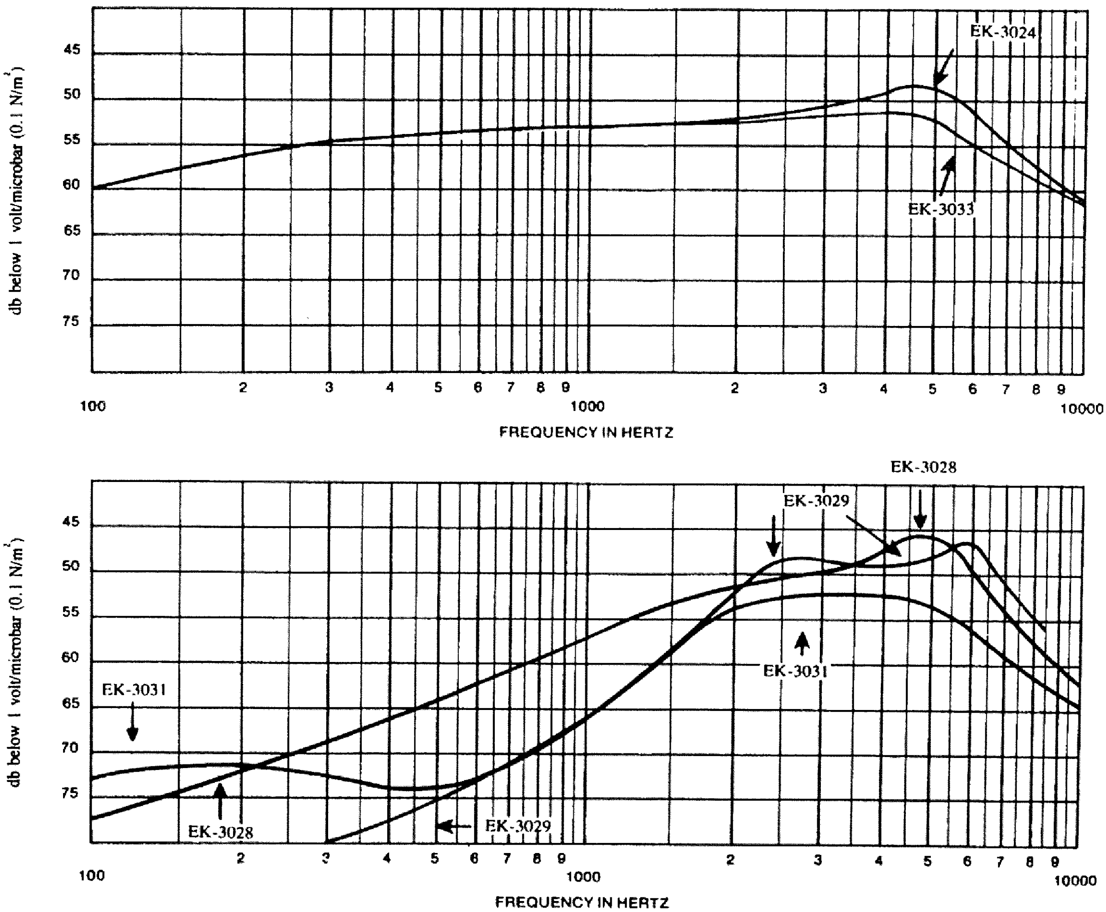

FIGURE 3.7. Variety of frequency response curves available in EK-series microphones. (Courtesy Knowles Electronics.)

ceiver.") A range of 0 to 20 dB high-frequency boost between .5 and 2 kHz can be obtained by choice of microphone. Of special interest is the EK-3031 microphone, which provides a smooth "stepped response" that would require several additional components to achieve by electronic means.

Directional Microphones. In addition to the standard omnidirectional electret microphones, directional-microphone capsules with a variety of internal rear-port time delays have been made available to make it possible for the hearing aid manufacturer to produce different directional characteristics with a given port spacing or to accommodate different port spacings. The properly utilized directional microphone provides an improvement of 3 to 5 dB in signal-to-noise ratio in difficult listening situations (Hawkins and Yacullo, 1984) and can provide an even greater benefit in some special situations.

A wide variety of performance has been achieved in directional-microphone hearing aids when measured *in situ*, ranging from just-barely-directional to good directional performance. The probable reason is that the location of the microphone has a marked effect on the effective port spacing: An over-the-ear location increases the effective spacing by about 1.4 times the physical spacing, whereas an ITE location decreases the effective spacing to about .7 times the physical spacing. Figure 3.8, from Madaffari (1983), illustrates different directional characteristics obtained with various combinations of effective port spacing and time delay. With careful attention to time delays, inlet-and-outlet phase shifts, and case and head diffraction, it is possible to make directional-microphone hearing aids whose *in-situ* directivity is good from low frequencies up through 4 or 5 kHz. Figure 3.9 illustrates such a design from the same Madaffari report. Even without careful design, it is possible to obtain good directivity up to 1 or 2 kHz, provided the microphone is chosen with time delay appropriate for the effective port spacing below 2 kHz.

Receiver-Plus-Earmold Response

Attempting to define the response of a hearing aid receiver without specifying the earmold is a nearly futile task. The earmold *controls* the delivered response of the combination, as illustrated in Figure 3.10. The curves in that figure were for a *single* receiver run with various "horn" and "reverse horn" earmold configurations. Only one of the curves represents a vented earmold. The topic of earmold acoustics is beyond the scope of this chapter, but several excellent references are available (Cox, 1979; Lybarger, 1980, 1985; Killion, 1984, 1988). Special earhooks that can provide low-frequency, high-frequency, or bandpass rejection are now also available for BTE hearing aids (Killion and Wilson, 1985).

Amplifier Source Impedance

A final response control measure available to the hearing aid designer should be mentioned. Figure 3.11 shows the change in earphone frequency response caused by changes in the source impedance (i.e., changes in the output impedance of the hearing aid amplifier). Since the receiver impedance rises with frequency, changing from a low-impedance (constant voltage) source to a high-impedance (constant current) source increases the high-frequency response. Resonating the coil with a shunt capacitor further boosts the high-frequency response under high-impedance-drive conditions. Although only one capacitor-value curve is shown in Figure 3.11, some hearing aids include a capacitor in series with an adjustable trimmer potentiometer to allow adjustment of the resonance peak frequency. As illustrated in Figure 3.11, the boost in response below resonance is accom-

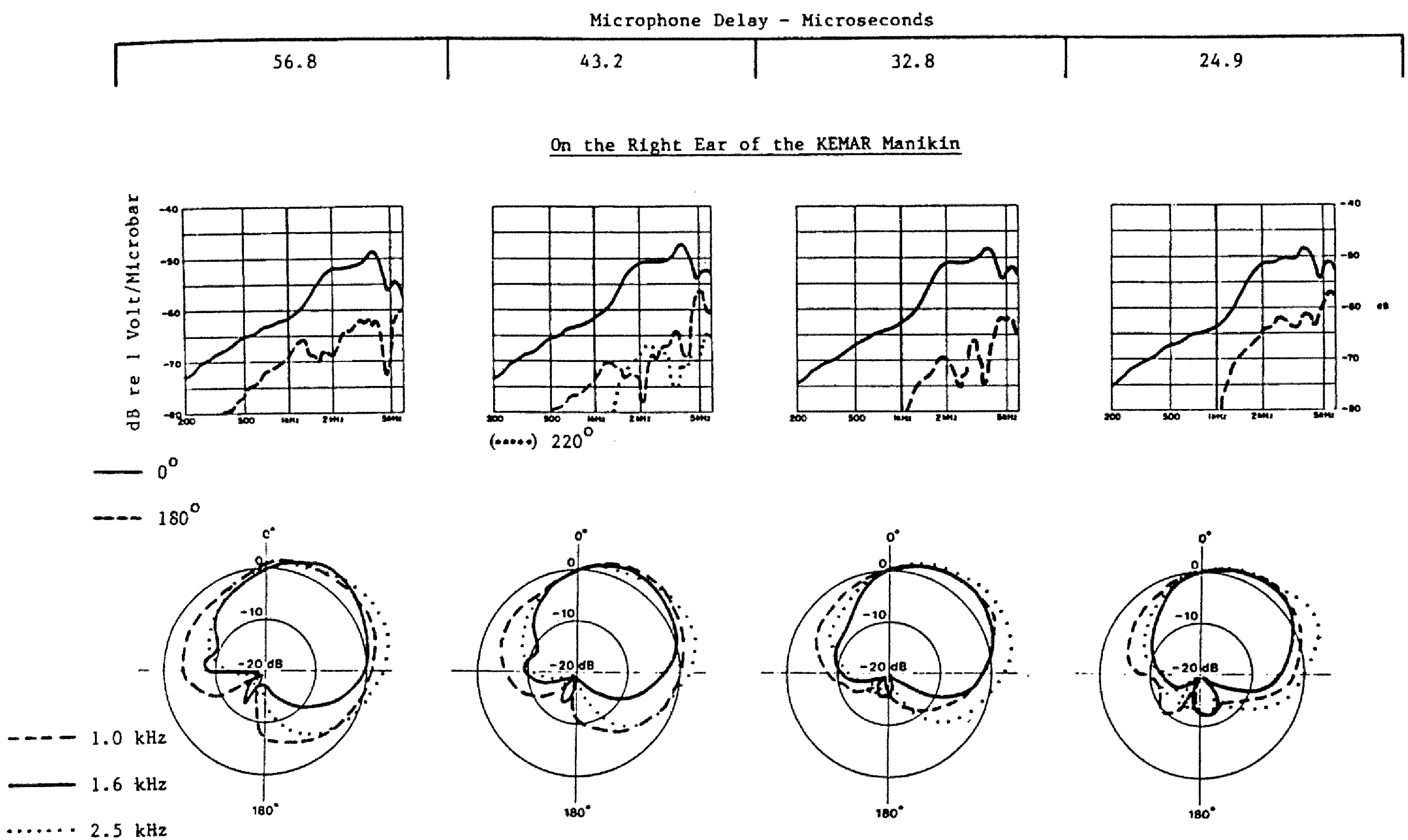

FIGURE 3.8. In-situ (KEMAR) directional microphone polar frequency and polar responses for 8.3 mm effective port spacing and various microphone time delays. (Reprinted with permission from Madaffari, 1983.)

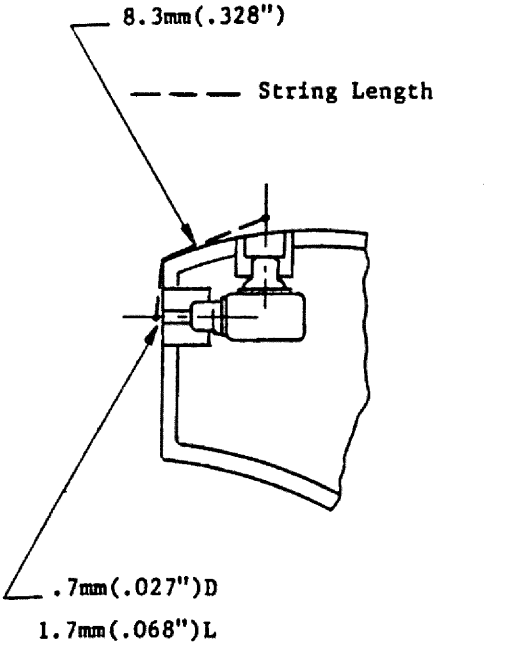

FIGURE 3.9. Diffraction-optimized directional-microphone hearing aid design giving good in-situ directional response over wide frequency band with 8.3 mm (.328") effective port spacing. (Reprinted with permission from Madaffari, 1983.)

panied by a reduced response above resonance, so such trimmers can also be used to roll off the highs as an on-the-spot high-frequency feedback fix. The hearing aid user ends up with less high-frequency emphasis than would probably be desirable when this happens, but it does save having to return the aid for a better shell fit or receiver mounting. Without this assurance, many manufacturers would probably be reluctant to make as good a high-frequency response available, knowing the number of returns they would experience (many of them due to inadequate ear impressions, which are out of their control).

The three curves in Figure 3.11 have been normalized to the same output level at 100 Hz to illustrate the change in frequency-response curve shape. The *maximum* undistorted output levels obtainable with a fixed-supply voltage are determined entirely by the receiver impedance (voltage clipping limits) and the current capability of the amplifier (current clipping limits) and have virtually *nothing* to do with the output impedance of the amplifier, which is generally determined by the electrical feedback

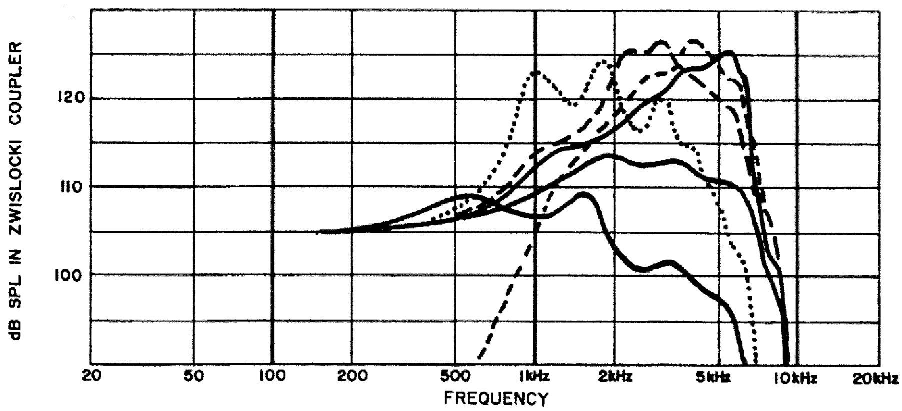

FIGURE 3.10. Frequency response tailoring using various earmold constructions with a single amplifier-receiver combination. (Reprinted with permission from Killion, 1980.)

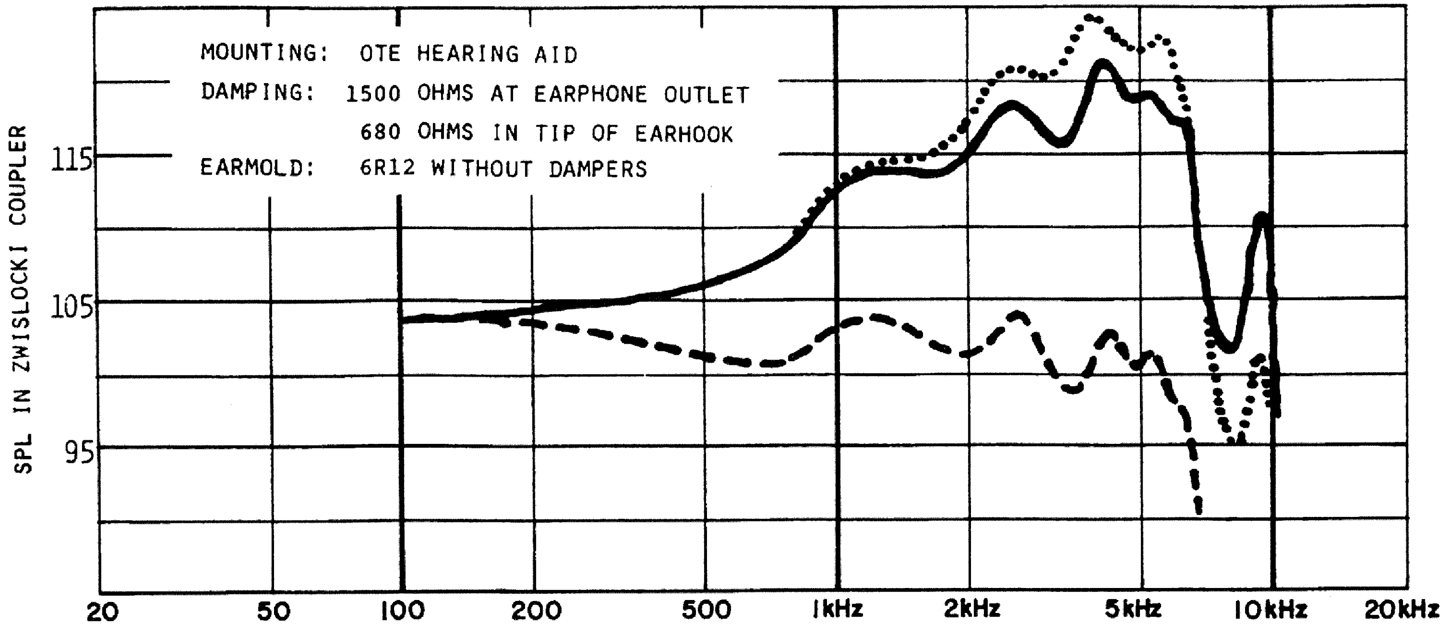

FIGURE 3.11. Frequency response tailoring with amplifier source impedance. Response of BP-series earphone obtained with electrical source of high impedance (constant current) (solid line), high impedance with shunt capacitor (dotted line), and low impedance (constant voltage) (broken line). (Reprinted with permission from Killion, 1980.)

conditions. In particular, the additional high-frequency response boost obtained by adding a capacitance in shunt with the receiver does *not* increase the undistorted high-frequency output capability of the hearing aid, as discussed in the section on distortion.

Noise

There are two basic "noise problems" with hearing aids: One is generally perceived as noise in the hearing aid, and one is generally perceived as a difficulty hearing in noisy surroundings.

"This hearing aid is noisy." An experienced dispenser with normal hearing knows that any hearing aid with sufficient gain can be turned up until its input noise becomes quite audible, even in a quiet room. The same would presumably be true for the noise in the normal auditory system if we could "turn up the gain" enough, as anyone who has listened carefully in a quiet anechoic chamber can attest. The important question is, "How close to normal thresholds can a tone in a sound field be before the amplified microphone noise masks the tone?" The answer is that with good amplifier design, the aided threshold determined by the microphone noise level can be within a few dB of normal threshold (Killion, 1976). Figure 3.12 shows the calculated HL equivalent of the BT-series microphone noise. In a recent double check of these 15-year-old calculations, *aided* sound field thresholds were obtained for a subject wearing a pair of broadband ITE aids containing EK-series microphones and a new low-noise "K-AMP" circuit. The across-frequency (250 to 6000 Hz) average was 4.3 dB HL in the first test and 2.1 dB HL in a repeat test. The across-frequency average of the *unaided* thresholds for the same normal-hearing subject was 1.4 dB HL, indicating that the sound field calibration was adequate. By any

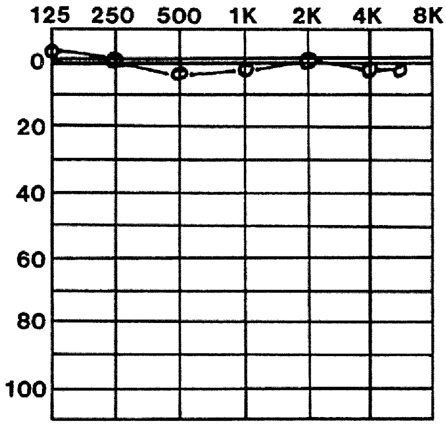

FIGURE 3.12. Equivalent HL of microphone noise: Expected aided thresholds in sound field when hearing aid gain is adjusted to exceed subject's hearing loss (BT-1751 microphone).

measure, the noise levels of modern subminiature microphones do not limit hearing aid performance.

Since modern subminiature microphones have A-weighted noise levels 5 dB lower than those found in a quiet concert hall and 15 to 20 dB lower than those found in a typical residence, it is often the amplified *ambient* noise level that is heard when the hearing aid gain is turned up. If the hearing aid does not have a smooth frequency response, even a listener with normal hearing may have difficulty recognizing these unnaturally altered and magnified background noises. A first-time hearing-aided listener, who may not have *heard* most ambient background noises for several years, may ascribe them to noise in the hearing aid regardless of the hearing aid response, although the period of adjustment appears to be much shorter for aids with smooth frequency responses.

"I can't understand speech in noise." When hearing aids had a real-ear bandwidth of only a couple of octaves, the experiments of

Tillman, Carhart, and Olsen (1970) indicated that the typical presbyacusic user would require a 15 to 30 dB *greater* signal-to-noise ratio when listening through the hearing aid than when the speech and noise were amplified by a wideband sound-field speech audiometer. Fortunately, no one needs to wear hearing aids like that anymore. Most studies since the pivotal Pascoe (1975) report have confirmed that a *properly selected* wideband hearing aid will *improve* the wearer's ability to hear speech in noise.

Distortion

Some hearing aids distort, as documented most recently by Preves and Newton (1989), usually with large input SPLs, which cause the amplified output to exceed the "headroom" designed into the hearing aid. Surprisingly high peak pressures are frequently presented to the input of the hearing aid; a spoon dropped onto a plate can produce a 110 to 114 dB SPL peak. Even commonly encountered speech can produce surprisingly high SPLs. Back when she was 5 years old, the writer's daughter's enthusiastic "Hi Dad" at 2½ feet was good for a 114 dB SPL peak at a head-worn hearing aid microphone inlet. The hearing aid user's own raised voice can easily cause 100 to 110 dB SPL peak pressures at the microphone inlet (which is, after all, only about 6" from the user's mouth).

As observed recently by Cole (1990), this distortion is basically a result of the necessary compromise between headroom and power consumption. Cole's tradeoff is shown in Table 3.1. As battery size is decreased to reduce the physical size of the hearing aid, the power consumption from the battery must go down if adequate battery life is to be preserved, and the headroom goes down with it. As Harry Teder once humorously observed, "All of my hearing aid output design problems are caused because I have to connect to an aspirin tablet rather than Hoover Dam!"

At low frequencies, the transducer manufacturer can help the designer preserve as much headroom as possible by winding more turns (of finer wire) on the internal coil as the allowable receiver bias current is decreased. With twice the number of turns, for example, the same maximum undistorted output of the hearing aid can be maintained at low frequencies with half the bias current.

The problem arises at *high* frequencies,

TABLE 3.1. Maximum SSPL-90 Versus Receiver Size and Battery Size.

Knowles Receiver Type	*Receiver Size (CC)*	*HFA-SSPL90 for 100 hr (ANSI) Battery Life*			
		#10	*#312*	*#13*	*#675*
CI	0.29	129	132	135	139
EF	0.19	125	129	132	136
ED	0.08	121	124	127	131
EH	0.056	114	118	121	125

Source: Cole, 1990; used with permission.

where headroom is *not* preserved by winding more turns on the coil. Indeed, just the opposite occurs. At high frequencies, where the electrical impedance of the receiver is increasing with frequency and thus the maximum output is voltage rather than current limited, each time the number of turns on the coil is doubled, the undistorted output drops by 6 dB. (The *impedance* of the receiver at each frequency increases by 4 times when the number of turns is doubled, and so the *voltage* required for a given high-frequency output will be doubled: one-half the current times four times the impedance). The result of increasing coil turns on maximum undistorted high-frequency output is shown in Figure 3.13.

Unfortunately, the high-frequency emphasis required of most hearing aids would require greater voltage swing at high frequencies even if receiver impedance did not rise with frequency. The combination is deadly. With the higher receiver impedances required by the lower bias currents (dropping to perhaps .3 mA in the smallest canal aids using class A amplifiers, as discussed above), the undistorted output capability at high frequencies drops so low that almost *any* sound will cause high-frequency clipping. Thus we have seen a recent rediscovery of the high-frequency intermodulation distortion problem discussed by Peterson (1951), who described a hearing aid that looked good on

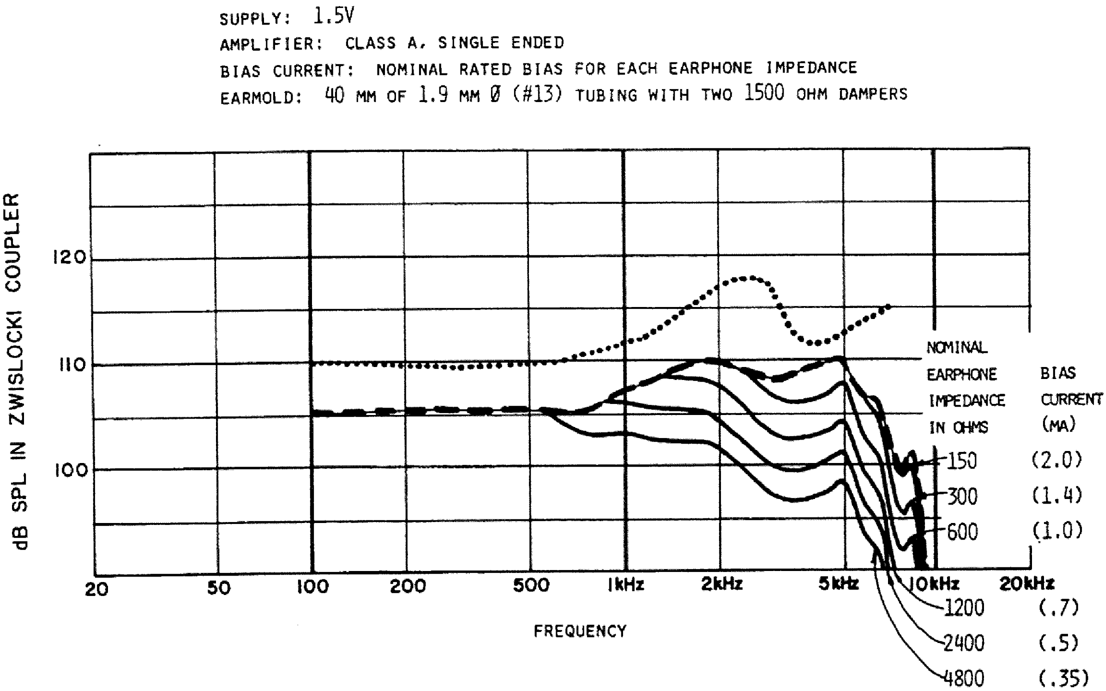

FIGURE 3.13. Maximum undistorted output of BP-series receiver with well-damped earmold, limited by earphone overload (dotted line), amplifier current clipping (broken line), and amplifier voltage clipping (solid line). (Reprinted with permission from Killion, 1980.)

paper but sounded awful. Peterson found that a high-frequency intermodulation distortion measurement correlated well with the subjective complaints.

Fortunately, anyone willing to pay a few more dollars can now obtain one of the class D amplified receivers described above. The EP-3074, for example, has an idling current of only .15 mA but a maximum undistorted output at high frequencies equivalent to the receiver with the 300-ohm impedance in Figure 3.13. Recent listening tests conducted by Palmer, Wilber, Killion, and Ballad (1990) suggest that users may be willing to pay $6.75 for each percentage-point increase in sound quality. The average sound-quality difference between a hearing aid using a high-impedance ED-series receiver with a .3 mA class A output amplifier and one using an EP-3074 (which contains an ED-series "motor" driven by a class D amplifier) was approximately 20 percentage points, worth about $135, according to Figure 3.14. [If the Palmer et al. (in preparation) results apply to fidelity ratings such as those in Figure 3.6, a hearing aid with a well-damped response should be worth $114 more than one with undamped tubing resonances. Having logged some 2000 hours wearing hearing aids, the writer as music lover and self-styled fidelity judge thinks both differences are worth much more.]

The receiver itself is seldom the cause of limited headroom in anything except very high powered hearing aid design. Readers interested in a comparison between the distortion levels in subminiature hearing aid receivers and in human ears, or in an extended discussion of the popular misconceptions about "transient response," are referred to the earlier version of this chapter (Killion, 1980). Fortunately, the widespread use of the FFT has made the intimate relationship between frequency-response shaping and oscilloscope transient response common knowledge.

SUMMARY

The summary of the previous edition of this chapter is even more valid now than in 1980. There are surprisingly few technical limita-

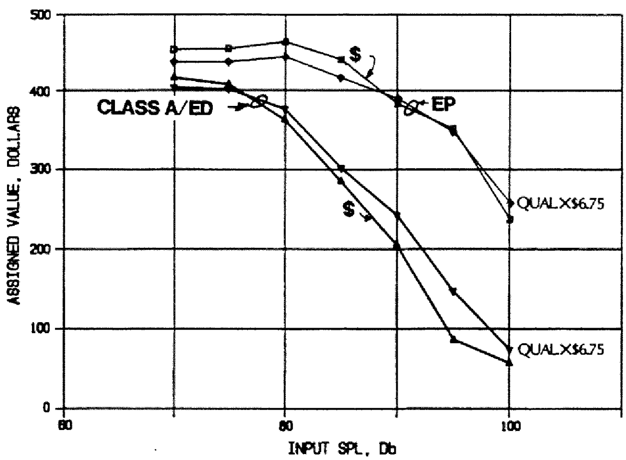

FIGURE 3.14. Sound quality and dollar value ratings for two hearing aids differing in their undistorted high-frequency outputs.

tions to hearing aid performance. Even in those cases where extremely high gains or sound pressure levels are required, the present state of the electronic and transducer art permits almost any conceivable combination of electroacoustic characteristics in head-worn hearing aids, up to and including high fidelity hearing aids.

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