CHAPTER 5

CORFIG AND GIFROC: REAL EAR TO COUPLER AND BACK

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The acronym "CORFIG," as used to describe the transformation that predicts the *2-cc coupler* response a hearing aid should have to provide a given *insertion response*, has gained a certain acceptance since the original version of this chapter was written 10 years ago. Whether or not its arithmetic inverse, "GIFROC," will gain similar acceptance remains to be seen, but the basic utility of the transformation from real-ear to 2-cc coupler measurements (and vice versa) is well established. In the 1960s no one knew for certain what the 2-cc coupler response of a hearing aid *should* be in order to provide a favorable real-ear insertion response to the hearing aid user. In the 1970s KEMAR and real-ear measurements were still viewed with caution. By the 1980s the widespread use of coupler correction curves gave rise to widespread improvements in hearing aid design and fitting.

We are now at the point where the first-order corrections are well known, and it is possible to concentrate on second-order effects and refinements. After defining the pertinent terms, this chapter concentrates on the factors affecting the real-ear-to-coupler and coupler-to-real-ear correction curves.

DEFINITIONS AND RELATIONSHIPS

2-cc Coupler Gain and Zwislocki Coupler Gain

The *2-cc coupler gain* of a hearing aid at a given frequency is the difference between the SPL developed (by the output of the hearing aid) at the microphone of the 2-cc coupler and the SPL delivered to the inlet of the hearing aid microphone. The *Zwislocki coupler gain* or *occluded-ear simulator gain* of a hearing aid is defined similarly, except that the SPL developed by the output of the hearing aid is measured at the microphone of the Zwislocki coupler, more formally called an occluded-ear simulator (ANSI S3.25-1979).

The difference between the 2-cc coupler and Zwislocki coupler gain of a hearing aid can be predicted almost exactly, in advance, for the technical reason that a hearing aid is a high-impedance acoustic source (compared to the low-impedance acoustic load presented by either of the couplers, or the ear for that matter), and thus the SPL developed in either coupler will be proportional to the acoustic impedances of those couplers (Dillon, 1985). Whatever the technical reason, the experimental results are shown in Figure 5.1, based on the early data of Sachs and Burkhard (1972).

Real-Ear-to-Coupler Level Difference (RECD)

For the practicing dispenser, the important difference is not the one between the level in the Zwislocki and the 2-cc coupler, but between the real ear and the 2-cc coupler: the *real-ear-to-coupler level difference* (RECD). Fortunately, the Zwislocki coupler is a good representation of the acoustic impedance of the

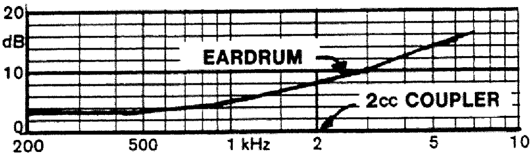

FIGURE 5.1. Real-ear-to-coupler level difference (RECD): The increase in average-eardrum or Zwislocki coupler SPL over 2-cc coupler SPL for a high-acoustic-impedance source such as a hearing aid or insert earphone.

average ear, so the curve of Figure 5.1 provides a good estimate of the difference between the *eardrum SPL* and the *2-cc coupler SPL* that a hearing aid will produce if the SPL at the inlet of the hearing aid microphone is the same in both cases.

The previous statement requires several additional qualifiers. In addition to the obvious assumptions that the volume control position was not changed between measurements and that the battery had not run down, the statement presumes that the identical earmold was used for both measurements. This is automatically true in the case of an in-the-ear (ITE) or canal aid, whose casing *is* the earmold; it is generally not true in the case of behind-the-ear (BTE) hearing aids, whose 2-cc coupler gain is typically measured with the "HA-2" version of the 2-cc coupler. The HA-2 coupler contains a built-in coupling (an 18 mm by 3 mm bore) that acts as an acoustic horn, producing 5 to 8 dB of high-frequency boost in the delivered output of the hearing aid. Unfortunately, earmold laboratories report that the majority of BTE earmolds are still ordered and delivered with constant-diameter couplings (typically the 1.93 mm inner diameter of #13 tubing), couplings that provide little of the high-frequency boost typically desired from a hearing aid.

Editorial comments aside, the difference between the eardrum SPL and the 2-cc coupler SPL that a BTE hearing aid will produce with identical SPL at the microphone inlet is determined *both* by the ear-to-coupler impedance differences and by the earmold configuration. Figure 5.2 shows the effect of different earmolds on the output of BTE hearing aids. The HA-2 earmold simulation is chosen as the reference condition (Dillon, 1985). Because the acoustic source impedance of any modern hearing aid will be many times greater than the load impedance presented by either the ear or a coupler (Lyregaard, 1982), essentially the same differences as seen in Figure 5.2 are obtained whether the measurement is made in a 2-cc coupler or in a real ear.

Functional Gain and Insertion Gain

The *functional gain* of a hearing aid at a given frequency, expressed in dB, is the difference between the aided and unaided sound-field thresholds at that frequency. The *insertion gain* of a hearing aid at a given frequency, also expressed in dB, is the difference between the aided and the unaided eardrum SPL at that frequency.

Both functional gain and insertion gain are *real-ear* measurements that require the hearing aid wearer to be present and wearing the hearing aid. They differ in that functional gain is a *behavioral* measurement that requires the hearing aid wearer to be alive and responsive, whereas insertion gain is an *objective*, probe-microphone acoustic measurement that can be performed on a live subject or on a manikin such as KEMAR. (In the latter case, the measurement is of the aided-unaided SPL difference at the eardrum position in the manikin's ear simulator and is called *simulated insertion gain.*)

As simple examples, the *unaided* sound-field threshold of an individual at 500 Hz might be 50 dB HL. If the *aided* sound-field threshold is 25 dB HL at 500 Hz, the measured *functional gain* of the hearing aid would be 25 dB. A probe-microphone measurement

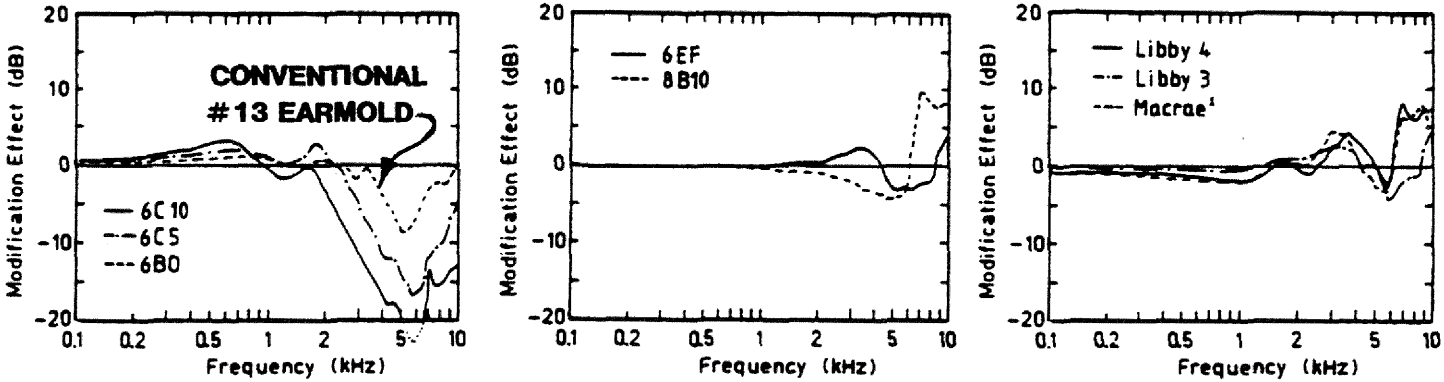

FIGURE 5.2. Difference, for a BTE hearing aid, between the 2-cc coupler SPL produced with common custom-earmold configurations and that produced by the 18 mm × 3 mm "horn" earmold simulation built into the HA-2 coupler. (Reprinted from Dillon, 1985, with permission of *Hearing Instruments* from Vol. 36, No. 12, 1985.)

at the unaided eardrum with the sound-field audiometer set to 50 dB HL should show an SPL of about 62 dB, since the minimum audible pressure *at the eardrum* at 500 Hz is 12 dB SPL. (Although it is beyond the scope of this chapter, the *eardrum* SPL corresponding to 0 dB HL is 12 dB at 500 Hz. This is true whether the stimulus is generated by an earphone, a hearing aid, or a sound field [Killion, 1978]. Table A-1, from ANSI S3.6-1989, gives the correspondence at other frequencies.) With the same 50 dB HL audiometer setting and the hearing aid in place, the probe-microphone measurement might show an SPL of, say, 89 dB, which would indicate an insertion gain of 27 dB, not much different from the 25 dB value for functional gain.

Recent studies have confirmed that, when a hearing aid is operating linearly and the measurements are properly performed, functional gain and the insertion-gain measurements yield equivalent answers to the question, "What is the real-ear gain of this hearing aid on that person?" Any differences will be the result of normal measurement variability or experimental error. (We include hearing aid noise and overload problems under "experimental error.") Indeed, one of the writers (Killion) once gave a paper under the intentionally humorous title "Insertion gain and functional gain: If they don't agree, you did it wrong!" (Killion, 1986).

The advantage of a careful insertion-gain measurement is that a complete response curve, with a standard deviation of 1 or 2 dB, can be generated in about 3 minutes (95 percent of the results will fall within ±2 to ±4 dB of the multiply-repeated measurement average) (Dillon and Murray, 1987; Killion and Revit, 1987; Trede, Morealli, Gudmundsen, and Killion, 1990). In contrast, a typical 7-frequency functional-gain measurement in the clinic requires a total of about 10 minutes to obtain the required unaided and aided sound-field audiograms and yields a standard deviation of about 5 dB under ideal conditions (95 percent of the measurements will fall within ±10 dB of the multiply-repeated average) (Byrne and Dillon, 1981; Killion, 1983) and a standard deviation closer to 7 dB under typical clinical conditions (Green, 1988). Because of measurement variability alone, therefore, the single-frequency difference between functional-gain and insertion-gain measurements will exceed 10.8 dB in 1 out of 20 (or "5 percent") measurements, even under ideal conditions; this difference can be even larger if functional gain and insertion gain are not measured in the same setting, taking care to move neither the hearing aid, the subject, nor the probe tube.

Finding such large individual differences between insertion gain and functional gain, some early investigators erroneously concluded that the two were in fact different, that they probably measured different things, and then argued that one or the other was a more realistic measure of real-ear gain. Fortunately, all recent careful experiments have shown that any differences are almost certainly due to experimental artifact or random variability (Zemplenyi, Dirks, and Gilman, 1985; Mason and Popelka, 1986; Dillon and Murray, 1987; Hawkins, Montgomery, Prosek, and Walden, 1987; Tecca and Woodford, 1987; Green, 1988).

The REAR, REUR, REIR, REIG, REOR Acronyms

A working group (ANSI S3.80) has been formed to standardize real-ear measurements. Their first semi-official output was a paper on terms (Schweitzer, Sullivan, Beck, and Cole, 1990), which are illustrated in Figure 5.3. In each case, the first two letters (RE) signify real-ear (as opposed to manikin or coupler) measurements. The last two stand for aided response (AR), unaided response (UR), insertion response (IR), insertion gain

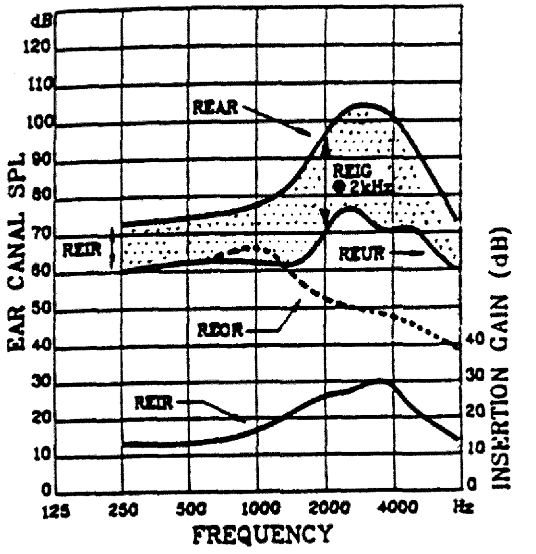

FIGURE 5.3. Illustration of real-ear measurement terms: REAR, REUR, REIR, REIG, and REOR. (Reprinted from Schweitzer et al., 1990, with permission of *Hearing Instruments* from Vol. 41, No. 2, 1990.)

(IG), and occluded response (OR). (*Gain* is a single-frequency measure, whereas *response* refers to the entire gain-versus-frequency characteristic.) The committee's hope is that with widespread adoption of the above acronyms, "A favorable improvement in interprofessional communication . . . will result" (Schweitzer, et al. 1990, p. 28)—presumably an improvement sufficient to compensate for their somewhat monotonous appearance and pronunciation. One clear advantage of these acronyms is that they permit the most important relation among real-ear measurements to be stated in a simple formula:

$$\text{REIR} = \text{REAR} - \text{REUR}$$

The real-ear insertion response is equal to the real-ear aided response minus the real-ear unaided response.

In the spirit of that working group's activities, we introduce an additional acronym: "2CCR," for *2-cc coupler response*. This acronym will be useful later in restating the two important results of this chapter as formulae.

CORFIG: The Insertion Response to Coupler Response Transformation

Although a real-ear measurement can give immediate information about the insertion response of a hearing aid once it has been delivered to the user, both the dispenser and the hearing aid manufacturer need a method of testing the hearing aid before its delivery to the end user. Since 1942, when Romanow described the design of a 2-cc coupler whose diameter-to-length ratio minimized problems with standing-wave modes up to about 8 kHz, the standard measurement of hearing aid performance for quality control purposes has been made using one or another version of that coupler.

In his paper, Romanow (1942) emphasized that the 2-cc coupler was not a real-ear simulator, but simply a convenient, easily fabricated coupler with which readily reproducible hearing aid measurements could be made. Romanow provided tentative correction curves, to be applied to the 2-cc coupler response curves for body-worn hearing aids, in order to estimate the "field-referenced response" (insertion response) of the hearing aid. A brief history of coupler corrections and real-ear measurements was included in the 1980 version of this chapter (Killion and Monser, 1980).

With the availability of the Zwislocki coupler real-ear simulator and the KEMAR manikin (Burkhard and Sachs, 1975), it became practical to obtain a *laboratory* measurement of the expected real-ear insertion response of hearing aids on a routine basis. Manikin measurements are invaluable for many engineering purposes, but mostly to define the *coupler* response of a hearing aid that will provide a desired real-ear insertion response. Once the coupler response target has been determined for the manikin in the sound field, the rest of the engineering development can take place using simple coupler measurements in a hearing aid test box.

The process of defining a coupler response for a particular insertion response can be broken down into a step-by-step, conceptual task. These steps take into account what is lost and what is gained when a hearing aid is taken off the coupler and inserted into the ear. All the factors that are lost when the hearing aid is inserted are added to the desired insertion response, while all the factors that are gained when the hearing aid is inserted are subtracted.

When the hearing aid is inserted, the unoccluded ear's natural amplifier, the REUR, is lost, so this factor is added into the desired coupler response. When the hearing aid is inserted, there is typically an increase in sound pressure level (the RECD) because the residual volume of the aided ear is typically smaller than the volume of the 2-cc coupler. So, this factor is subtracted from the desired coupler response. When the hearing aid is inserted, the microphone of the hearing aid is placed near the surface of the head (or body), giving an increase in sound pressure level, so we subtract this factor as well. Altogether, the desired coupler response can be obtained by adding to the desired insertion response the following transformation: **the REUR, minus the RECD, minus the effect of the placement of the hearing aid microphone.**

For example, in a case where the desired insertion response is a constant 0 dB insertion gain at all frequencies, then the desired coupler response is "CORFIG," **CO**upler **R**esponse for **F**lat **I**nsertion **G**ain. CORFIG is nothing more or less than the transformation at the end of the previous paragraph.

Note that all CORFIG curves in this chapter are "2-cc coupler CORFIG" curves except those specifically designated as "Zwislocki coupler CORFIG" curves, and all are based on average-subject data. In other words, unless otherwise noted, these curves describe the 2-cc coupler response curve required of a hearing aid if it is to produce a flat real-ear insertion response curve for the hypothetical "average subject."

It is clear that the CORFIG curves just described have applications more important than the unusual case in which a flat real-ear response is desired, because the target 2-cc coupler response for *any* desired real-ear insertion response can be obtained *by adding the CORFIG curve to the desired insertion-gain curve*. In terms of the acronyms presented earlier,

Target 2CCR = Target REIR + CORFIG.

Before finishing this chapter, we need to consider in detail the (mostly relatively small) differences that occur among the various average-subject CORFIG curves, depending on type of hearing aid, measurement condition and—in the case of BTE hearing aids measured on the HA-2 coupler—any differences between the actual custom earmold and the horn-earmold simulation built into the HA-2 coupler. But first, we pursue an example in which CORFIG is used for its intended purpose: improving the hearing aid fitting.

ORDERING THE 2-CC COUPLER RESPONSE

Transforming Target Insertion Response to Target 2-cc Coupler Response

Figure 5.4 shows the audiogram of a hypothetical subject, Mr. Youngmind, along with the target insertion response calculated according to the "⅓-gain rule," roughly similar to the latest National Acoustic Laboratories' (NAL-R) procedure (Byrne and Dillon, 1986) without the low-frequency corrections or the reserve gain allowance. This is only one of the methods for estimating a *reasonable* frequency response for a hearing aid whose frequency

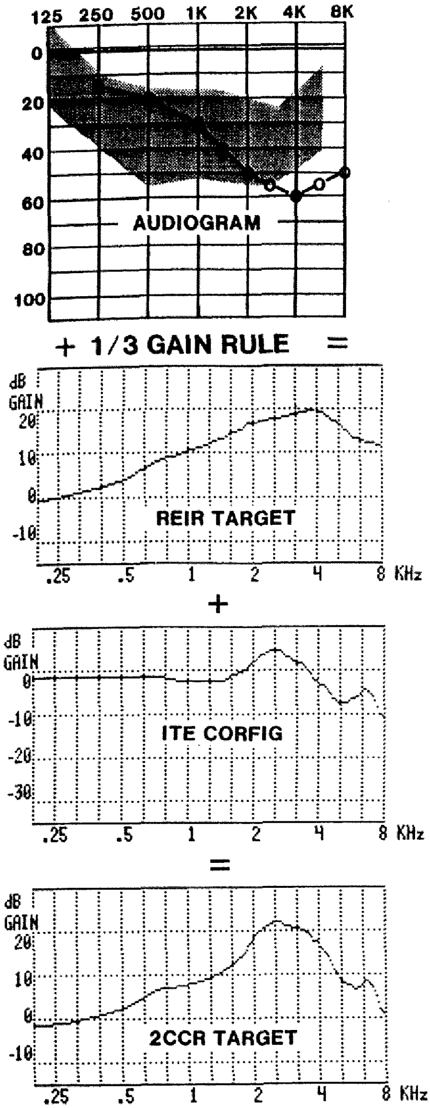

FIGURE 5.4. Estimation of target 2-cc coupler response: Desired insertion response plus CORFIG equals target 2-cc coupler response.

response does not change with input level; other methods are discussed elsewhere in this book. None of these methods attempts to restore aided thresholds to 0 dB HL, a procedure that would result in more gain and high-frequency emphasis than most adults will tolerate, but each attempts to reach a reasonable compromise between insufficient audibility and excess gain. What is important for the present purpose is that with the NAL-R and similar procedures, a change of 5 dB HL in the measured threshold audiogram will result in a change of only about 2 dB in the target insertion response. This means that minor errors in the original audiogram will not have much effect on the prescribed target response, which is fortunate because earphone-based audiograms sometimes provide imperfect estimates of the sound-field audiograms for the same listener.

Let us assume the insertion response (REIR) of Figure 5.4 has been chosen as the target for Mr. Youngmind's ITE hearing aid. The hearing aid manufacturer, not having Mr. Youngmind's ear available, must test the hearing aid in a test box with a 2-cc coupler. In order to have a reasonable chance of achieving the target REIR when the hearing aid is delivered, i.e., in order for Mr. Youngmind to *experience* the insertion response chosen as the target, the manufacturer needs to know the 2-cc coupler equivalent of the target REIR curve. Either the manufacturer or the dispenser can calculate the 2-cc coupler target response curve by adding the average-subject ITE CORFIG to the target REIR, as illustrated in Figure 5.4. If the 2-cc coupler response of the resulting ITE aid matches the target 2-cc response of Figure 5.4, and if Mr. Youngmind's ear acoustics are close to average, then the insertion response he experiences will be close to the original target insertion response.

Many dispensers simply send in the audiogram and ear impression and let the manufacturer do everything else. Others, however, choose to generate a target insertion response and calculate the target 2-cc response themselves, sending only the target 2-cc response along with the impression to the manufacturer. The labor of those calculations is now much simpler for those who choose to do it themselves, as more than one probe-system manufacturer includes software for (a) calculating the target insertion response curves from the entered audiogram (using NAL, ½-gain, or other rule) and (b) immediately plotting the corresponding target 2-cc response using *either* the pre-programmed average-subject CORFIG data *or* the individual-subject CORFIG calculated automatically from the previously measured unaided response and real-ear-to-coupler level differences for that individual (Punch, Chi, and Patterson, 1990; Revit, 1990).

Individual Differences

A nagging problem for both the manufacturer and the dispenser has been that even when the 2-cc coupler response of the hearing aid matches the target 2-cc response almost exactly, the *measured* insertion response (REIR) may not match the *desired* (target) REIR. This can happen for several reasons. First, although the effect of both deliberate and inadvertent venting typically dominates the insertion response at low frequencies, the 2-cc coupler response of the hearing aid is almost always measured with the vent blocked. Second, an unusual external ear and/or eardrum can cause the unaided response (REUR), the aided response (REAR), or both to be substantially different from these responses in the average ear. A hearing aid on that unusual ear will produce an insertion response that is unusual, substantially different from what one would expect based on average data. Third, measurement error—both random and not-so-random—often causes an *apparently* significant difference that disappears with repeated measurements or is peculiar to the measurement setup. These are discussed next.

Venting. A first-order estimate of the real-ear effect of a given vent channel can be obtained from the well-known vent-response tables of Lybarger (1985). This estimate can be included as a correction to the target 2-cc response, so that the delivered real-ear insertion response will be closer to the target insertion response. But the correction is tricky and depends on the expected gain setting of the hearing aid. To explain, even with the hearing aid turned off, the sound entering directly into the ear through the vent will provide 0 dB "gain" at low frequencies so that when the calculated vent-response-corrected gain of the hearing aid is less than 0 dB it is ignored: The hearing aid no longer controls the insertion response at those frequencies. Dillon (1991) gives an excellent treatment of this problem.

A more serious problem in the measurement of the aided response is the effect of the probe tubing, which can cause an appreciable leak that disappears when the probe tubing is removed. It is not uncommon to see published papers demonstrating that "the size of the vent channel or vent plug has almost no effect on the measured insertion response," with graphs indicating that *all* of the measured response curves were probably dominated by an inadvertent leak caused by a large-diameter probe tube lying between the ear canal wall and the hard shell of the hearing aid. The reason is easily seen by analogy: If there are two holes in the bottom of a bucket, one large and one small, it is hard to notice any difference in the rate of water loss when the smaller hole is blocked. Indeed, some hearing aid shells fit so loosely that, even without a probe tube in place, the dominant venting action is from the slit leak around the shell rather than from the nominal vent channel. Similarly, the 5 to 10 dB peak that is often introduced by the Helmholtz resonance between the vent channel and the ear canal volume is frequently damped by the resistance in the slit leak caused by the presence of the probe tube.

Yet another real-ear effect of venting is the tendency for sound emanating back out of the ear canal through a vent channel to raise and sharpen resonant peaks in the aided response, even to the point of audible feedback (Cox, 1982). Unfortunately, the extent of these effects cannot be predicted in advance unless the precise dimensions and impedance of the ear are known (Kates, 1988). This is one of the reasons why standard 2-cc coupler measurements are made with vent channels plugged.

Because of these and other potential effects of venting described earlier, it should be clear to the reader that any time a vented hearing aid is ordered by means of a target 2-cc coupler response, the individual fit of the delivered instrument must be confirmed by a real-ear measurement. But caution must be observed to avoid extraneous venting effects introduced by the presence of the probe tube.

Unaided Response. The unaided response of the average ear shows a primary peak at 2.7 kHz of 15 to 20 dB and a secondary peak at 4 to 5 kHz of 10 to 15 dB, depending on measurement conditions. The peak at 2.7 kHz is determined by the principal quarter-wave resonance of the distributed "horn" formed by the pinna, concha, and ear canal, the combined effective length of which, including end effects, amounts to about 32 mm. The magnitude of the peak is influenced somewhat by the efficiency of the horn action, but is mostly controlled by the damping introduced at the eardrum (through the ossicular coupling) by the cochlea, which is normally resistive at that frequency.

Frequency. An unusually large pinna, concha, and ear canal can result in an unaided response curve that peaks close to 2 kHz, rather than the average-ear 2.7 kHz value; the opposite can happen with a small external ear, which may exhibit a peak near 3.5 kHz.

Magnitude. A sharp peak height of 30 dB or more can occur with a "concrete eardrum" (a psuedo-medical term introduced here), where the mechanical input resistance from the cochlea provides no resonance damping, the eardrum impedance being dominated by a stiffening disease (tympanosclerosis or otosclerosis, for example). At the other extreme, a peak height of only 5 to 10 dB or less can occur when a flaccid, high-loss eardrum effectively damps the peak. Ear wax or a badly collapsed ear canal can also result in a drastic reduction of peak height. In the extreme case of the badly collapsed ear canal, the unaided response exhibits only a high-frequency rolloff; no peak at all can be seen.

Ear (Load) Impedance. Another consequence of an unusual middle ear impedance is that the *output* of the hearing aid may be significantly higher or lower than would be expected from average data (Gilman, Dirks, and Stern, 1981; Kates 1988).

Figure 5.5 shows the measured ear canal SPL produced by an insert earphone, compared to its 2-cc coupler output, for several individual ears, some common and some unusual. The acronym RECD (real-ear-to-coupler level difference) describing the type of curves shown in Figure 5.5 was introduced earlier. The average-ear RECD is also shown in Figure 5.5 for comparison. (Note: The 1972 Sachs and Burkhard data shown in Figure 5.1 had been corrected for the estimated difference between eardrum and ear canal SPL. The data in Figure 5.5 are uncorrected because ear *canal* SPL rather than ear*drum* SPL is what is of interest here: Most probe placements result in a tip location 5 to 10 mm back from the eardrum.)

Curve a in Figure 5.5 appears to belong to an ear with tympanosclerosis ("concrete eardrum"); curve e appears to belong to an ear with a flaccid eardrum, such as is sometimes found with a healed perforation; curve f *appears* to belong to an ear with a perforated eardrum, but checking for leaks between the eartip and ear canal during measurement would be a good idea before sending the person out for medical treatment. The real-ear aided response of a hearing aid whose target response was chosen on the basis of average-ear data would be much higher on the ear of curve a and much lower on the ear of curve e. More importantly, neither difference is a frequency-independent level shift, which could be accommodated by changing the volume control setting. Either will produce a substantial tilt in the aided frequency response curve.

The curves in Figure 5.5 are of particular interest because they can be readily measured by anyone with a probe measurement setup. These curves were obtained on a commercially available probe system by Fikret-Pasa and Revit (1992), using a technique similar to that used by Sachs and Burkhard (1972), who recorded the output of an insert earphone in a 2-cc coupler and then in real ears. Figure 5.6 shows the measurement configuration. A probe tube was threaded through the foam eartip of an insert earphone (Figure 5.6a), which was then sealed onto the surface of an HA-1 2-cc coupler as though the earphone's eartip were an earmold or hearing aid shell (Figure 5.6b). The 2-cc coupler response of the earphone was measured and stored as a calibration curve. The real-ear response of the earphone was then measured by placing the probe-tube/eartip combination into the subject's ear, with the eartip at a depth anticipated to be that of a typical hearing aid (12 mm) and the probe tube extending approximately 5 mm beyond (Figure 5.6c). The curves of Figure 5.5 show the real-ear earphone response minus the 2-cc coupler earphone response (i.e., the RECD response) for several subjects.

Cox (1983), Hawkins (1984), and Libby (1985) reported similar measurements with an insert receiver or insert earphone for the purpose of selecting the maximum 2-cc coupler output a hearing aid should have in order not to exceed the patient's discomfort level. Similar measurements can also be made using a hearing aid with a direct audio input. Punch et al. (1990) report the use of sound-field measurements with a BTE hearing aid for similar purposes, although the latter method introduces the additional variability inherent to sound field measurement.

In general, real-ear-to-coupler level difference measurements can be quite accurate up to about 4 kHz. At higher frequencies, there is a good chance of error due to a quarter-wave resonance minimum in the SPL at the probe tube inlet.

Modifying the Target 2-cc Coupler Response

In cases where the unaided response is substantially different from that of the average ear, Mueller (1989) suggests that the average-CORFIG-based target 2-cc response can be adjusted by correcting for the difference between the *individual* unaided response and the *average* unaided response. This may make good sense in cases of ears that are unusually large or small, whose resonance frequencies are lower or higher than normal. Matching the hearing aid peak frequency to the real-ear peak frequency will give a smoother insertion response, which in turn should minimize the time required for the user to adjust to the hearing aid so that maximum benefit may be obtained.

Completely modifying the target 2-cc response makes less sense in the case of the concrete eardrum or the collapsed canal. Re-

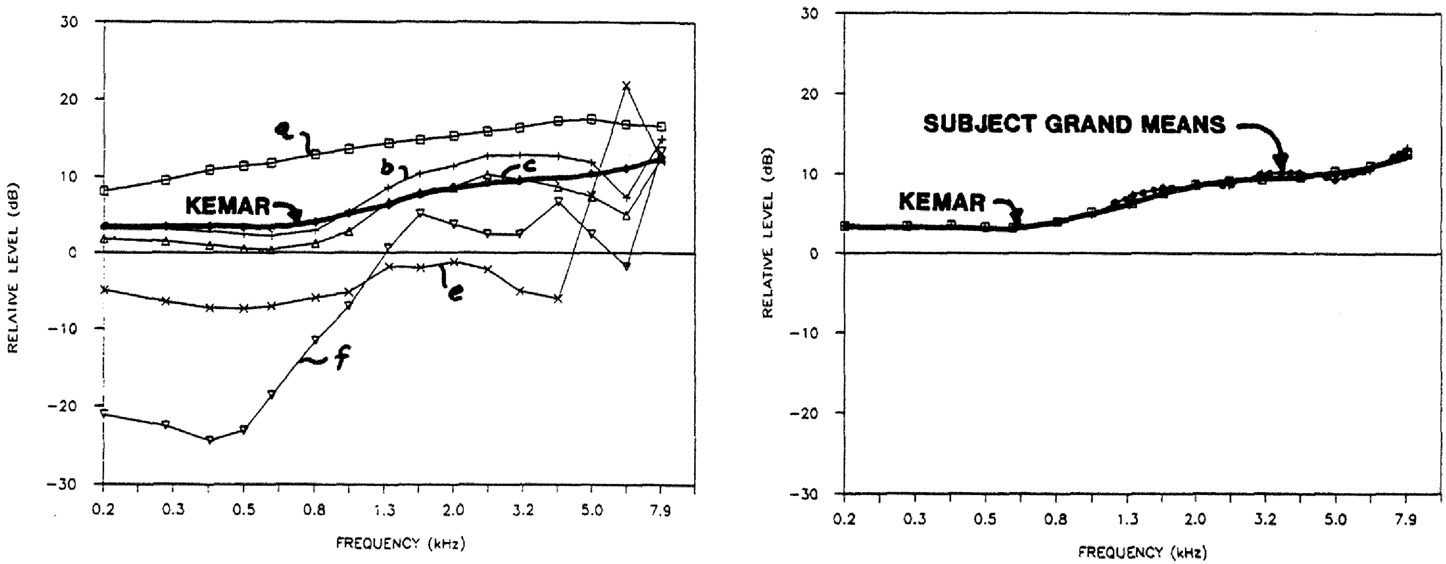

FIGURE 5.5. Examples of ear canal SPL developed in real ears by an insert earphone, normalized to the 2-cc coupler SPL developed by the same earphone.

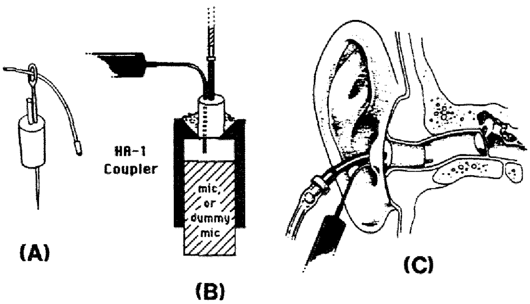

FIGURE 5.6. Use of insert earphone to determine real-ear-to-coupler level differences (RECD): (a) Threading probe tubing through foam eartip; (b) Measuring earphone output in 2-cc coupler; (c) Measuring earphone output in real ear. (Modified drawing of ear reproduced with permission from Elliot Berger, E-A-R Corporation.)

producing a 30 dB unaided response peak in the aided response would indeed produce a smoother insertion gain curve (and might at first sound "more natural" to the first-time hearing aid wearer with this type of ear), but common sense warns that such a pathological ear receives little benefit from that sort of spectral shaping. Similarly, the usual solution to a collapsed ear canal is to fit the individual with a tube or hollowed-out plastic shell, which is sometimes all the "hearing aid" that is required. Reproducing a depressed unaided response peak in the aided response would deprive that ear of the spectral shaping that was part of its pre-collapse design. (Interestingly enough, the normal speech spectrum *measured at the eardrum* is almost *flat* up to 3 to 4 kHz because of the 15 to 20 dB boost supplied by the normal external ear resonances [Bergenstoff, 1981]).

We believe that it is *not* a good idea to incorporate large corrections for individual peak magnitudes into the target 2-cc response. Some dispenser or manufacturer intervention in any automated procedure for generating the target 2-cc response seems called for in such cases.

Analogous with Mueller's suggestion for correcting for an individual's real-ear unaided response into the target 2-cc coupler response, a correction for the individual's real-ear-to-coupler level difference could also be incorporated into the target 2-cc coupler response. In theory, the result would be a perfect fit every time if the manufacturer were able to duplicate the target 2-cc coupler response; the measured insertion response on the delivered aid would duplicate the original insertion response target response exactly. At least two probe-equipment manufacturers permit exactly those corrections to be incorporated into the target 2-cc response printout, based on simple probe measurements that can be made before the hearing aid is ordered (Punch et al, 1990; Revit, 1990).

Measurement Error Pitfall. A problem facing both manufacturers and dispensers is that of measurement error and variability. Improper probe tube placement, leakage around the probe tube, probe tube movement between the unaided and aided response measurement, movement of the subject between the two measurements, movement of the *tester* between the two measurements, and severe reflections or noise in the test space can all cause appreciable error in the measurement of hearing aid insertion response. Some of these are discussed in more detail later, but the present problem is twofold: (a) deciding how much of the individual-ear correction should be incorporated into the target 2-cc response, and (b) deciding when a hearing aid should be sent back to the manufacturer for modification.

An examination of test-retest variability from several experiments indicates that, with great care, it is possible to obtain a standard deviation for a single measurement of 1 to 2 dB at low frequencies. Above 4 kHz, extreme care plus good equipment and a reasonable size room are required to keep the standard deviation below 3 to 5 dB. This is unfortunate, because the frequency region above 4 kHz is all too often ignored while, in the opinion of the present writers, that region is one of the important regions for optimizing a hearing aid fitting. Given the present state of the art, however, the writers are forced to recommend that *individual-ear corrections above 4 kHz be restricted* to a smooth extension of any curve corrections below 4 kHz. (In other words, the target curve above 4 kHz should connect smoothly to the corrected target curve below 4 kHz.) There are so many sources of error above 4 kHz that the dispenser is as likely to cause a problem as to solve one by attempting any other corrections.

The ½-Correction Suggestion: A Statistical Compromise. At 4 kHz and below, where careful measurements are on reasonably firm ground but some error is inevitable, one attractive compromise between no individual-ear correction and full correction would be to use only ½ of the correction. In other words, instead of correcting the average-ear CORFIG by the full difference between individual-ear and average-ear REUR or RECD, correct by only ½ that difference. This compromise was initially suggested by a common finding in all experiments, a finding known as "regression to the mean": The farther from the average a given result is, the more likely a repeat measurement will be closer to the average. The use of a "½-correction rule" reduces the likelihood of making a serious over-correction caused by measurement error. It also gives some weight to nature's design for the spectral shaping of the ear as exhibited in average data, without ignoring the fact that ears are sometimes dramatically different.

There is further rationale for a ½-correction compromise, for the REUR correction at least. After extended arguments too long to be reproduced here, the writers conclude the following: If the hearing loss is measured in a *sound field*, thus taking into account the individual's REUR, a "½-gain" fitting formula would compensate for any individual-versus-average REUR difference by prescribing a gain change equal to half that difference. To achieve the equivalent result starting with *earphone* thresholds (especially insert-earphone thresholds), the individual-versus-average REUR difference should be multiplied by the same ½ rule before it is added as a correction to the target 2-cc coupler response.

However reasonable it may be, no arbitrary rule can or should take the place of good clinical judgment. In this, as in many of the decisions that must go into hearing aid selection and fitting, a good guess as to how much change in "sound" the candidate is likely to tolerate while adjusting to the hearing aid can help avoid the "dresser-drawer solution" to the fitting problem.

GIFROC: THE COUPLER RESPONSE TO INSERTION RESPONSE TRANSFORMATION

It should be readily apparent that if *adding* a CORFIG to the target real-ear insertion response results in the target 2-cc coupler response required of the hearing aid, then *subtracting* a CORFIG from the 2-cc coupler response of an aid will result in an estimate of the real-ear insertion response for that aid. Although perhaps it *should* be readily apparent, years of experience teaching some very bright students indicate it usually *isn't*. To make matters worse, some writers have used the term "CORFIG" to describe the upside-down inverse curve.

In the hope of reducing the confusion, we introduce the name for the inverse of CORFIG: "GIFROC" (CORFIG spelled backwards). The GIFROC curve is the mirror image of the CORFIG curve around the 0 dB axis. Numerically, the GIFROC value at any frequency is the exact negative of the CORFIG value, so that *adding the GIFROC* value is equivalent to *subtracting the CORFIG* value. This definition leads to what is perhaps the most important statement of this chapter: *An estimate of the insertion response of a hearing aid whose 2-cc coupler response is known is obtained by adding the GIFROC curve to the 2-cc coupler response.* Or, using the acronyms,

Estimated REIR = 2CCR + GIFROC

Figure 5.7 illustrates applying a GIFROC to estimate the real-ear insertion response for a full concha ITE hearing aid whose 2-cc coupler response has been obtained in a test box. The real-ear insertion response of a *BTE* hearing aid can be estimated from the manufacturer's published response by adding the BTE GIFROC *and* correcting for any differences in earmold horn effect between the published HA-2 simulated-horn-earmold curve and the expected earmold response. (See Figure 5.2.)

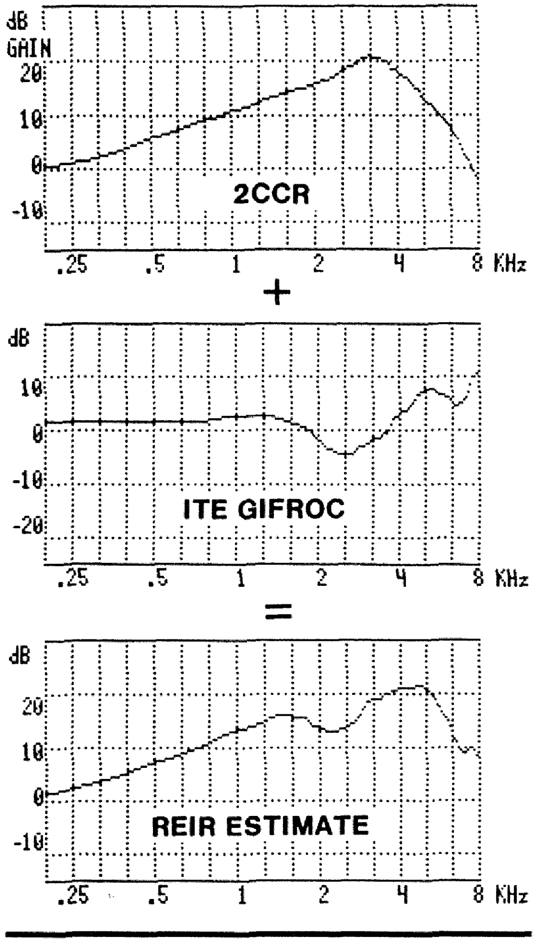

FIGURE 5.7. Estimation of real-ear insertion response: 2-cc coupler response plus GIFROC equals REIR estimate. Curves shown for full-concha ITE aid.

AVERAGE-EAR CORFIG AND GIFROC DATA

In the beginning, most real-ear measurements were made at 0° incidence in conditions as close to anechoic as could be obtained. The problem is that we seldom listen under anything remotely approaching anechoic conditions. Under most listening conditions, we are presented with a mixture of direct and reflected sound. Even in close face-to-face conversations, 10 to 20 percent of the energy arriving at the listener's ears is typically reflected energy. In group settings or when listening to music, the sound sources are generally located much farther away from the listener, so the majority of the energy arriving at the listener's ear is reflected energy. With a TV set in a typical living room, for example, the curve given by Olson (1967, p. 285) indicates that 90 percent of the energy arriving at the listener's ear can be reflected energy.

Diffuse-Sound-Field Data

Figure 5.8 shows the average-ear CORFIG and GIFROC curves for various types of hearing aids. The curves in Figure 5.8 are based on diffuse-sound-field measurements of the SPL at the eardrum on 16 ears (8 males

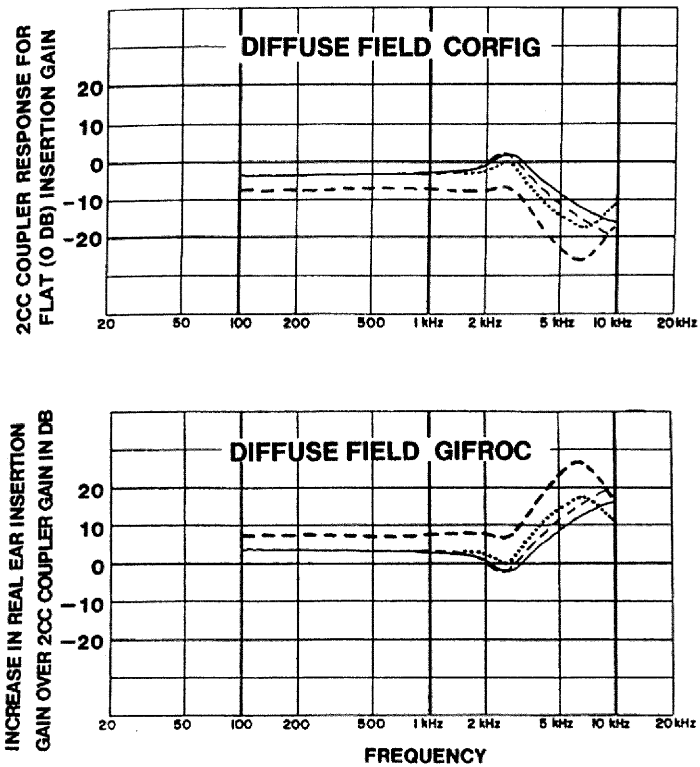

FIGURE 5.8. 2-cc coupler diffuse-field CORFIG and GIFROC curves for various types of hearing aids: BTE (solid curve); full-concha ITE (dashed curve); small ITC (dotted curve); deep, extremely long canal-tip ITC (heavy dashed curve).

and 8 females) and diffuse-sound-field measurements of the SPL on the KEMAR manikin at the microphone position for BTE, ITE, and canal hearing aids (Killion, Berger, and Nuss, 1987).

By definition, a "diffuse" or "random incidence" sound field is one in which sound arrives with approximately equal probability from all directions. In theory, this condition could be realized in an anechoic chamber with many independent loudspeakers located around the listener. In practice, a reverberation chamber is normally used to produce a diffuse sound field, with the multiple reflections from the walls, floor, and ceiling providing the required diffusion. The data of Figure 5.8 were obtained in the large reverberation chamber at the E-A-R facilities in Indianapolis, Indiana, for example. Although usually performed in a research laboratory setting, such diffuse-field measurements probably come closer to representing the real-world situation than any other measurement. They thus provide basic engineering data to guide hearing aid design.

0-, 45-, and 90-Degree Sound-Field Data

Fortunately, the measured real-ear insertion response is relatively independent of the

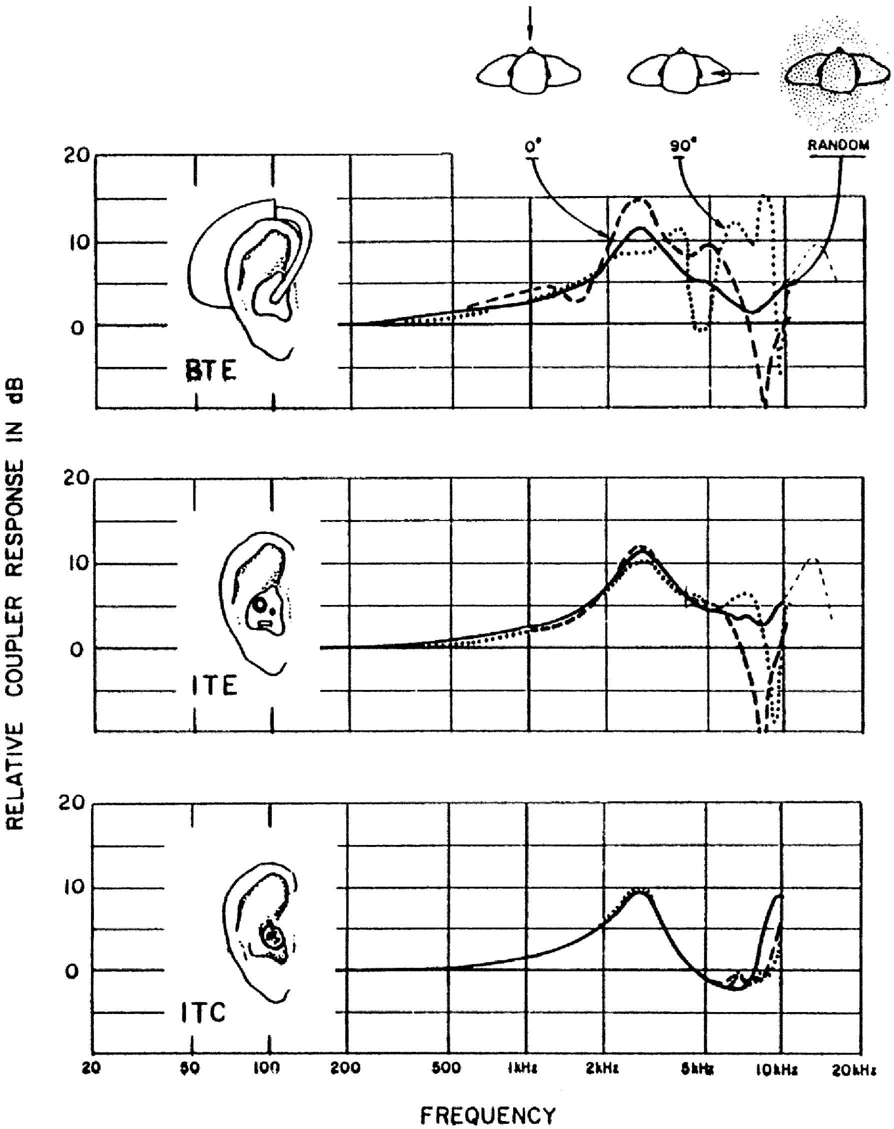

FIGURE 5.9. Zwislocki coupler CORFIG curves illustrating effect of hearing aid type and sound-field conditions. (Reprinted with permission from Killion and Monser, 1980.)

sound-field conditions. This is almost exactly true for canal aids at all frequencies, and it holds true for ITE hearing aids up to 4 to 5 kHz, and for BTE hearing aids up to 3 to 4 kHz. Figure 5.9 shows the Zwislocki coupler CORFIGs for BTE, ITE, and canal (ITC) aids obtained under three sound-field conditions in an anechoic chamber.

In clinical practice, probe measurements are typically made in relatively small rooms

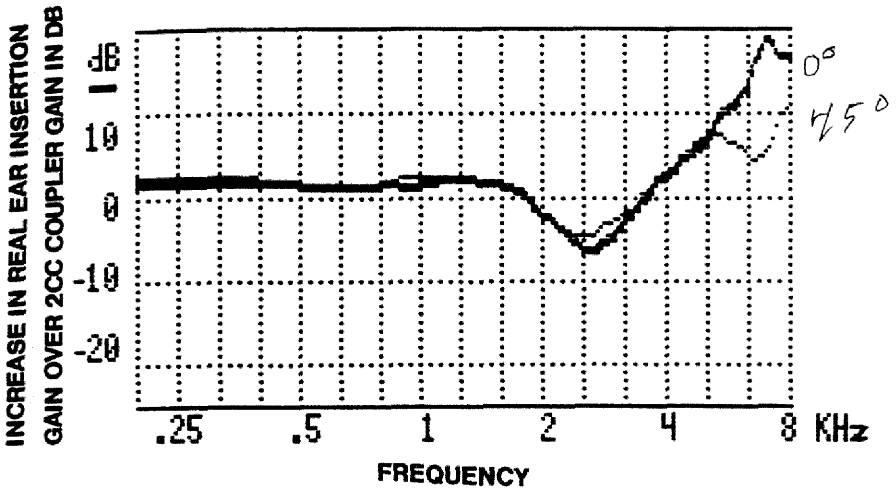

FIGURE 5.10. Comparison between 0° incidence and 45° incidence 2-cc coupler GIFROC curves for a BTE hearing aid.

with the loudspeaker located either directly in front of the listener (0° incidence) or toward the ear under test (45° and 90° incidence). There is a problem with locating the loudspeaker at 0°: The normal concha anti-resonance causes a severe dip in the unaided response curve in the 6 to 8 kHz range and—because the unaided curve is subtracted from the aided curve to obtain insertion gain—causes an artificial *peak* in the insertion-gain curve of BTE and ITE hearing aids because their microphones do not experience the concha anti-resonance.

Figure 5.10 shows a comparison between the 0° incidence GIFROC obtained by Burnett and Beck (1987) on 10 real ITEs measured in an anechoic chamber (they labeled it the "2-cc coupler to insertion response correction") and the 45° incidence GIFROC obtained by Revit (1990). Note first the excellent agreement at 5 kHz and below, and second, the peak in the 0° GIFROC curve at 7 kHz. (To see the same peak, mentally invert the 0° ITE CORFIG curve of Figure 5.9 to obtain the GIFROC curve.)

Table 5.1 shows average CORFIG and GIFROC values for BTE, ITE, and canal aids. These values are suitable for use with a probe microphone measurement system that uses an over-the-ear reference microphone and a loudspeaker at 45°. For the reasons discussed earlier, the BTE values above 3 kHz and the ITE values above 4 kHz should be viewed with caution. The GIFROC values in Table 5.1 are similar to, but not identical with, those given by Lybarger and Teder (1986) in their Table 1. The Lybarger and Teder values were obtained from sound-field measurements where no reference microphone was used.

Minimizing Measurement Variability

One problem with 0° incidence measurements is that they appear to produce the largest measurement variability. Locating the loudspeaker at 45° reduces the effect of inadvertent head motion and room reflections. Figure 5.11 shows the effect of measurement angle on the *variability* of the measured insertion response of a single BTE hearing aid

TABLE 5.1. CORFIG and GIFROC data; 45° speaker azimuth, reference microphone over the ear

Frequency (kHz)	*BTE*		*ITE*		*Canal*	
	CORFIG	*GIFROC*	*CORFIG*	*GIFROC*	*CORFIG*	*GIFROC*
0.2	−1.7	1.7	−1.1	1.1	−2.0	2.0
0.3	−1.2	1.2	−1.4	1.4	−1.7	1.7
0.4	−0.7	0.7	−1.3	1.3	−1.6	1.6
0.5	−0.3	0.3	−1.4	1.4	−1.4	1.4
0.6	−0.2	0.2	−1.5	1.5	−1.4	1.4
0.8	−0.6	0.6	−1.6	1.6	−1.8	1.8
1.0	−1.7	1.7	−2.4	2.4	−2.6	2.6
1.3	−2.0	2.0	−2.3	2.3	−2.1	2.1
1.6	−0.5	0.5	−1.5	1.5	−1.6	1.6
2.0	4.0	−4.0	1.5	−1.5	0.7	−0.7
2.5	6.0	−6.0	4.5	−4.5	2.9	−2.9
3.1	4.3	−4.3	2.3	−2.3	0.8	−0.8
4.0	2.8	−2.8	−2.4	2.4	−6.5	6.5
5.0	0.0	0.0	−7.2	7.2	−9.1	9.1
6.3	−1.0	1.0	−5.4	5.4	−9.3	9.3
8.0	−8.3	8.3	−10.4	10.4	−13.1	13.1

Source: Courtesy, Frye Electronics, Inc., Tigard, OR.

measured 5 times on each of 10 subjects (Killion and Revit, 1987). The measurements were performed in a 10' by 9' by 7.5' double-walled IAC booth with the loudspeakers located 18" from the subject's ear.

It is noteworthy that the variability in this study was less than previously reported (although Hawkins et al., 1989, and Humes, Hipskind, and Block, 1988, have since reported similar data). In addition to taking great care in locating the probe tube deeply in the ear for each subject, two other precautions may have contributed to the excellent repeatability in this study. First, each subject was asked to sight across the tip of his or her nose, selecting a spot on the wall for each eye's line of sight, and then to maintain that head location throughout the experiment. Second, the experimenter was in a separate room during the actual measurements. As a result, the disturbing effect of *subject* head motion and *tester* body motion were minimized.

With larger loudspeaker-to-subject spacings, reflections can introduce substantial variability in real-world probe measurements. This was confirmed in informal preliminary experiments that used a loudspeaker-to-ear spacing of 30 inches (Trede et al., 1990). Measured differences of several dB in canal aid insertion response between 0° and 45° loudspeaker orientations were *reversed* by changing the test location in the room or by bringing reflecting panels within a few feet of the test subject. This was true even though the test system employed a reference microphone located directly under the ear to normalize the incoming sound pressure. With large loudspeaker-to-ear distances, it is almost as important for the *tester* not to move between unaided and aided response measurements as for the subject. Lack of attention to tester movement may explain

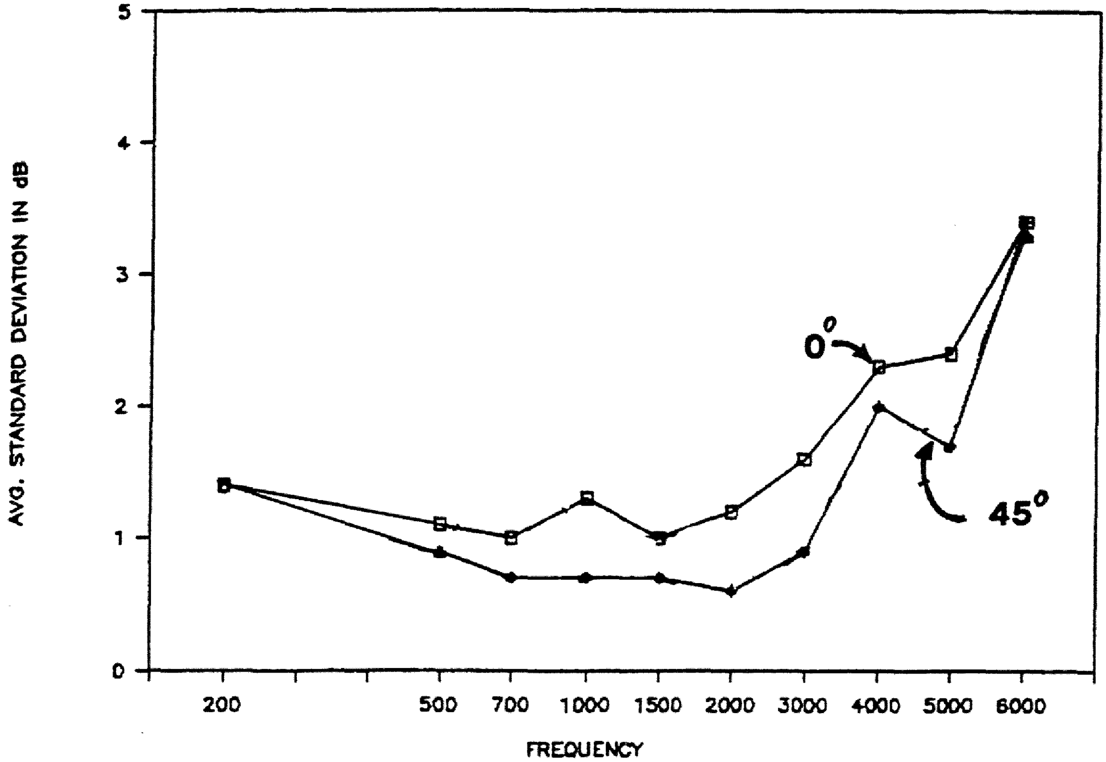

FIGURE 5.11. Variability of measured insertion response from test-retest experiments with one BTE aid on 10 subjects using two loudspeaker orientations: 0° and 45°. (Modified with permission from Killion and Revit, 1987, © by Williams & Wilkins, 1987.)

some of the large test-retest variability often reported in insertion response measurements. We now recommend a loudspeaker-to-ear spacing of 12 inches for routine clinical measurements and never more than 18 inches. A loudspeaker orientation of 45° is also recommended except when measuring CROS-type hearing aids, where 0° is obviously required.

An important exception to the 45° recommendation occurs when an estimate of the REIR above 4 kHz is desired. In that case, informal observation shows that averaging measurements of REIR at 0° and 45° appears to provide a reasonable estimate of the diffuse-field REIR and thus the "real world" response of the hearing aid above 4 kHz. Taken singly, the 0° measurement tends to overestimate the high-frequency gain while the 45° measurement tends to underestimate it.

The requirements for reliable real-ear measurements are as follows:

1. The probe tube is deeply placed in the ear canal.
2. Nothing moves between unaided and aided testing—not (a) the probe nor (b) the test subject, nor (c) the tester, nor (d) anyone or anything else in the test environment.

3. The sound field is free from extraneous noises, including those made by the subject.
4. The tester occasionally verifies that essentially the same REIR can be obtained twice in a row.

THE UNANSWERED QUESTIONS

It seems clear that one can now *measure* the insertion response of a hearing aid with much greater accuracy than an aid can typically be manufactured to match a given target response. On a group of 119 ears fitted with custom ITE hearing aids and adjusted to equal the target response at 2 kHz, Mueller (1990) showed deviations from the desired target response at 4 kHz exceeding 10 dB in 29 percent of the fittings.

What is unclear is just how much deviation is acceptable and over what frequency range. If a hearing aid with a response peak at 3500 Hz is placed on an ear whose unaided response peaks at 2200 Hz, the real-ear insertion response will show a dip at 2200 Hz and a 5 to 10 dB peak at 3500 Hz. Is that worth fixing? What if fixing that problem by shifting the hearing aid earphone resonance downward introduces a general high-frequency rolloff above 4 kHz? Which is the more important problem? These and other questions are now addressable *because* reliable real ear measurements are available.

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