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HEARING AID TRANSDUCERS

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1 INTRODUCTION

This chapter describes several decades of transducer progress that was driven primarily by one dictum: Make it smaller. In 1954, the smallest hearing aid microphone occupied 3 cc in volume and most hearing aids were pocket size and worn on the body. Thirteen design generations later, the smallest microphone occupies 0.02 cc, the smallest receiver (earphone) is nearly as small, and hearing aids are routinely fitted deeply and entirely within the ear canal. Somewhat surprisingly, the performance of some of the smaller transducers is dramatically *better* than that of their larger predecessors, particularly with regard to (a) frequency response and bandwidth, (b) available gain before feedback (previously limited by magnetic coupling and vibration coupling between the microphone and receiver), and (c) resistance to shock damage. Improvements in materials, associated amplifier design, and understanding that has cleared up long-standing misconceptions have made these results possible. Hearing aid microphone and earphone cartridges are now used in recording studio microphones and monitor earphones, providing a fidelity of reproduction meeting or exceeding that of any available transducers.

2 TRANSDUCER MINIATURIZATION

Figure 1 shows several hearing aid transducers. The largest (3 cc) is the Knowles AJ series microphone introduced in 1954; the smallest (0.02 cc) is the 12th-generation Knowles EM series microphone, introduced in 1988. (Note: The transducer model designations throughout this chapter are those of Knowles Electronics, founded by ninth ASA President Hugh Knowles, where the author studied transducer design under Elmer Carlson from 1962 to 1983.)

In addition to the obvious cosmetic advantages, the reduction in transducer size 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 decibels of direct "binaural squelch" for interfering noise and reverberation as well as a much more pleasant sense of auditory space. With in-the-ear (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.[1] 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 only one working ear to experience a surprisingly good sense of auditory space and ability to localize sounds.

3 TRANSDUCER TYPES

The most energy-efficient receiver designs are magnetic, typically using a push–pull type of "balanced armature" magnetic construction such as illustrated in Fig. 2.[2] Other transducer mechanisms have been proposed, but to the writer's knowledge none come close to providing

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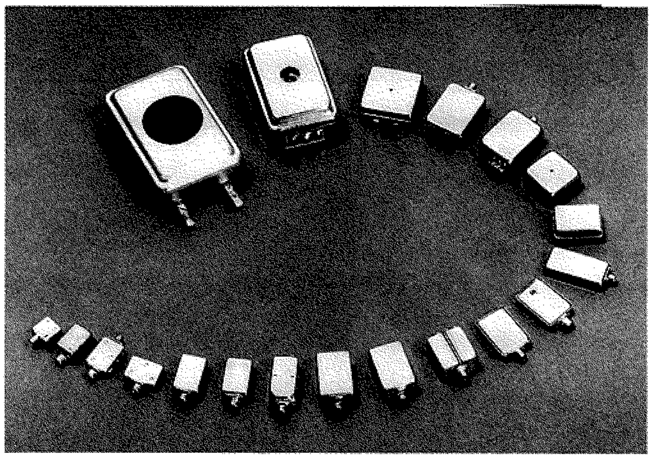


Fig. 1 Selected hearing aid transducers from 1954 to the present. (Courtesy Knowles Electronics.)

the same electromechanical coupling coefficiency; piezoelectric bimorphs, the next nearest competitor, appear to be 10 dB less efficient in converting electrical energy to acoustic output. The electromechanical coupling coefficient for balanced-armature receivers is $k = 0.8$, while for bimorphs $k = 0.25$, so the proportions of electrical energy converted to the mechanical side (k^2) are 0.64 and 0.06, respectively. The dominance of the magnetic receiver is thus readily understood. Hearing aid wearers must carry the power source around with them, and the receiver consumes 50–95% of the power in a well-designed hearing aid. Thus receiver efficiency determines almost directly how often the battery must be changed: A 10-dB reduction in efficiency would require purchasing five to nine times as many batteries each year!

The first headworn hearing aid microphones were also magnetic, as 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 their low input impedance. Once low-noise junction field-effect transistor (JFET) preamplifiers became available, however, energy efficiency in the microphone ceased to be important since the JFET could be driven by a low-current, low-energy signal. The lower efficiency but higher *voltage sensitivity* piezoelectric ceramic and electret microphone elements thus became practical.

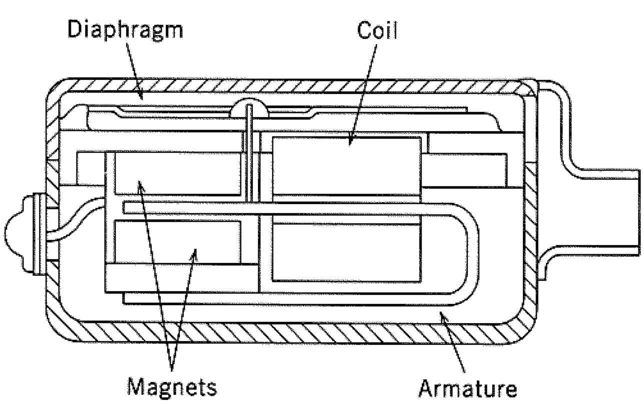


Fig. 2 Balanced armature magnetic receiver. (Courtesy Knowles Electronics.)

Figure 3 illustrates the dramatically different rank ordering that results when magnetic, ceramic, and electret microphones are compared using a low input-impedance amplifier (Fig. 3*a*) and a high-input-impedance amplifier (Fig. 3*b*).

Figure 4 shows the construction of a typical hearing aid electret-condenser microphone.[3] 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 neighborhood of 1000 Ω.

4 PERFORMANCE VERSUS SIZE

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. As an illustration of the latter improvement, Figure 5 shows the improved (reduced) vibration sensitivity of hearing aid 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—are a direct result of reduced magnetic and vibration coupling between microphone and receiver. (The remaining limitation is acoustic coupling in the form of sound leakage from the ear canal back to the microphone opening. This leakage can be due to intentional "venting" of the ear mold or an inadequate seal of the ear mold in the ear canal.) The minimal shock mounting routinely used in modern ITE and canal hearing aids has been made practical because of increasingly rugged transducers: One modern hearing aid receiver will withstand 20,000g shock; modern microphones will all withstand 20,000g or more without damage.

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 as

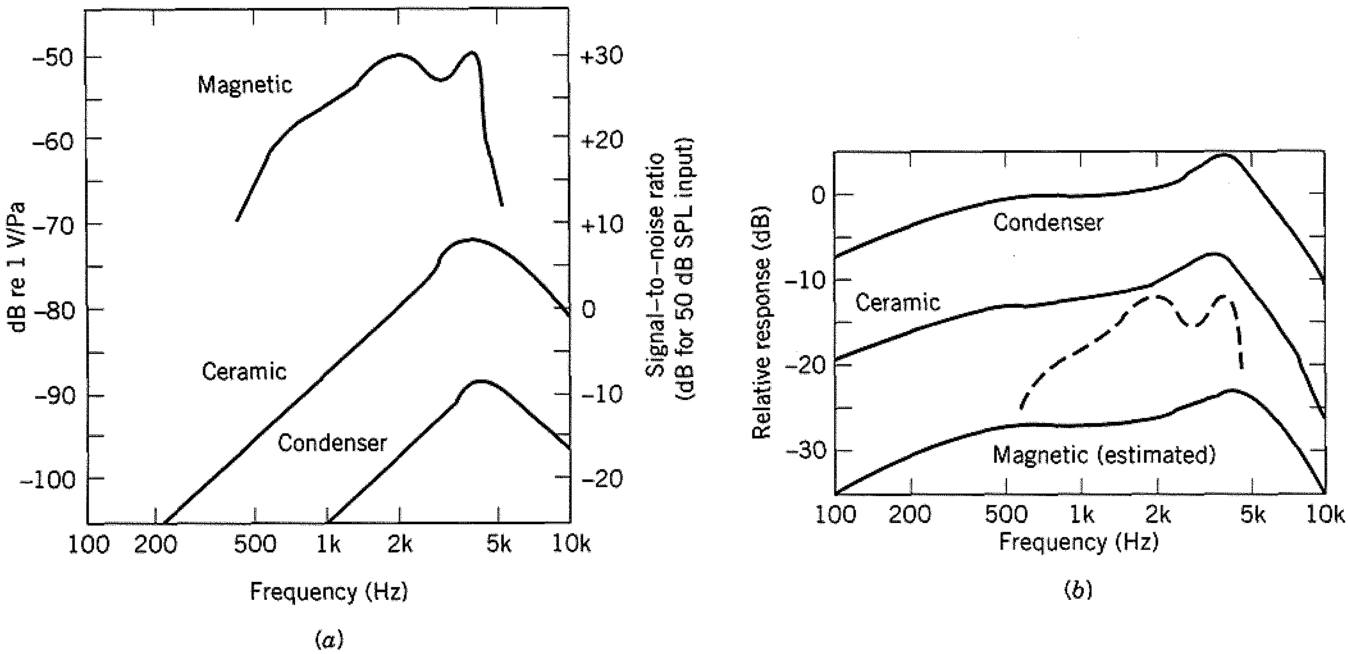


Fig. 3 Rank ordering of the sensitivity of magnetic, piezoelectric ceramic, and electret–condenser microphones operating into (*a*) low-input-impedance (5000-Ω) amplifier and (*b*) high-input-impedance (100 mΩ) amplifier. [The dashed curve in (*b*) shows the sensitivity of a commercial narrow-band magnetic microphone. The lowest curve is the estimated sensitivity of the same microphone with sufficient damping to produce the same frequency response as the condenser and ceramic microphones.] (Reprinted, with permission, from Killion, 1991.)

low a frequency as you would wish (less than 1 Hz with a high-quality ear mold seal and a high-resistance 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 tendency 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.

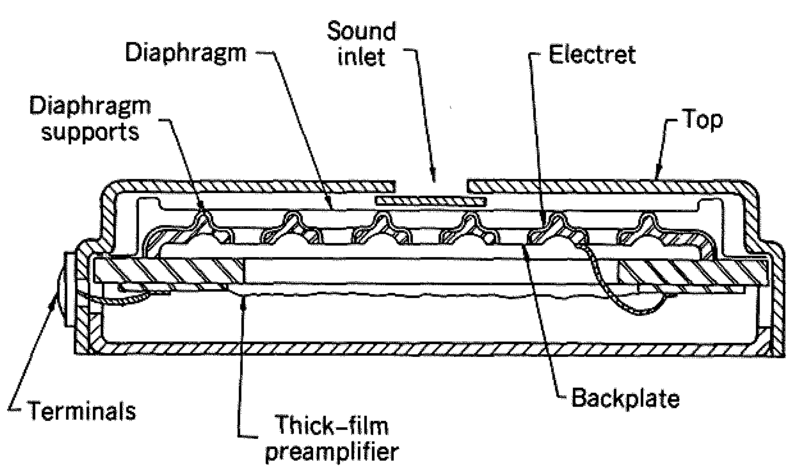


Fig. 4 Electret–condenser microphone with built-in preamplifier. (Reprinted, with permission, from Ref. 3.)

4.1 Microphone Noise

The signal-to-noise level of any microphone that is small compared to a wavelength should be proportional to the square root of the case volume, as can be seen by considering the 3-dB improvement obtained if two microphones are T-connected to the same sound inlet tube and their electrical outputs are connected in series. The resulting *signal* sensitivity will be 6 dB greater, because the signal voltage outputs of the two microphones 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.

While this reasoning holds in general, improvements in materials and design can sometimes provide surprising results. As an example, based on size considerations, the EK series (Fig. 5) electret microphone should be 9 dB noisier than the BA series magnetic microphone because the BA microphone has eight times greater volume, but the EK microphone is just as quiet. While the energy efficiency of the EK microphone is much lower than that of the BA, the EK's intrinsic noise level is lower still due to its lower internal losses (real part of its acoustical, mechanical, and electrical impedances). The equivalent sound pressure level (SPL) of the noise in a 1-Hz bandwidth at 1 kHz is about −14 dB for both microphones. Neither is any limitation to hearing, as discussed in Section 7.

4.2 Receiver Sensitivity

A similar theoretical expectation holds for the sensitivity and maximum undistorted output of hearing aid receivers, both of which 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 dominant impedance determining the motion of a subminiature receiver's diaphragm is not that of the 0.6 cc or so of air in the occluded ear canal (or of the total 1.3 cc equivalent volume when the compliance of the eardrum is included) but that of the much smaller volume of air trapped *behind* the receiver diaphragm. This amounts to less than 0.04 cc in the smaller receivers, so that producing 100 dB SPL in the ear canal at low frequencies requires the production of 130 dB SPL in the "back volume" behind the receiver diaphragm.

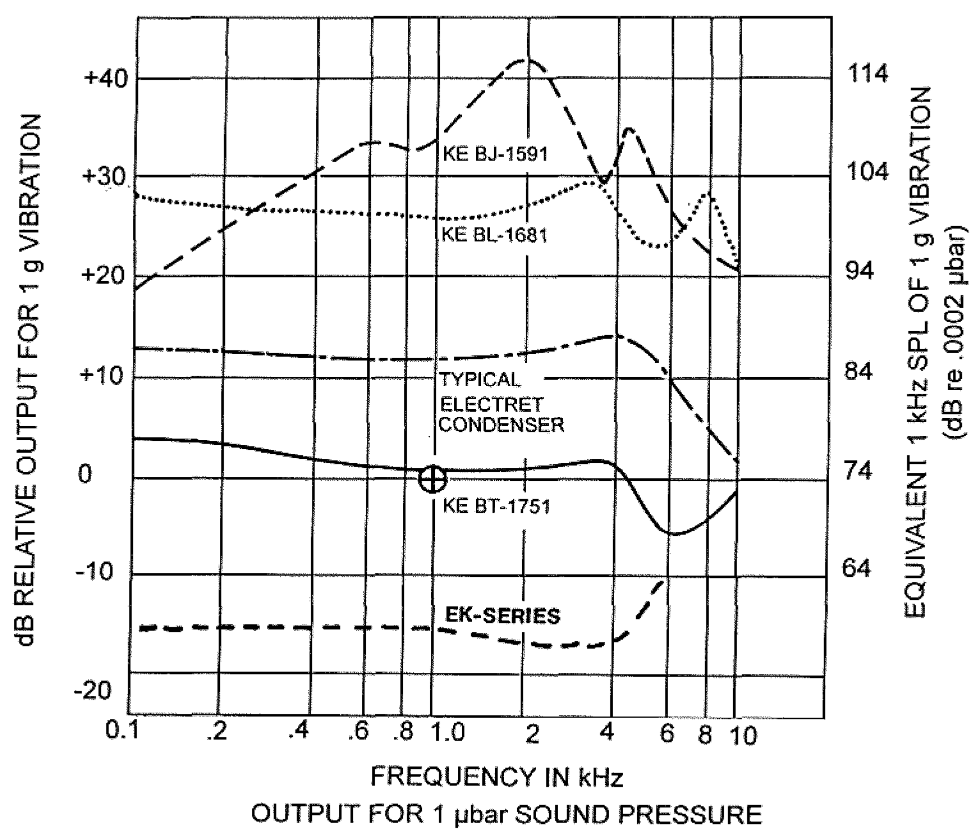


Fig. 5 Vibration sensitivity of magnetic, piezoelectric ceramic, and three generations of electret–condenser microphones. (Adapted, with permission, from Ref. 4.)

Thus if the outputs of two equal receivers are combined into a single double-area 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 virtually the same but the amplifier output current will drop by half. This can be seen as follows: With the two receiver coils in series, their combined electrical impedance Z will be double and, by Ohm's law, each will experience half the former coil current ($i = e/Z$). Since each receiver's diaphragm motion is determined almost completely by the impedance of the back volume and thus is virtually unaffected by the presence of the other receiver, each receiver will now produce half the volume displacement. The *total* volume displacement of air therefore, 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 (0.12 cc) has a sensitivity only 1 dB less than that of the BK series receiver (0.18 cc), which occupies 1.5 times the volume and thus might be expected to be nearly 2 dB less sensitive. Here, again, steady improvements in materials and design have partially held off the presumed automatic loss of performance with decreasing size.

When the same improvements in materials and design are applied to the redesign of an older, larger receiver, a substantial performance improvement can result. The CI series receiver, for example, occupies the same case volume as the older BI series receiver (0.29 cc) yet requires only half the power to generate the same SPL and 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 receivers, which required the highest battery drain, *and* used the smallest (lowest milliampere-hour capacity) battery. In order to provide the one-week battery life that is generally considered the minimum acceptable, the manufacturer was limited to approximately 0.3 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, which left it marginal at best (see Section 8).

4.3 Class D Receivers

Just as the availability of JFET preamplifiers allowed a breakthrough in microphone design, the availability of low-voltage silicon gate complementary metal–oxide–semiconductor (CMOS) circuitry made it possible to include a subminiature class D (high-efficiency switching) drive amplifier inside the receiver case.[5,6]

The battery consumption of a class D amplifier is intrinsically low at idle (low-output SPL) and 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 0.02 mA over its idling value of 0.15 mA. In cooperation with appropriate amplifier design, the result can be a complete hearing aid that averages only 0.32 mA of battery drain yet has an undistorted output capability of 110–115 dB SPL. Since the instantaneous peaks in speech and music are typically 10–15 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!

4.4 Receiver Vibration and Feedback

One important characteristic of subminiature receivers *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 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 Ref. 4.) The most recent microphones can be safely cemented directly to the hearing aid housing with either no increase in feedback problems or, when previously a rubber tube allowed the microphone to "pump" on the tube, a decrease in feedback problems.

The sound leakage from an intentional vent channel in the ear mold or an inadvertent slit leak along the ear canal exiting near the microphone inlet is a common cause of feedback, but when those causes are corrected, the limitation is usually the SPL developed by the entire hearing aid case acting like a miniature speaker 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[7] routinely obtained from two receivers mounted "belly to belly" so their mechanical vibration outputs subtract but their acoustic outputs add. This arrangement of two receivers can provide an increased acoustic sensitivity (as described in Section 4.2) while virtually cancelling the vibration output and is now commercially available as the EJ series of dual receivers. Even without dual receivers, most manufacturers have developed compliant mounting and careful orientation of the receiver into a finely tuned art in order to provide high-gain instruments.

5 BANDWIDTH

At one time hearing aid transducers were intentionally designed with a bandwidth restricted to the region between 500 and 3000 or 4000 Hz.[8] The common belief at the 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,[9] since the frequency bands below 400 Hz contained only 5% of the speech information and those bands above 4000 Hz contained only 15% of the speech information.

More recent experiments, however, have indicated that (1) filtering out low- and high-frequency noise simultaneously filters out useful low- and high-frequency speech cues and (2) 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 traditional telephone with its 500–3000-Hz passband and its 30% total harmonic distortion typically yielded nearly perfect sentence intelligibility if the talker was in a reasonably quiet room. The problem comes in the presence of noise, where a 10% increase in articulation index may result in a 30–50% increase in sentence intelligibility and the maximum possible audible bandwidth is called for.

Fortunately, the hearing aid 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–16 kHz of the ear) for nearly 15 years. The flat-frequency-response BT-1759 version of the BT series electret hearing aid microphone has been regularly used in broadcast and recording studios since the early 1970s, and a standard response ED series receiver is used in what are arguably two of the most accurate high-fidelity earphones available (the 16-kHz-bandwidth Etymotic Research ER-1 and ER-4 insert earphones). A direct demonstration that hearing aid transducers permitted high-fidelity sound reproduction in a series of simulated-live-versus-recorded fidelity rating experiments has also been reported.[10]

6 FREQUENCY RESPONSE SHAPING

A real-ear frequency response that has been tailored to the hearing loss often provides improved speech intelligibility—especially in noise—for the hearing aid user. To accomplish such response tailoring, the manufacturer typically chooses from among the wide variety of available microphone and receiver frequency responses, combines those with frequency response shaping in the hearing aid amplifier itself, and delivers the finished product to the dispenser who, in recent years, tests the insertion response of the delivered aid on the intended user. In the case of the behind-the-ear aids, the dispenser might modify the ear mold to obtain additional response control.

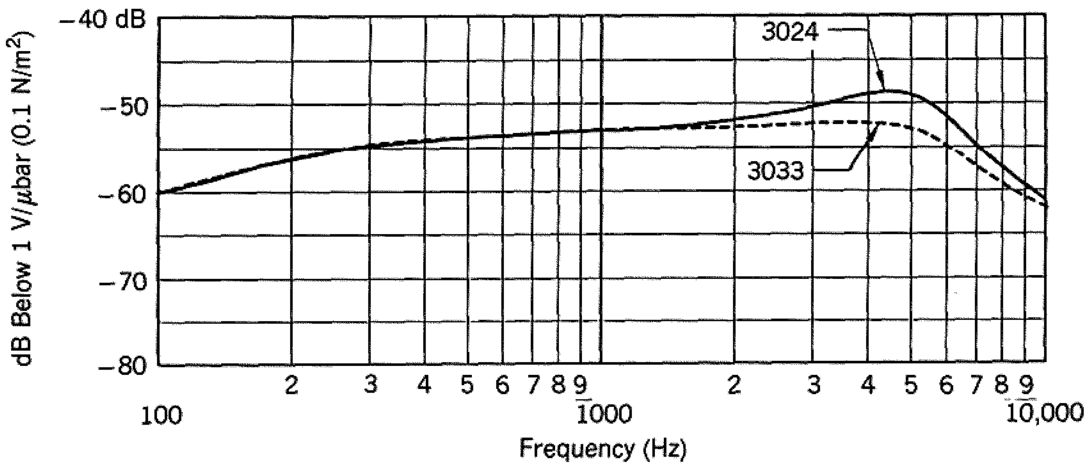
6.1 Microphone Responses

A sampling of the available electret microphone responses is shown in Fig. 6. (These are plotted to the old General Radio 30 dB/decade engineering scale rather than the ANSI standard 50 dB/decade hearing aid scale, a transducer tradition that goes back nearly as far as the use of the term *receiver*). A range of 0–20 dB high-frequency boost between 0.5 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 could only be achieved electronically by adding several additional components.

6.2 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 in order 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 3–5 dB in signal-to-noise ratio even under difficult reverberant listening conditions[11] and can provide an even greater benefit in some special situations.

In order to obtain good in situ performance in directional-microphone hearing aids, the effect of head and ear diffraction must be taken into account. The loca-



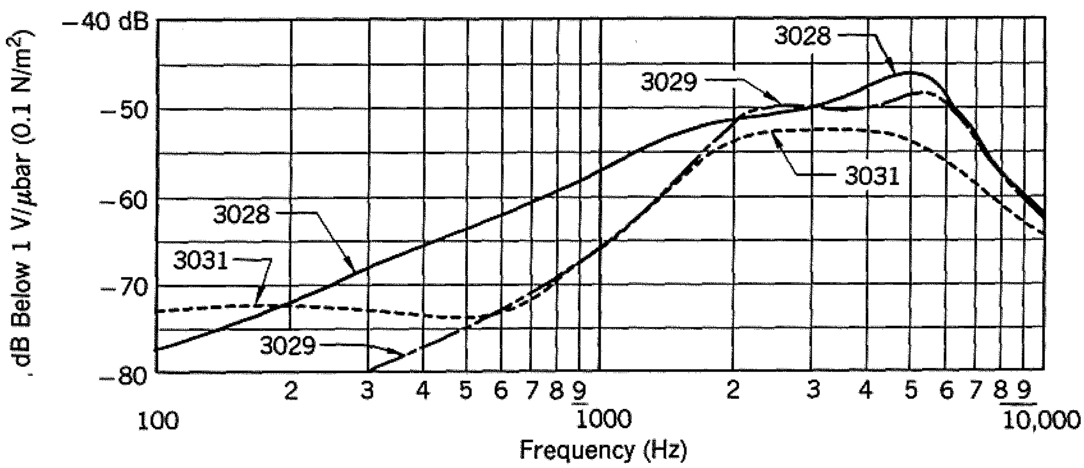


Fig. 6 Variety of frequency response curves available in EK series microphones. (Courtesy of Knowles Electronics.)

tion 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 0.7 times the physical spacing. Figure 7 illustrates different directional characteristics obtained with various combinations of effective port spacing and time delay.[12] Many early directional-microphone hearing aid designs had poor directivity above 1–2 kHz, but Madaffari demonstrated that 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.

6.3 Receiver-Plus-Ear-Mold Response

Attempting to define the response of a hearing aid receiver without specifying the ear mold is a nearly futile task. The ear mold *controls* the delivered response of the combination, as illustrated in Fig. 8. The curves in that figure were for a *single* receiver run with various "horn" and "reverse horn" ear mold configurations. Only one of the curves represents a vented ear mold. The topic of ear mold acoustics is beyond the scope of this chapter, but several excellent references are available.[14–17] Further response shaping can be obtained with the use of acoustic filter ear hooks that can provide 24 dB/octave low-frequency, high-frequency, or bandpass rejection when used with BTE hearing aids.[18]

6.4 Amplifier Source Impedance

A final response control measure available to the hearing aid designer should be mentioned. Figure 9 shows the change in earphone frequency response caused by changes in the source impedance, that is, 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-

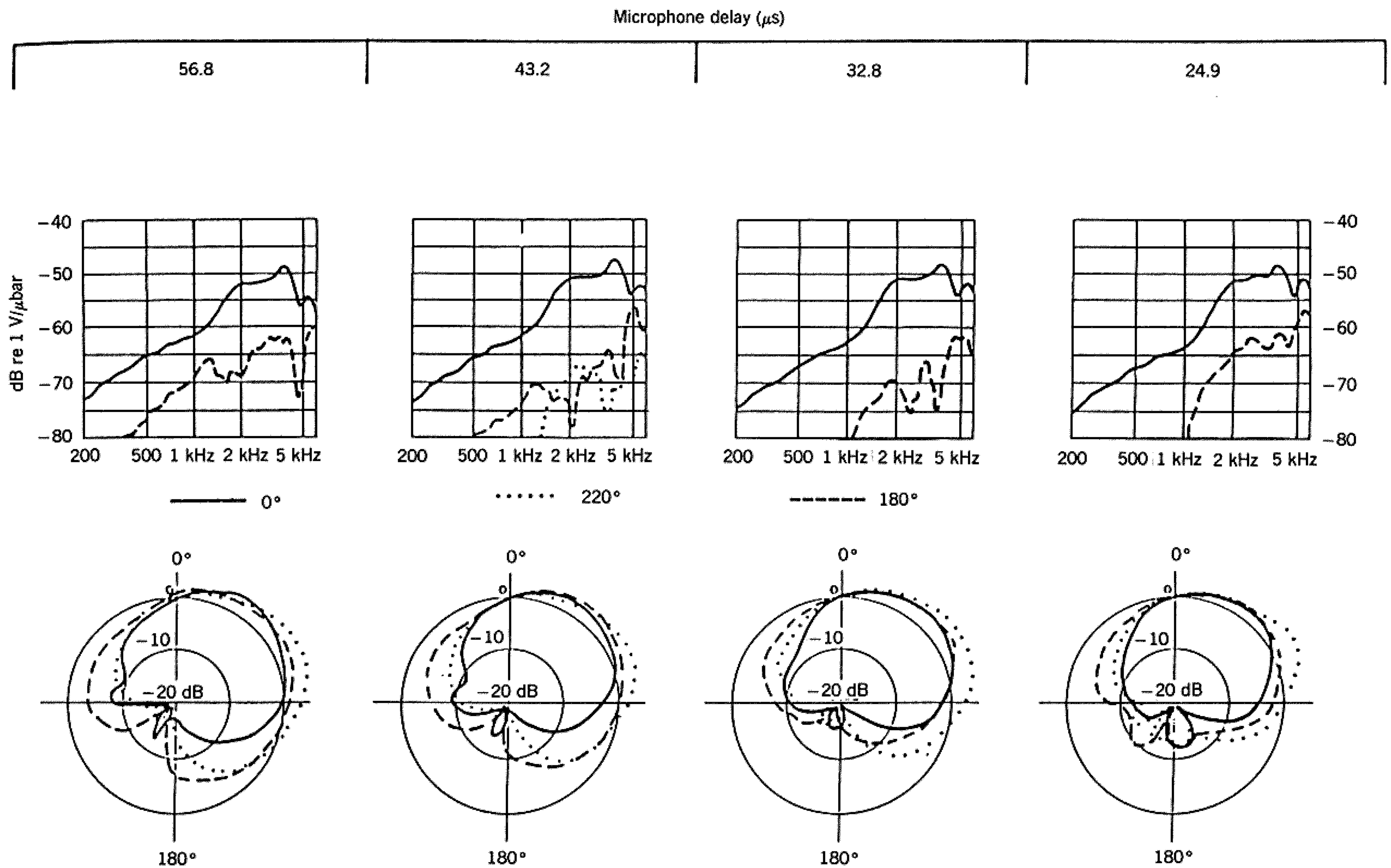


Fig. 7 In situ (KEMAR manikin) directional microphone frequency and polar responses for 8.3 mm effective port spacing and microphones with various internal time delays. (Reprinted, with permission, from Ref. 12.)

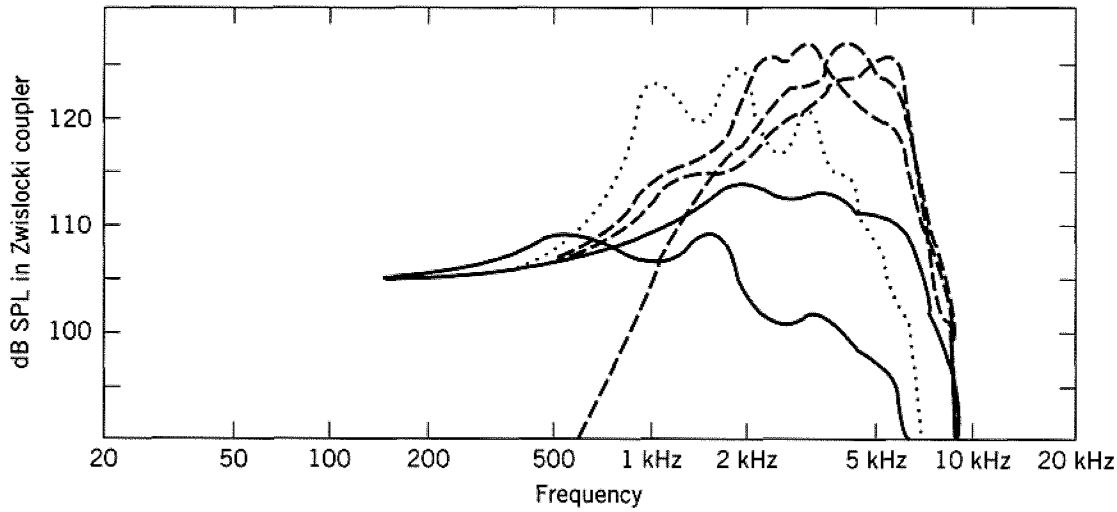


Fig. 8 Frequency response tailoring using various ear mold constructions with a single amplifier–receiver combination. (Reprinted, with permission, from Ref. 13.)

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 Fig. 9, some hearing aids include a capacitor in series with an adjustable trimmer potentiometer to allow adjustment of the resonance peak frequency. As illustrated in Fig. 9, the boost in response below resonance is accompanied 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 three curves in Fig. 9 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 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-

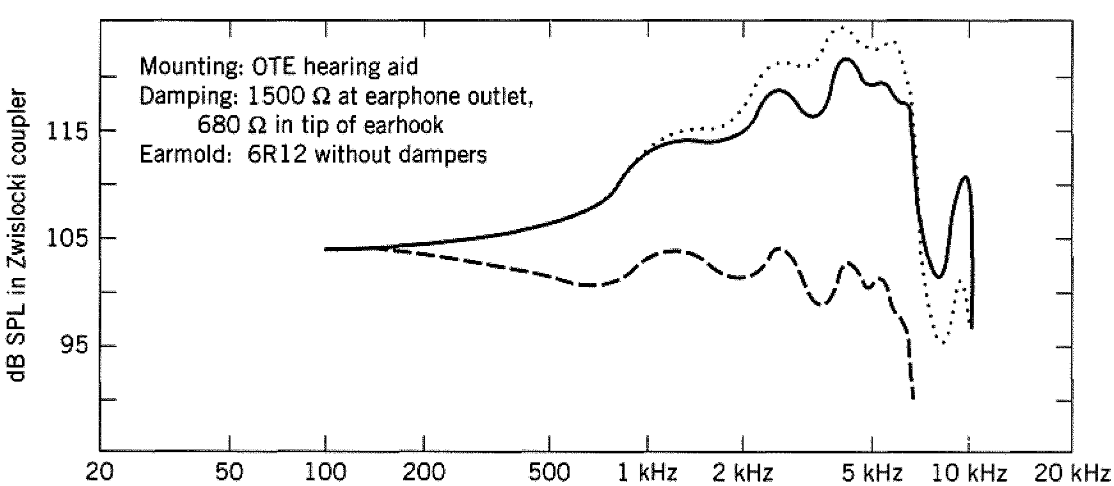


Fig. 9 Frequency response tailoring with amplifier source impedance. Response of BP series earphone obtained with electrical source of high impedance (solid line), high impedance with shunt capacitor (dotted line), and low impedance (broken line). (Reprinted, with permission, from Ref. 13.)

frequency output capability of the hearing aid, as discussed in Section 8.

7 NOISE

The complaint that "this hearing aid is noisy" often results not from noise in the hearing aid but from noise in the room that is not recognized or localized as such.

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 be obtained before the amplified microphone noise masks the tone in the sound field?" The answer is that with good amplifier design, the aided threshold determined by the microphone noise level can be within a few decibels of normal threshold.[19] The calculated audiometric hearing level (HL) equivalent of the BT series microphone noise at the 250–4000 octave frequencies ranges from 0 to 4 dB HL, with an average value of 1.6 dB at those five frequencies. In a recent double check of these 15-year-old calculations, *aided* sound field thresholds were obtained for a subject wearing a pair of commercially available broadband ITE hearing aids containing EK series microphones. The across-frequency average of the aided thresholds was 3.2 dB HL, essentially equal to the calculated values and indicating that aided thresholds near audiometric zero levels can now be routinely obtained in practical hearing aids. By any measure, the noise levels of modern subminiature microphones do not limit hearing aid performance.

Since modern subminiature microphones have A-weighted noise levels some 5 dB lower than that found in a quiet concert hall and some 15–20 dB lower than that 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 shorter for aids with smooth frequency responses.

8 DISTORTION

Surprisingly high peak pressures are frequently presented to the input of the hearing aid; a spoon dropped onto a plate can produce a brief 100–114-dB SPL peak. More pleasant but similar peak pressures are produced by the Chicago Symphony Orchestra in Chicago's Orchestra Hall. Even commonly encountered speech can produce surprisingly high SPLs. Back when she was five years old or so, the writer's daughter's enthusiastic "Hi Dad" at $2\frac{1}{2}$ 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 100–110 dB SPL peak pressures at the microphone inlet (which is, after all, only about 6 in. from the user's mouth).

It is thus often difficult to provide adequate "headroom" (the available undistorted output reserve above the nominal operating levels) before overload—and audible distortion—sets in.[20] This overload distortion is most often a direct result of the traditional compromise between headroom and power consumption. Table 1 shows this trade-off.[21] As battery size is decreased to reduce the physical size of the hearing aid, the power consumption from the battery must decrease if adequate battery life is to be preserved and headroom decreases with it.

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, 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

TABLE 1 Maximum Hearing Aid Output versus Receiver Size and Battery Size

Knowles RCVR Type	RCVR Size (cc)	HFA-SSPL90 for 100 HR Battery Life by Battery Type Number (dB SPL)			
		No. 10	No. 312	No. 13	No. 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

Note: HFA SSPL90 is the ANSI standard abbreviation for high-frequency-average (1.0, 1.6, and 2.5 kHz) saturation SPL output of a hearing aid, measured with volume control set at maximum and a 90 dB SPL input.
From Ref. 21. Used with permission.

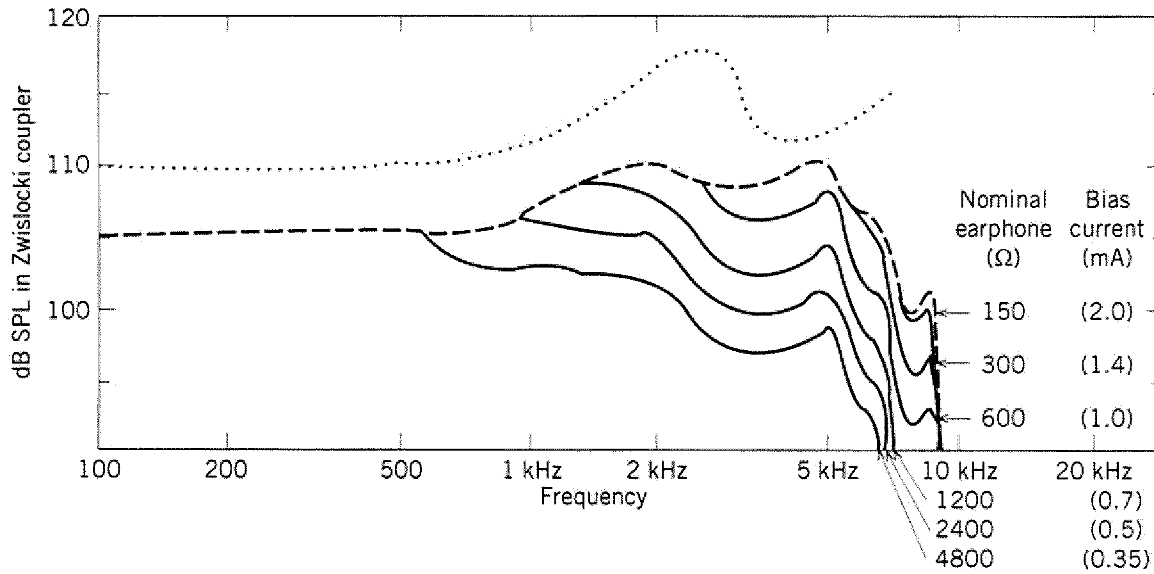


Fig. 10 Maximum undistorted output of BP series receiver with well-damped ear mold limited by earphone overload (dotted line), amplifier current clipping (broken line), and amplifier voltage clipping (solid line). Each curve obtained with the same class A amplifier biased to the currents shown. (Reprinted, with permission, from Ref. 13.)

drops by 6 dB. (The *impedance* of the receiver at each frequency increases by four 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 relationship between receiver impedance, DC bias current, and maximum undistorted high-frequency output is illustrated in Fig. 10.

Unfortunately, the combination of the high-frequency emphasis required of most hearing aids with the use of higher receiver impedances and lower bias currents (dropping to perhaps 0.3 mA in the smallest canal aids using class A amplifiers, as discussed above) caused the undistorted output capability at high frequencies to drop so low that almost *any* sound would cause high-frequency clipping and intermodulation distortion.[22] The class D amplified receivers described above provided a convenient means of circumventing the dilemma described above.

The EP-3074, for example, has an idling current of only 0.15 mA but a maximum undistorted output at high frequencies equivalent to the receiver with the 300-Ω impedance in Fig. 10. The receiver mechanism itself is seldom the cause of limited headroom in anything except very high powered hearing aid designs: A comparison between the distortion levels in subminiature hearing aid receivers and in human ears, as well as an extended discussion of some popular misconceptions about "transient response," is found in Ref. 13.

9 APPLICATIONS

A few examples of non-hearing-aid applications of hearing aid transducers may be useful.

The excellent noise performance of recent subminiature electret–condenser microphones has made it possible to measure the acoustic emissions—stimulated and spontaneous—that arise from the proper operation of normal outer hair cells in the cochlea. The apparent noise level of the normal ear (Killion[19]) is 10–15 dB greater than that of a commercial cochlear-emissions probe microphone such as the Etymotic Research ER-10.

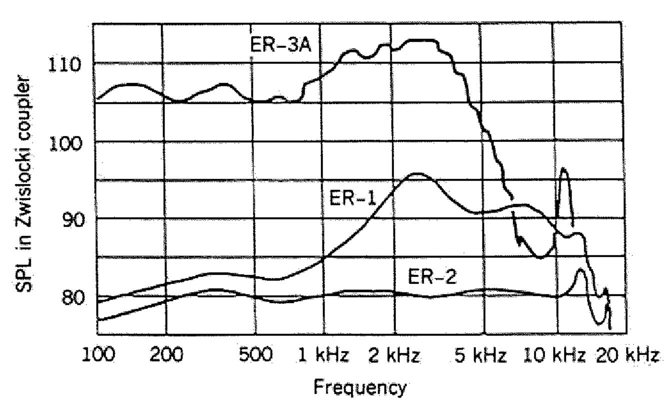


Fig. 11 Frequency response of three insert earphones with 0.1-V (1-mW) drive.

The large undistorted output of the CI series receivers (Section 4.2) has allowed the design of insert earphones that mimic the frequency response and calibration sensitivity of the more traditional supra-aural audiometric earphones such as the TDH-39 headphone that was supplied with the vast majority of audiometers for several decades.

The large bandwidth of the ED series earphone has permitted the design of insert earphones that have a nearly flat eardrum pressure response from 100 Hz to 16 kHz or mimic the normal diffuse-field pressure response of the external ear. Figure 11 shows a frequency response and sensitivity comparison of three commercially available insert earphones. The availability of earphones originally designed for hearing aids has made possible insert earphones that can provide much better noise exclusion, interaural isolation, and real-ear response smoothness than more traditional supra-aural and circumaural earphones.

REFERENCES

1. E. Shaw, "The Acoustics of the External Ear," in G. A. Studebaker and I. Hochberg (Eds.), *Acoustical Factors Affecting Hearing Aid Performance*, Williams and Wilkins, Baltimore, MD, 1980, pp. 109–125.
2. E. V. Carlson, F. W. Cross, and M. C. Killion, Miniature acoustic transducer of improved construction, U.S. Patent 3,588,383, 1971.
3. M. C. Killion and E. V. Carlson, "A Subminiature Electret–Condenser Microphone of New Design," *J. Audio Eng. Soc.*, Vol. 22, 1974, pp. 237–243.
4. M. C. Killion, "Vibration Sensitivity Measurements on Subminiature Condenser Microphones," *J. Audio Eng. Soc.*, Vol. 23, 1975, pp. 123–127.
5. M. C. Killion, Class D Hearing Aid Amplifier, U.S. Patents 4,592,087 and 4,689,819, 1986 and 1987.
6. E. V. Carlson, "An Output Amplifier Whose Time Has Come," *Hearing Instrum.*, Vol. 39, 1988, pp. 30–32.
7. M. Harada, Single Port Coupler for Resonance Splitting in a Hearing Aid, U.S. Patent D300,660, 1989.
8. M. C. Killion and E. V. Carlson, "A Wideband Miniature Microphone," *J. Audio Eng. Soc.*, Vol. 18, 1970, pp. 631–635.
9. N. R. French and J. C. Steinberg, "Factors Governing the Intelligibility of Speech Sounds," *J. Acoust. Soc. Am.*, Vol. 19, 1947, pp. 90–119.
10. M. C. Killion and T. W. Tillman, "Evaluation of High-Fidelity Hearing Aids," *J. Speech Hear. Res.*, Vol. 25, 1982, pp. 15–25.
11. D. B. Hawkins and W. S. Yacullo, "Signal-to-Noise Ratio Advantage of Binaural Hearing Aids and Directional Microphones under Different Levels of Reverberation," *J. Speech Hear. Disorders*, Vol. 49, 1984, pp. 278–286.
12. P. L. Madaffari, "Directional Matrix Technical Report," IRPI Report #10554, Knowles Electronics, Franklin Park, IL, 1983.
13. M. C. Killion, "Problems in the Application of Broadband Hearing Aid Earphones," in G. A. Studebaker and I. Hochberg (Eds.), *Acoustic Factors Affecting Hearing Aid Performance*, Williams and Wilkins, Baltimore, MD, 1980, p. 229.
14. R. M. Cox, "Acoustic Aspects of Hearing Aid–Ear Canal Coupling Systems," Monographs in Contemporary Audiology, Vol. 1, pp. 1–44, Maico Hearing Instruments, Minneapolis, 1979.
15. S. F. Lybarger, "The Earmold as an Acoustical Control Factor," in G. A. Studebaker and I. Hochberg (Eds.), *Acoustic Factors Affecting Hearing Aid Performance*, Williams and Wilkins, Baltimore, MD, 1980, pp. 197–217.
16. M. C. Killion, "Recent Earmolds for Wideband OTE and ITE Hearing Aids," *Hear. J.*, Vol. 37, No. 8, 1984, pp. 15–22.
17. M. C. Killion, "Earmold Design: Theory and Practice," in J. H. Jensen (Ed.), *Hearing Aid Fitting Theoretical and Practical Views*, 13th Danavox Symposium, Stougaard Jensen, Copenhagen, 1988.
18. M. C. Killion and D. W. Wilson, "Response-Modifying Earhooks for Special Fitting Problems," *Audecibel*, Fall, 1985, pp. 28–30.
19. M. C. Killion, "Noise of Ears and Microphones," *J. Acoust. Soc. Am.*, Vol. 59, 1976, pp. 424–433.
20. D. A. Preves and J. R. Newton, "The Headroom Problem and Hearing Aid Performance," *Hear. J.*, Vol. 42, No. 10, 1989, pp. 19–26.
21. W. A. Cole, Comments on the "headroom problem," in M. Valente (Ed.), *Hearing Aids: Standards, Options and Limitations*, 1996, Chapter 1 (Written by D. Preves).
22. A. P. G. Peterson, "Intermodulation Distortion," *Gen. Radio Experim.*, Vol. 25, 1951, pp. 179–188.