# Chapter Five

# Prescriptive Procedures

*by Robert M. Traynor, EdD*

**Robert M. Traynor, EdD, is Clinical Professor of Audiology at the University of Colorado in Greeley, Colorado.**

Since its inception, Audiology has been concerned with hearing aid selection and the interaction of the hearing impaired person and amplification. The classic work of Watson and Knudsen (1) and Lybarger (2) suggests a long clinical interest in prescribing parameters for hearing aids according to audiometric measurements. In the early days, prescriptive procedures were neglected due to technologic limitations and viewpoints of that time. Teter (3) suggests that the combination of the Harvard Report (4) and the comparative hearing aid evaluation approach (5) led hearing aid selection away from prescriptive technology. It has only been in the last decade that we have returned to the prescriptive approach. The development of sophisticated in-the-ear (ITE) hearing aids and the probe microphone have facilitated the renaissance of prescriptive fitting techniques.

## PREPARATION

Clinically, prescriptive hearing aid fitting should begin with incorporating historical, situational, and audiometric information into the overall diagnosis and treatment of the particular hearing impairment. Information regarding the person's impairment typically suggests the type, style, circuit, programmability, and other general parameters of the instrument that will facilitate a good prescriptive hearing aid fitting. Once the person's data are obtained, the clinician must be cautious in coupling the appropriate prescriptive fitting technique with the specific type hearing loss and hearing aid circuitry. Except for some recent approaches presented in this chapter, basic research fundamental to prescriptive hearing aid selection has been conducted only with linear hearing instruments. New procedures, specifically offered for nonlinear hearing aid circuits, include the DSL I/O (6), IHAFF (7), and the FIG6 (8). The appropriate choice of the algorithm may determine the success of the prescriptive hearing aid fitting. Sullivan (9) suggests that there is a significant interaction effect among prescriptive fitting techniques and persons with different hearing loss characteristics. The challenge is in choosing the right prescriptive fitting technique or modifying it so it becomes the right approach for a given client.

### Audiologic Evaluation

It is essential that the client is given a comprehensive audiologic evaluation that includes pure tone air and bone conduction audiometry, speech audiometry, immittance audiometry, speech at its most comfortable level (MCL), and speech at its uncomfortable loudness level (UCL). While most prescriptive fitting procedures require only pure tones to facilitate their calculations, some approaches will require warble tone loudness growth measurements in high and low bands. Also, some proprietary programmable systems require special equipment and specific measurement protocols as part of their fitting. Though not essential, it is reasonable to assess the

Address all correspondence to: Robert M. Traynor, EdD, Audiology Associates of Greeley, 2528 16th St., #100, Greeley, CO 80631.

interoctave frequencies, particularly 1500 and 3000 Hz. Not only do some prescriptive formulae calculate gain and output at these interoctave frequencies, but the hearing impairment could change somewhat over a full octave, creating a less than adequate fitting that might require modification of the hearing aids performance parameters in these interoctave areas. In addition, no matter what fitting approach is utilized, it appears that there should be strong clinical contraindications to a binaural fitting before clients are directed to the monaural use of hearing aids (10,11). Auditory deprivation is one issue, but the other is how much better people do with two hearing aids.

## PRESCRIPTIVE PROCEDURE

### Where Do These Prescriptive Formulae Come From?

According to Hawkins (12) approaches to the specification of gain and frequency response can be categorized in several ways. Researchers decide systematically what they want the hearing aid to do and why; then specific formulae are developed to achieve the stated goal. Usually, these formulae are based upon research that supports a particular approach to the hearing aid fitting question. Cornelisse et al., discuss prescription as a frequency-specific gain function that can be prescribed for each individual with hearing impairment on the basis of audiometric data (6). These gain algorithms can be subdivided into two classes: formulae that make use of threshold audiometric data and formulae that incorporate suprathreshold audiometric data in deriving electroacoustic prescriptions for listeners with hearing impairment. While Byrne (13) states that gain and frequency response requirements appear to be about equally predictable from threshold or suprathreshold measures, particular formulae may or may not accurately reflect specific relationships between the audiological measure and the amplification parameter. The specifics of prescriptive formula generation are beyond the scope of this presentation, but detailed discussion can be found in Humes and Halling (14) and McCandless (15).

### What Will A Prescriptive Procedure Do For My Clients?

All prescriptive approaches have some common objectives (16):

1. to provide gain appropriate to achieve functional threshold shifts toward "normal" hearing
2. to present average speech spectrum at a comfortable level to the ear
3. to provide maximum dynamic range
4. to provide signals that restore equal loudness function
5. to provide aided speech signals at the MCLs in the speech frequencies
6. to provide gain based on the size and shape of the dynamic range
7. to provide gain based upon the discomfort level.

In a busy clinic, it is difficult to "eyeball" a hearing loss and consider all of the above variables equally. Prescriptive technique allows for an organized, systematic approach to hearing aid fitting. Further, if these approaches are not utilized when aids are ordered, the manufacturer often makes the decision regarding the specific hearing aid circuit based upon their average data, or hearing instruments built for similar hearing losses that were not returned. Obviously, information that is average data will not usually apply to a specific person, and instruments not returned for credit may be instruments that are not being utilized. Prescriptive approaches are important in that they bring to the clinician a scientifically based procedure that maximizes user benefit in a short amount of time.

*Which Formulae Should I Use?*

A detailed discussion of the advantages and limitations of the various popular and obscure prescriptive fitting approaches can be obtained from various sources (12,14,15). This discussion will focus upon the two most often used approaches for linear instruments, as well as present the more recent prescriptive methods specifically designed for nonlinear hearing aids, including:

- National Acoustic Laboratory (NAL)
- National Acoustic Laboratory Revised (NAL-R)
- Prescription of Gain and Output (POGO) (POGO II)
- Desired Sensation Level Input/Output (DSL I/O)
- Independent Fitting Forum (IHAFF)
- FIG6

The NAL and POGO procedures have been available since 1976 and 1983, respectively. The updated versions, NAL-R (1986) and POGO II (1988), created greater accuracy and utility facilitating great popularity among those that use prescriptive procedure. Since the most popular existing methods were developed from research with linear instruments (17), nonlinear circuits that adjust according to the input signal are not well re-

flected with these approaches and they do not assist clinicians in the prescriptive application of the nonlinear devices to adjust gain on the basis of input level. The rise of the use of conventional and programmable nonlinear hearing aid circuits has created a focus on new prescriptive methodologies to accommodate nonlinearity and assist clinicians in their ability to fit these products easily and accurately. Specifically, approaches that are the result of this research include the DSL I/O (6), the IHAFF (18) and the FIG6 (8,19).

## Prescriptive Procedures for Linear Hearing Aids

*National Acoustics Laboratory Revised (NAL-R)*

The NAL was first described by Byrne and Tonnisson(20) and revised by Byrne and Dillion (21) as an attempt to prescribe a frequency response that optimizes the potential for the understanding of speech after the hearing aid volume control had been adjusted to the wearer's preferred level (13). Speech optimization involved an attempt to maximize the amount of available speech signal, averaged over a wide frequency range. It was reasoned that this would most likely be achieved if all frequency bands of speech contributed about equally to the loudness of the signal. This procedure did not achieve the goal of amplifying all speech bands to equal loudness at a comfortable listening level or to MCL, which was demonstrated to be approximately equal. There was a general trend to provide too little low frequency gain relative to the mid- and high frequency gain, and to provide too much variation in frequency response slopes for the variation in the subjects' audiogram slopes (20). To correct this difficulty and add more utility to the NAL procedure, Byrne and Dillion (21) proposed a revised version of the National Acoustics Laboratory procedure, the NAL-R. This procedure specifically refined and optimized the energy available for certain hearing losses in all speech bands, particularly the low frequency band between 500 and 1000 Hz and reduced the calculation from a "half slope" or "half gain" rule to a "third slope" or a "one-third gain" rule. Although the calculation for threshold is a third slope, the NAL-R procedure requires a calculation of the three frequency average as well. The three frequency average calculation increases at a rate of 46 percent of the three frequency audiometric average. The NAL formula includes 10 dB of reserve gain (22). The specific NAL-R formula for the calculation of real-ear insertion gain (REIG) is presented in **Table 1 (5-1).**

Two other modifications to the NAL-R formula shown in **Table 2 (5-2)** have been suggested for those

**Table 1.**

| | | | | | |
|---|---|---|---|---|---|
| 250 | X + 0.31 | HL @ 250 | -17 | = | REIG |
| 500 | X + 0.31 | HL @ 500 | -8 | = | REIG |
| 1000 | X + 0.31 | HL @ 1000 | -3 | = | REIG |
| 1500 | X + 0.31 | HL @ 1500 | +1 | = | REIG |
| 2000 | X + 0.31 | HL @ 2000 | +1 | = | REIG |
| 3000 | X + 0.31 | HL @ 3000 | -1 | = | REIG |
| 4000 | X + 0.31 | HL @ 4000 | -2 | = | REIG |
| 6000 | X + 0.31 | HL @ 6000 | -2 | = | REIG |

**X= .05 (HL @ 500 + HL @ 1000 + HL @ 2000)**

**Table 5-1**
**NAL - R Formula for Real Ear insertion Gain**
**Byrne and Dillion (1986)**

with severe sensorineural hearing impairment (23,24). Modification 1 is an increase in the X factor equation if the 3-frequency average (500, 1000, 2000) exceeds 60 dB (12). The second modification increases the gain in the low and reduces the gain in the high frequencies if the degree of hearing loss at 2000 Hz exceeds 90 dB. This adjustment alters the hearing aid response for the necessary changes in the high and low frequency regions to accommodate a person with severe hearing impairment who needs more low frequency energy for power and less high frequency energy to reduce feedback problems.

*Calculation of the NAL-R*

Specific calculations of the NAL-R formula are presented in **Table 3 (5-3).** These are conducted in two steps: step 1 is the calculation of the X factor, or three frequency average, by adding the frequencies 500, 1000, and 2000. The total is then multiplied by a factor of 0.05. For the hearing loss presented in the table, the X factor is 6.5 dB. Step 2 is computation of the REIG for each frequency by multiplying the hearing loss by the factor of 0.31. For example, at 500 Hz, the REIG is equal to 0.31×30 dB or 9.30 dB plus the X Factor of 6.5 or 15.80 minus the 8 dB correction factor equals an REIG of 7.80 dB. This calculation is conducted for each frequency 250 through 6000 Hz. These modifications to the calculated REIG suggested for those with severe hearing losses

**Table 2.**

Modification 1.

If the X Factor sum of 500, 100, and 2000 exceeds 180, then :

0.116 (X - 180 )

is added to the X Factor.

(Where X= Combined total of HL @ 500, 1000, 2000)

Modification 2.

| Hearing Loss at 2 kHZ | Frequency in kHz. | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| dB HL | .25 | .5 | .75 | 1 | 1.5 | 2 | 3 | 4 | 6 |
| 95 | 4 | 3 | 1 | 0 | -1 | -2 | -2 | -2 | -2 |
| 100 | 6 | 4 | 2 | 0 | -2 | -3 | -3 | -3 | -3 |
| 105 | 8 | 5 | 2 | 0 | -3 | -5 | -5 | -5 | -5 |
| 110 | 11 | 7 | 3 | 0 | -3 | -6 | -6 | -6 | -6 |
| 115 | 13 | 8 | 4 | 0 | -4 | -8 | -8 | -8 | -8 |
| 120 | 15 | 9 | 4 | 0 | -5 | -9 | -9 | -9 | -9 |

NAL - R Adjustment for Hearing Losses when they exceed 90 dB at 2 kHZ.

Table 5-2
NAL - R Formula Modifications for Severe to Profound Losses (Byrne, Parkinson & Newall, 1990, 1991)

**Table 3.**

| Patient Audiometric Thresholds | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| 250 | 500 | 750 | 1000 | 1500 | 2000 | 3000 | 4000 | 6000 |
| 25 | 30 | 35 | 45 | 45 | 55 | 65 | 75 | 90 |

Step 1
X= .05 (HL @ 500 + HL @ 1000 + HL @ 2000)
X= .05 (30+45+55)
X= .05 (130)
X= 6.5

Step 2

| | | | | | | NAL-R REIG |
|---|---|---|---|---|---|---|
| 250 | 6.5 + 0.31 | (25) | = 14.25 | -17 | = | -2.75 |
| 500 | 6.5 + 0.31 | (30) | = 15.80 | - 8 | = | 7.80 |
| 1000 | 6.5 + 0.31 | (35) | = 17.35 | -3 | = | 14.35 |
| 1500 | 6.5 + 0.31 | (45) | = 20.45 | +1 | = | 21.45 |
| 2000 | 6.5 + 0.31 | (45) | = 20.45 | +1 | = | 21.45 |
| 3000 | 6.5 + 0.31 | (55) | = 23.55 | -1 | = | 22.55 |
| 4000 | 6.5 + 0.31 | (65) | = 26.65 | -1 | = | 25.65 |
| 6000 | 6.5 + 0.31 | (45) | = 20.45 | +1 | = | 21.45 |

Table 5-3
NAL - R Formula Real Ear insertion Gain calculations
Byrne and Dillion (1986)

(23,24) and are added to the calculated REIG to facilitate better understanding ability with less feedback. Fortunately, most probe-microphone systems can compute the NAL-R REIG automatically.

*Prescription of Gain and Output (POGO)*

The objective of the POGO approach was to present a practical approach that had some basis in what was known about gain preferences of persons with hearing impairment (25). An additional concern was to outline an approach that presented gain and output as critical characteristics of the prescription (16). Essentially, POGO is Lybarger's 1/2 Gain Rule, with a correction factor in the low frequencies to facilitate better speech understanding ability (12). A later version of this technique, POGO II, modified the gain when the hearing loss was greater than 65 dB HL (26), increasing the gain by half the amount that the hearing loss exceeds 65 dB (12). Since the POGO II modification refines the POGO approach, this is the advocated procedure. Similar to other prescriptive procedures, the POGO II calculation generates REIG; its formula is presented in **Table 4 (5-4)** and its calculation of REIG and output is presented in **Table 5 (5-5).**

*Calculation of POGO II*

**Table 5 (5-5)** presents the same hearing impairment as in **Table 3 (5-3),** except UCL measurements at 500, 1000, and 2000 have been added. To compute the formula, the first step is to divide the potential user's hearing loss by 2 at each frequency and record the product.

**Table 4.**

| | Real Ear insertion Gain | | Saturation |
|---|---|---|---|
| 250 | 1/2 HL + 1/2 (HL-65) | -10 | |
| 500 | 1/2 HL + 1/2 (HL-65) | -5 | Peak SSPL 90 = [(UCL @ 500 +1 K + 2K)/3] + 4 |
| 1000 | 1/2 HL + 1/2 (HL-65) | | |
| 2000 | 1/2 HL + 1/2 (HL-65) | | |
| 3000 | 1/2 HL + 1/2 (HL-65) | | |
| 4000 | 1/2 HL + 1/2 (HL-65) | | |

Table 5-4
POGO II formula for calculating Real Ear Gain
Schwartz, Lyregaard and Lundh (1988)

**Table 5.**

| Patient Audiometric Thresholds | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| 250 | 500 | 750 | 1000 | 1500 | 2000 | 3000 | 4000 | 6000 |
| 25 | 30 | 35 | 45 | 45 | 55 | 65 | 75 | 90 |
| UCL = 95 | | UCL= 95 | | UCL= 100 | | | | |

| | | | | POGO II REIG |
|---|---|---|---|---|
| 250 | 12.5 + 0 | -10 | = | 2.5 |
| 500 | 15 + 0 | -5 | = | 10 |
| 1000 | 22.5 + 0 | | = | 22.5 |
| 2000 | 27.5 + 0 | | = | 27.5 |
| 3000 | 32.5 + 0 | | = | 32.5 |
| 4000 | 37.5 + 5 | | = | 43.5 |

Saturation

96.66 + 4 dB = Peak SSPL90 100.66

Table 5-5
POGO II formula Real Ear Gain calculations
Schwartz, Lyregaard and Lundh (1988)

Step 2 is to consider whether the hearing loss at the particular frequency being calculated is greater than 65 dB. If so, subtract 65 dB from the impairment, then divide by 2 and add that factor to the figure obtained in step 1. In **Table 5 (5-5),** for 500 Hz, the hearing impairment is 30 dB. Step 1 is to divide 30 by 2, and obtain 15 dB. Since this impairment is less than 65 dB, the correction is 0 dB. Thus, the POGO II REIG for 500 Hz is 15 dB plus 0 dB correction, or 15 dB minus the low frequency factor for 500 Hz of -5 dB for a final REIG of 10 dB.

For the same hearing loss, the calculation for 4000 Hz is slightly different as it exceeds 65 dB. Step 1 is to divide 75 by 2 and obtain 37.5 dB. Since 75 is greater than 65 dB, step 2 is to subtract 65 dB from 75 dB and obtain 10 dB. Next, divide 10 dB by 2 and obtain 5 dB, which is added to 37.5 dB obtained in step 1 for a total REIG of 43.5 dB.

The POGO II formula also offers a calculation for output that utilizes tonal UCL measurements for 500, 1000, and 2000 Hz. To compute the output for the hearing impairment presented in **Table 5 (5-5),** average these UCLs and add 4 dB to arrive at the recommended peak SSPL 90. In this example, add 95+95+100 and divide that sum by 3 to obtain 96.66 then add the 4 dB to yield a total of 100.66 peak SSPL 90. As in NAL-R, these formulae are readily available on computer software to facilitate computation and are computed by most probe-microphone systems automatically.

### The 1980s Fitting Dilemma

Market studies provide information regarding the hearing impaired population that relate to the 1980s fitting dilemma (27). The data indicate that 88.1 percent of hearing losses are either mild (42.2 percent) or moderate (45.9 percent) with the remainder being severe (8.9 percent) and profound (3.0 percent) losses. In the daily clinical routine, it is obvious that the majority of these individuals with hearing impairment (95 percent or more) have sensorineural losses.

For many years, researchers have documented the nonlinear function of the cochlea and thus, the nonlinearity of sensorineural hearing loss (28–31). This can be seen in individual cases by conducting loudness growth measures and observing the person's dynamic range at various frequencies. Specifically, cochlear nonlinearity refers to the fact that, according to frequency, the cochlea tends to compress high intensity sounds. Cochlear pathology is known to 1) result in reduced acuity in low, middle, high, or all frequencies; 2) reduce the dynamic range unequally across frequencies; 3) produce inherent distortion; 4) reduce frequency resolution; 5) show abnormal growth of loudness (recruitment); and 6) create other alterations of the incoming acoustic signal (16).

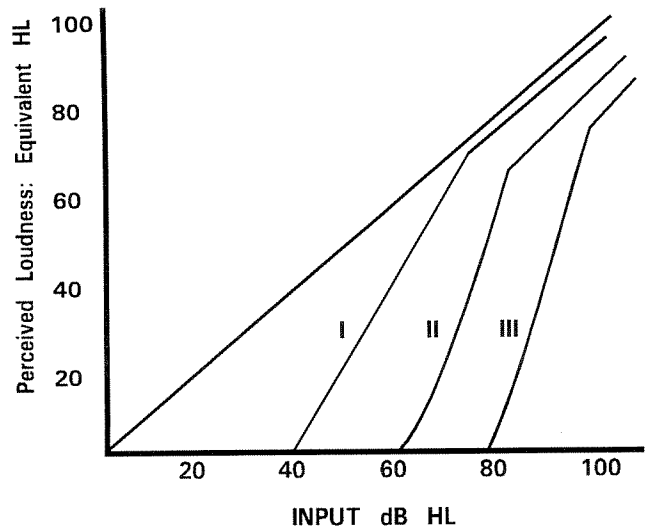



**Figure 1.**
Three types of sensorineural hearing loss: I = 40 dB hearing loss with normal or near-normal loudness sensations; II = 60 dB leveling off below normal-loudness curve; III = 70 dB+ loss with reduced curve for high intensity and reduced dynamic range in noise.

Recently, Killion and Fikret-Pasa (32) differentiated three types of sensori-neural hearing losses, Type I, Type II, and Type III, each with a somewhat different fitting rationale (**Figure 1**). The Type I hearing loss is characterized by individuals who exhibited about a 40 dB hearing loss and normal or near normal loudness sensations for all sufficiently intense sounds. Type II is characterized by about a 60 dB hearing loss and a loudness growth function that levels off below the normal-loudness curve but doesn't reach it. Type III is characterized by a 70 dB or greater hearing loss, a reduced loudness curve for high intensity sounds, and a reduced dynamic range for intelligibility in difficult situations. Berlin (28) suggests that Type I individuals probably have some outer hair cell loss, the Type II impairments have substantial outer hair cell loss and some inner hair cell loss, while the Type III have substantial outer hair cell loss and substantial inner hair cell loss. He concludes that fitting linear hearing aids to people with primarily outer hair cell loss is much less likely to work than fitting compression hearing aids, as these aids can mimic the gain function of

the outer hair cells at low input intensities. Comparing the Kochkin (27) data with that of Killion and Fikret-Pasa (32) would suggest that about 45.9 percent of the hearing impaired population could fall into the Type I category, 42.2 percent could be Type II, and only about 12 percent could be Type III. Thus, it could be reasoned that most hearing losses, probably as many as 85–90 percent, are mild-to-moderate sensorineural ones of Type I and Type II.

As the 1980s closed and the 1990s began, nonlinear hearing instruments were recognized as appropriate for the majority of persons with hearing impairment, since most of their hearing losses were sensorineural and theoretically nonlinear. The question was, "Does it make sense to use a prescriptive fitting method designed to fit a linear hearing aid to fit a nonlinear hearing aid?" In reality, clinicians either utilized linear hearing aids and a prescriptive technique, or they used nonlinear hearing aids and did not utilize prescriptive measures. This dilemma may still contribute to the fact that only 50 percent of the clinicians utilize prescriptive procedures to fit their clients (33). There have been some recent attempts to design prescriptive techniques for nonlinear hearing aid circuits, including the DSL I/O (6), the IHAFF protocol (7), and the FIG6 (8,19). These procedures are still in their developmental stages and will continue to evolve and become the methods for prescriptive fitting of nonlinear and/or programmable hearing instruments.

## Prescriptive Procedures for Nonlinear Hearing Aids

*Desired Sensation Level Input/Output*

The DSL I/O approach is a close relative of the DSL approach first used as a prescriptive method for children (34,35). Cornelisse et al. (6) indicate that their purpose was to develop a device-independent formula for the specification of the electroacoustic characteristics of a personal hearing aid or, theoretically, the ideal amplified output for a range of input levels. They summarize the DSL I/O approach as a series of mathematical equations (**Table 6 [5-6]**) that describe the relationship between the input level of a signal delivered to a hearing aid and the output level produced by the hearing aid. The I/O approach divides the input dynamic range into three regions: 1) input levels below compression threshold, or $I_{min}$; 2) input levels that will exceed the compression threshold when amplified, or $I_{max}$; and, 3) the area between these two limits. For input signal levels equal to or less than $I_{min}$, linear gain is applied to the signal (i.e., below compression threshold). For signal levels equal to

**Table 6.**

1. $I < I_{min}$ = (Linear Gain): $O = O_{min} I_{min} I$ )

2. $I_{min} < I < I_{max}$ (Linear Compression):

$$O = \left[\frac{I - I_{min}}{I_{max} - I_{min}}\right] X ( O_{max} - O_{min} ) + O_{min}$$

3. $I \geqslant I_{max}$ (Output Limiting):

$$O = O_{max}$$

**Table 5-6**
**Desired Sensation Level Input / Output**
**Formula for calculating Real Ear Insertion Gain**
**Cornelisse, Seewald and Jamieson (1995)**

or greater than $I_{max}$, the output is limited to $O_{max}$ (output limiting). For input signals between $I_{min}$ and $I_{max}$, the I/O formula applies compression to the signal. The I/O formula is frequency-specific, (i.e., the I/O is described for each frequency band). The I/O formula is also device-independent; that is, the formula was not designed for any particular model but rather was designed to be applicable to the class of wide dynamic range compression hearing aids in general. Furthermore, the I/O approach can be applied to fit either a single channel or a multichannel compression device. Specification of frequency-specific gain and maximum output are sufficient to describe the electroacoustic characteristics of a linear gain hearing aid. Available selection procedures are inadequate for the compression hearing aid because these algorithms only specify one gain value. Their approach was to describe the frequency-specific I/O characteristis of a compression device.

*Calculation of the DSL I/O*

**Figure 2** presents a typical plot for one frequency band (6). The input SPL (sound field level: dBSF) is plotted along the abscissa and the output SPL (ear canal level: dBEC) is plotted along the ordinate. The diagonal line (I+SFt) indicates the transfer function from the unaided input level to the unaided output level. When a horizontal line and vertical line intersect along the diagonal line (I+SFt), then the horizontal and vertical correspond to equivalent levels at the two measurement points (i.e.,

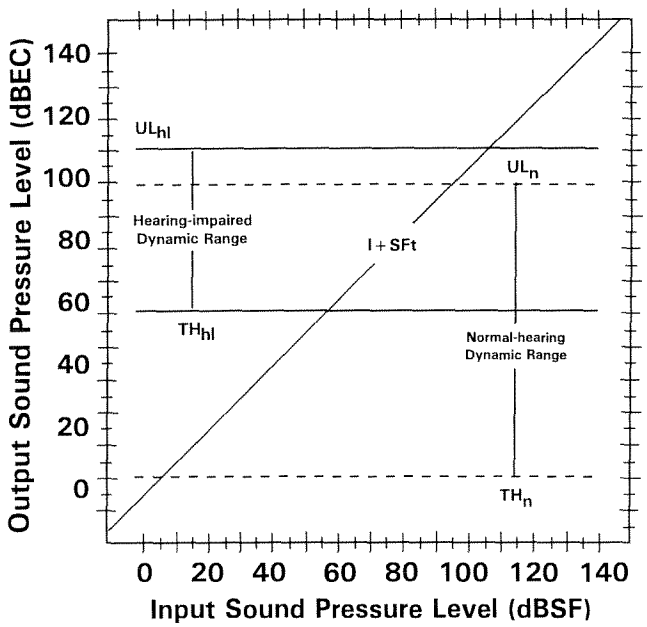



**Figure 2.**
A typical I/O plot showing the auditory dynamic range for normal-hearing and an individual with a 50-dB hearing loss at 1000 Hz. The normal threshold ($TH_n$) and upper limit ($UL_n$) are indicated by dashed lines. The hearing-impaired threshold ($TH_{hl}$) and upper limited ($UL_{hl}$) are indicated by solid lines. The diagonal line (I+SFt) indicates the normal transfer function from the input SPL to the output SPL (6).

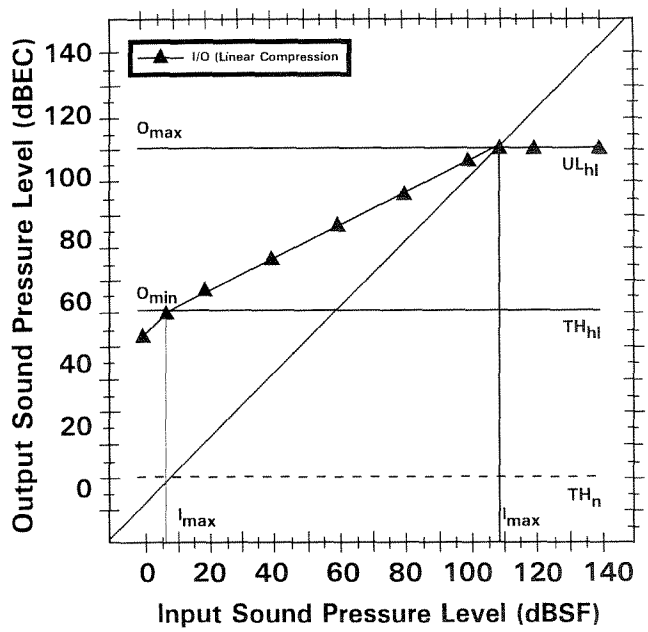



**Figure 3.**
The I/O linear compression function for the 50-dB hearing loss at 1000 Hz, as depicted in **Figure 2** in this chapter (6).

dBEC and dBSF). Hypothetical data used in this example are for an individual with a 50 dB hearing loss at 1000 Hz; the DSL I/O must be plotted for each frequency of interest, usually 250, 500, 750, 1000, 1500, 2000, 3000, 4000, and 6000 Hz. The standard hearing-impaired auditory dynamic ranges are indicated with horizontal lines in **Figure 2.** The nonimpaired-hearing threshold ($TH_n$) is 9.6 dBEC for a 1000 Hz pure tone. The hearing-impaired threshold ($TH_{hi}$) is 59.6 dBEC, which is equal to $TH_n$ plus 50 dB. The standard hearing upper limit ($UL_n$) of comfort is 99 dBEC, which corresponds to the real-ear saturation response (RESR) for a 0-dB hearing loss (35). The upper limit ($UL_{hi}$) of comfort for individuals with hearing impairment is 110 dBEC, which corresponds to the RESR for a 50 dB hearing loss (35). The standard auditory dynamic range ($DR_n$) equals $UL_n$ minus $TH_n$, which is 89.4 dB. The hearing impaired unaided (output) dynamic range ($DR_u$) equals $UL_{hi}$ minus $TH_{hi}$, which is 50.4 dB. The linear compression I/O function provided by the formula for an individual with a 50 dB hearing loss is plotted in **Figure 3.** In this example, $I_{max}$ is equal to 107.4 dBSF and $I_{min}$ is equal to 7.0 DBSF. The aided (input) dynamic range ($DR_a$) equals $UL_{hi}$ minus $TH_n$ which is 100.4 for this example (6).

The DSL I/O is a complex formula that considers many variables at many frequencies simultaneously to facilitate accurate calculation of the specific parameters of the fitting. The use of a computer program that only requires the clinician to enter threshold data and automatically calculates the variables and presents ideal I/O curves for a particular hearing impairment is a great help (6).

## Independent Hearing Aid Fitting Forum (IHAFF)

The IHAFF was formed by a group of researchers, engineers, and clinicians with the goal of developing a standardized comprehensive hearing aid fitting protocol to assist in the selection and fitting of contemporary nonlinear and programmable hearing aids. The IHAFF protocol assumes that if an appropriate hearing aid operates to restore the sensation of loudness perception over a wide bandwidth, the wearer of the hearing aid will have a good opportunity to achieve maximum speech recognition ability (18). The primary goals of the IHAFF protocol are to provide for amplification so that soft speech is perceived as soft; conversational level speech is perceived as comfortable; and loud speech and sounds are loud but not uncomfortable (36). To facilitate these goals, the IHAFF

group has recommended the use of software-assisted loudness judgment measures, a selection of specific electroacoustic characteristics of hearing aids, and a self-assessment questionnaire that is useful before and after the fitting. The rationale for the circuit selection process is based upon the loudness growth information (36), obtained by the use of the Contour Test that involves the use of warble tones presented to the client at a minimum of two frequencies, typically 500 Hz and 3000 Hz (7). The client judges their loudness on a scale of seven categories of loudness. At the heart of the IHAFF procedure (i.e., Visualization of Input/Output Locator Algorithm (VIOLA), that assists in the management of the variables involved in hearing aid circuit selection (7,17). It calculates the relationship between overall speech input levels for soft, average, and loud speech at the hearing aid microphone and the user's loudness judgments for warble tones (36). It then displays (See **Figure 4**) individual loudness judgment categories along with an I/O graph noting input (30 to 90 dB in 10 dB increments) at the hearing aid microphone versus output of the hearing aid as measured in HA-1, 2 cc coupler. Ultimately, the VIOLA software allows the clinician to enter different combinations of overall gain at an input level of 40 dB, output and compression characteristics (threshold kneepoint and compression ratio), maximum output, overall gain at 60 dB and slope characteristics at the frequencies for which loudness judgments were measured. The calculated I/O curve is then displayed and evaluated by the clinician for compliance with the target. If the instrument chosen does not meet the target, then characteristics of devices can be changed (if programmable) or the instrument can be changed (if conventional). The VIOLA software screen will predict 2 cc coupler gain for hearing aids. Although the IHAFF is still being researched, it promises to be a precise, predictive formula for nonlinear instruments.

### FIG 6 Procedure

The Fig 6 Procedure is a loudness-based fitting formula designed to accommodate the types of hearing losses described earlier by Killion and Fikret-Pasa (32); indeed the name of this approach is derived from the loudness growth concept presented in Figure 6 of that article. Killion has added a loudness-based fitting formula to the array of prescriptive fitting procedures. In its current form, it is a spreadsheet approach to estimating the level-dependent gain and frequency response of nonlinear hearing aids (8,9). Since these aids can change their gain and frequency response depending upon the input level, the FIG6 can be utilized to calculate gain and frequency response for low-level, moderate-level, and high-level sounds. The formula for the FIG6 is presented in **Table 7 (5-7).** Calculations of REIG utilizing this method are presented in **Figures 6–10.** The FIG6 method is a software program that asks the clinician to enter the person's audiogram (**Figure 5**) and graphically presents a calculation of REIG (**Figure 6**). If a hearing aid is to be ordered for this, FIG6 automatically calculates the REIG and the appropriate average coupler response for flat insertion gain (CORFIG) for BTE, as shown in **Figure 7** (37,38); ITE (**Figure 8**), ITC (**Figure 9**), and CIC (**Figure 10**) hearing aids (8,19).

### Once I Figure the Gain/Output Requirements, How Do I Order a Hearing Aid?

Prescriptive procedures can be utilized to choose either a hearing aid circuit offered in the manufacturers' matrix book, or to order a custom-made ITE or ITC instrument. Unless the clinician allows the manufacturers to utilize their own methods to choose the circuit, it is necessary to give them specific information to facilitate the order. This is why computing the REIG by a formula is not enough to insure that the instrument ordered will provide the appropriate amount of gain and appropriate frequency response. To obtain an instrument constructed to the appropriate amounts of REIG, the calculated prescription must be presented in 2 cc coupler terms.

*Conversion of the REIG to 2 cc Coupler*

These conversions are conducted by the use of the CORFIG, which is either Coupler Response for Flat Insertion Gain (32) or CORRection FIGure (37). The specific theoretical issues regarding CORFIGs are beyond the scope of this discussion and are presented elsewhere (37,38). The values in **Table 8 (5-8),** when subtracted from the REIG, convert the REIG into 2 cc coupler gain. Transforming the 2 cc coupler gain back into REIG, is GIFROC (CORFIG spelled backwards): the CORFIG values are simply added to the 2 cc coupler values (38).

Since there are different microphone locations for various types of hearing instruments, the CORFIGs are different depending upon the type of hearing instrument used, BTE, ITE, ITC, and CIC. In **Table 9 (5-9),** the NAL-R calculations for REIG are presented for the same hearing impairment. To change these data into that for a 2 cc coupler, the clinician decides on the type of hearing instrument to be utilized and subtracts the average COR-

Visual Input/Output Locator Algorithm (VIOLA)
09/04/95

NAME : SAMPLE FILENAME :
EAR : RIGHT HEARING AID TYPE : ITE

| | 500 Hz | 3000 Hz |
|---|---|---|
| Gain at 40 dB | | |
| Comp. Threshold 1 | | |
| Comp. Ratio 1 | | |
| Comp. Threshold 2 | | |
| Comp. Ratio 2 | | |
| Max. Output | | |
| Gain at 60 dB | | |
| Slope (dB/oct) | | |

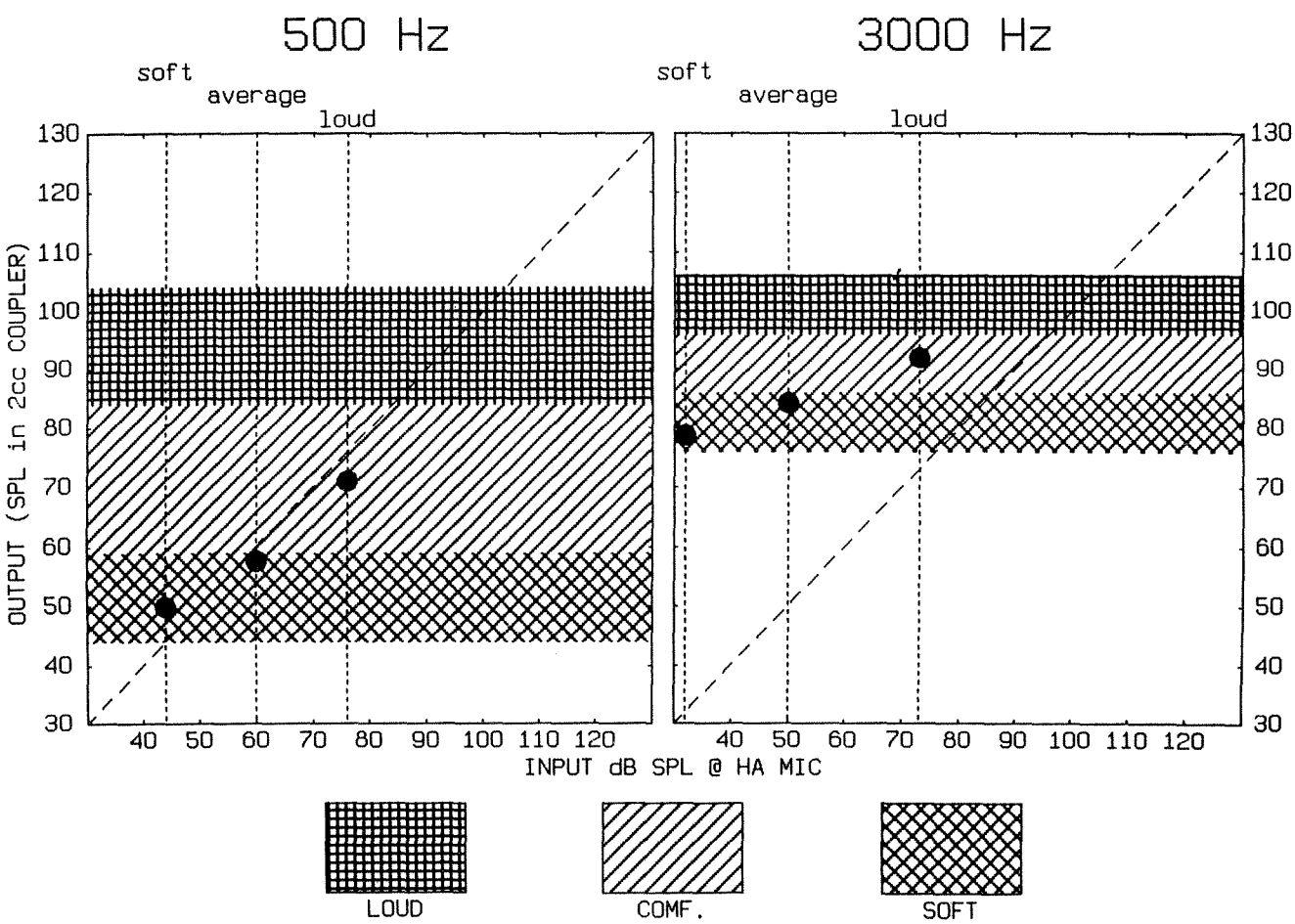



**Figure 4.**
Visual input/output locator algorithm (VIOLA).

Table 7.

**A. Gain for low level sounds:**

1. G = 0 for 0 to 20 dB HL
2. G = HL-20 for 20 to 60 dB HL
3. G = HL-20 - .5*(HL-60) for HL >= 60 dB

**B. Gain at MCL:**

1. G = 0 for 0 to 20 dB HL
2. G = 0.6 *(HL-20) for 20 to 60 dB HL
3. G = 0.8 * HL - 23 for HL >= 60 dB

**C. Gain for high-level sounds:**

1. G = 0 for 0 to 40 dB HL
2. G = 0.1 * (HL-40)^1.4 for HL >= 40 dB

Table 5-7
FIG6 Formula, Killion (1994)

Table 8.

| Hearing Aid Type | Frequency in kHz. .25 | .5 | 1 | 2 | 3 | 4 | 6 |
|---|---|---|---|---|---|---|---|
| BTE | -3 | -2 | -2 | -7 | -10 | -10 | 0 |
| ITE | 0 | 0 | 0 | -2 | -6 | 0 | 3 |
| ITC | 0 | 0 | 0 | -2 | -3 | 3 | 7 |

Note: To convert REIG to 2 cc coupler gain, the above numbers are subtracted from REIG. To convert 2 cc coupler gain to predicted REIG, the above numbers are added to 2cc coupler gain.

Table 5-8
Average CORFIGs for BTE, ITE, ITC
Hearing Aids (Hawkins 1992b)

FIG for that type of hearing aid to the calculated REIG. This gives the manufacturer specific 2 cc coupler parameters on which to base the instrument. In the example in **Table 9 (5-9),** the use of an ITE instrument for the hearing loss presented at 500 Hz, the REIG is 7.80 dB−CORFIG. The result is 0 dB=7.80 dB in the 2 cc coupler. At 3000 Hz, the REIG is 23.55 dB−CORFIG or −6 dB for a total of 18.55 dB in the 2 cc coupler.

A full explanation of the specifics of the hearing aid ordering process are available elsewhere (22,38), and the protocol for calculating desired 2 cc coupler values used in ordering a custom ITE are summarized below:

1. Obtain desired REIG using a prescriptive fitting procedure.

Table 9.

| Patient Audiometric Thresholds | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| 250 | 500 | 750 | 1000 | 1500 | 2000 | 3000 | 4000 | 6000 |
| 25 | 30 | 35 | 45 | 45 | 55 | 65 | 75 | 90 |

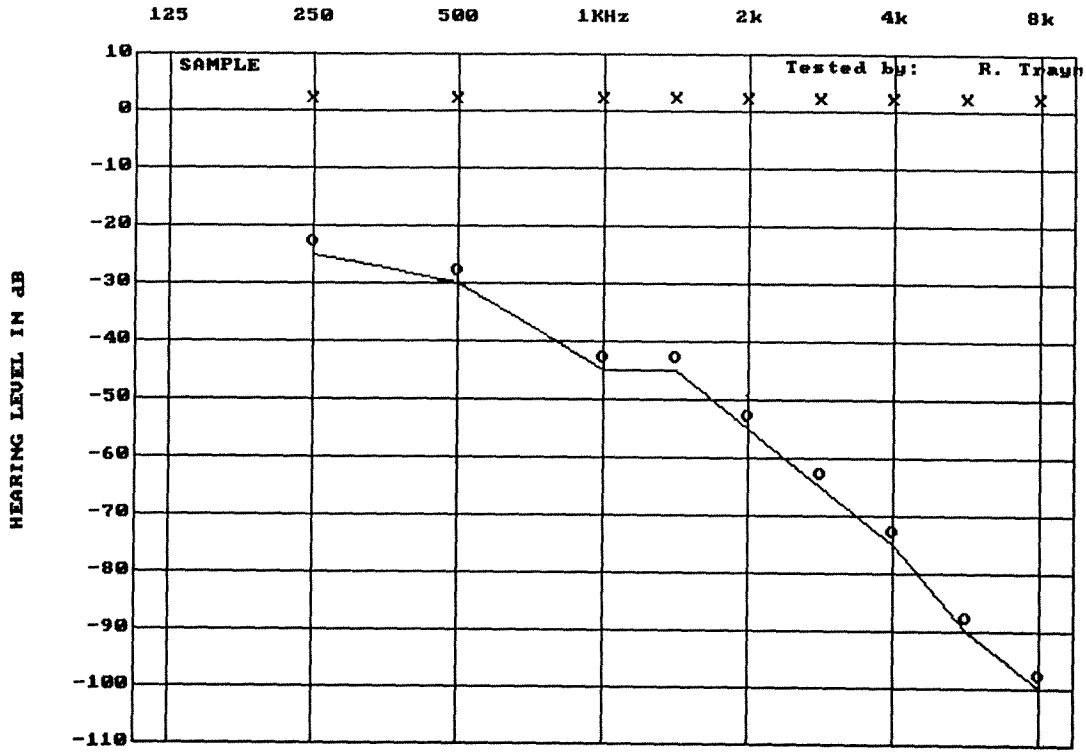
X= .05 (HL @ 500 + HL @ 1000 + HL @ 2000)
X= .05 (30+45+55)
X= .05 (130)
X= 6.5

| | | | | | | | NAL-R REIG | Ave. ITE CORFIG | 2 cc Coupler Gain |
|---|---|---|---|---|---|---|---|---|---|
| 250 | 6.5 + 0.31 | (25) | = 14.25 | -17 | = | | -2.75 | 0 | -2.75 |
| 500 | 6.5 + 0.31 | (30) | = 15.80 | - 8 | = | | 7.80 | 0 | 7.80 |
| 1000 | 6.5 + 0.31 | (35) | = 17.35 | -3 | = | | 14.35 | 0 | 17.35 |
| 1500 | 6.5 + 0.31 | (45) | = 20.45 | +1 | = | | 21.45 | - | |
| 2000 | 6.5 + 0.31 | (45) | = 20.45 | +1 | = | | 21.45 | -2 | 19.45 |
| 3000 | 6.5 + 0.31 | (55) | = 23.55 | -1 | = | | 22.55 | -6 | 16.55 |
| 4000 | 6.5 + 0.31 | (65) | = 26.65 | -1 | = | | 25.65 | 0 | 25.65 |
| 6000 | 6.5 + 0.31 | (45) | = 20.45 | +1 | = | | 21.45 | 3 | 24.45 |

Table 5-9
NAL - R Formula Real Ear insertion Gain calculations with CORFIG for 2 cc coupler gain
Byrne and Dillion (1986), Hawkins, 1992b)

2. Add reserve gain (if not figured as part of the formula).
3. Apply CORFIGs from **Table 8 (5-8)** to totals obtained in step 2 to obtain desired 2 cc coupler full on gain values.
4. Use calculated 2 cc coupler values to select desired matrix or provide actual 2 cc coupler values to the manufacturer for circuit selection.

One method of choosing a circuit for a hearing instrument is matrix approximation, involving the use of the above 2 cc coupler data and a manufacturer's matrix book (22). Most manufacturers provide a book that includes the 2 cc coupler specifications available for all of their custom ITE models. A matrix is provided for each model, defining such features as gain, SSPL-90, slope, and frequency bandwidth. These designations vary somewhat from one manufacturer to another but will still offer the same information. For example, one manufacturer may present its matrix as 40/12/108/W, where 40 is gain, 12 is the slope, 108 is the output, and W refers to the use of a wideband receiver. Another might present its matrix as 107/30/15, where 107 is output, 30 is gain, and 15 is slope, with no designation as to the type of receiver. For fully programmable circuits, the same procedure applies but does not require the manufacturer to specifically build the hearing aid according to prescriptive specifications. Each instrument is manufactured with the same capabilities available in a custom shell. Since the hearing instrument is fully programmable, the full range of parameters is available for the client at the time of fitting. These instruments can be thought of as




| Freq. | RIGHT | LEFT | Freq. | LEVEL dB SPL 40 | 65 | 90 | LEVEL dB SPL 40 | 65 | 90 |
|---|---|---|---|---|---|---|---|---|---|
| 125 | | | 125 | 0 | 0 | 0 | 0 | 0 | 0 |
| 250 | 25 | 0 | 250 | 5 | 3 | 0 | 0 | 0 | 0 |
| 500 | 30 | 0 | 500 | 10 | 6 | 0 | 0 | 0 | 0 |
| 1000 | 45 | 0 | 1K | 25 | 15 | 1 | 0 | 0 | 0 |
| 1500 | 45 | 0 | 1.5K | 25 | 15 | 1 | 0 | 0 | 0 |
| 2000 | 55 | 0 | 2K | 35 | 21 | 4 | 0 | 0 | 0 |
| 3000 | 65 | 0 | 3K | 43 | 29 | 9 | 0 | 0 | 0 |
| 4000 | 75 | 0 | 4K | 48 | 37 | 15 | 0 | 0 | 0 |
| 6000 | 90 | 0 | 6K | 48 | 37 | 15 | 0 | 0 | 0 |
| 8000 | 100 | 0 | 8K | 48 | 37 | 15 | 0 | 0 | 0 |
| | | | 12K | 45 | 35 | 12 | -3 | -3 | -3 |
| | | | 16K | 43 | 32 | 10 | -5 | -5 | -5 |

NOTES:
GAIN AT 6 & 8KHz IS LIMITED TO 4KHz VALUE
GAIN AT 12KHz = 8KHz GAIN - 2.5dB
GAIN AT 16KHz = 12KHz GAIN - 2.5dB

° LOW = 40dB SPL
° MCL = 65dB SPL
° HIGH = 95dB SPL

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**Figure 5.**
Individual loudness judgment categories. (*Permission granted by M.C. Killion, PhD.*)

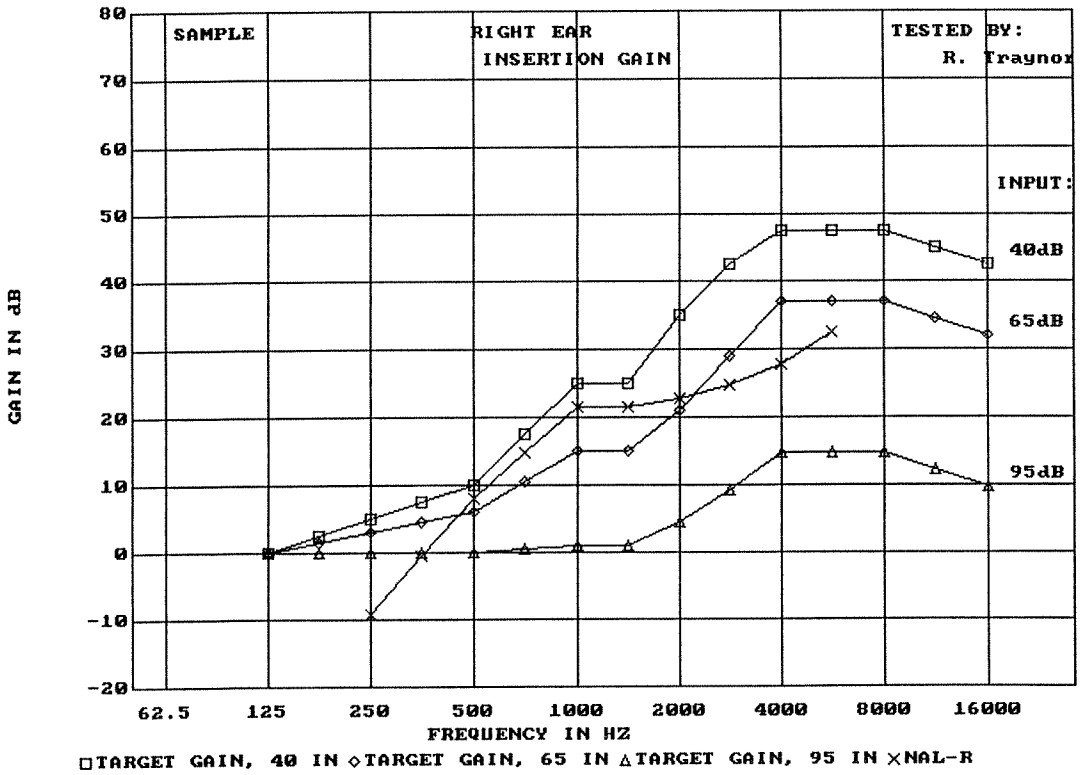


**Figure 6.**
Target gain vs. input level: Right ear insertion gain.

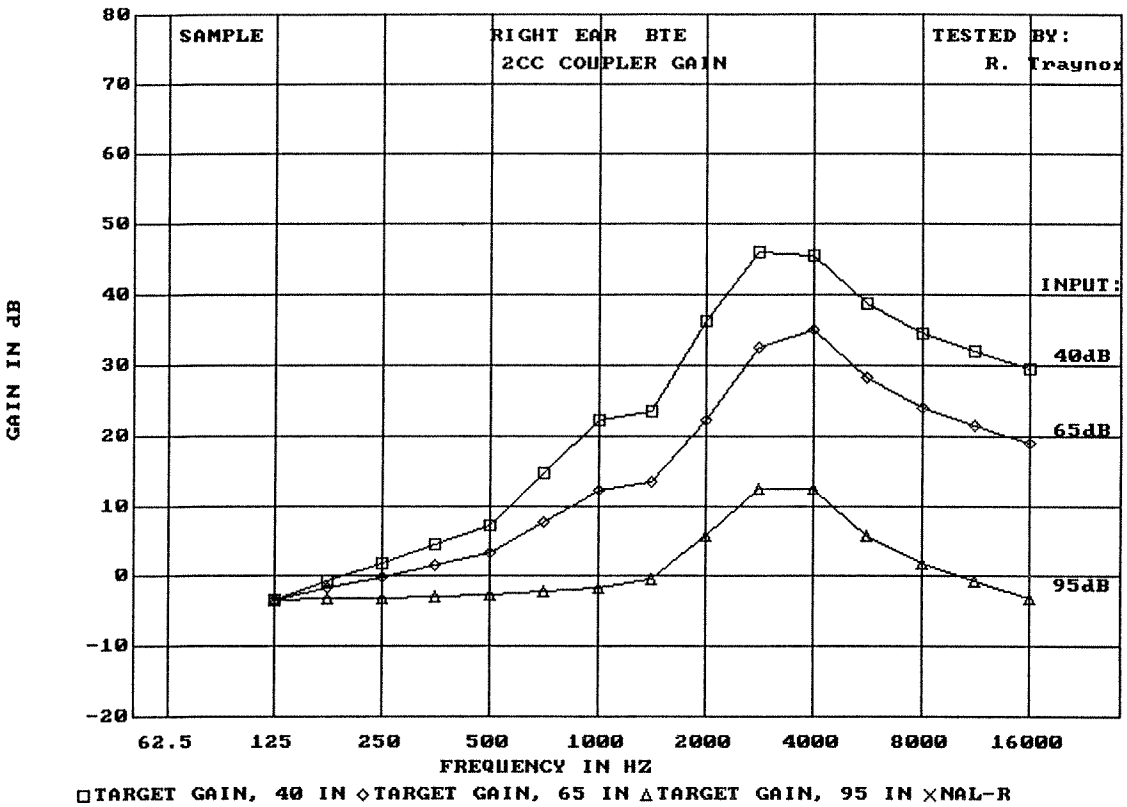


**Figure 7.**
Target gain vs. input level: Right ear BTE 2 cc coupler gain.

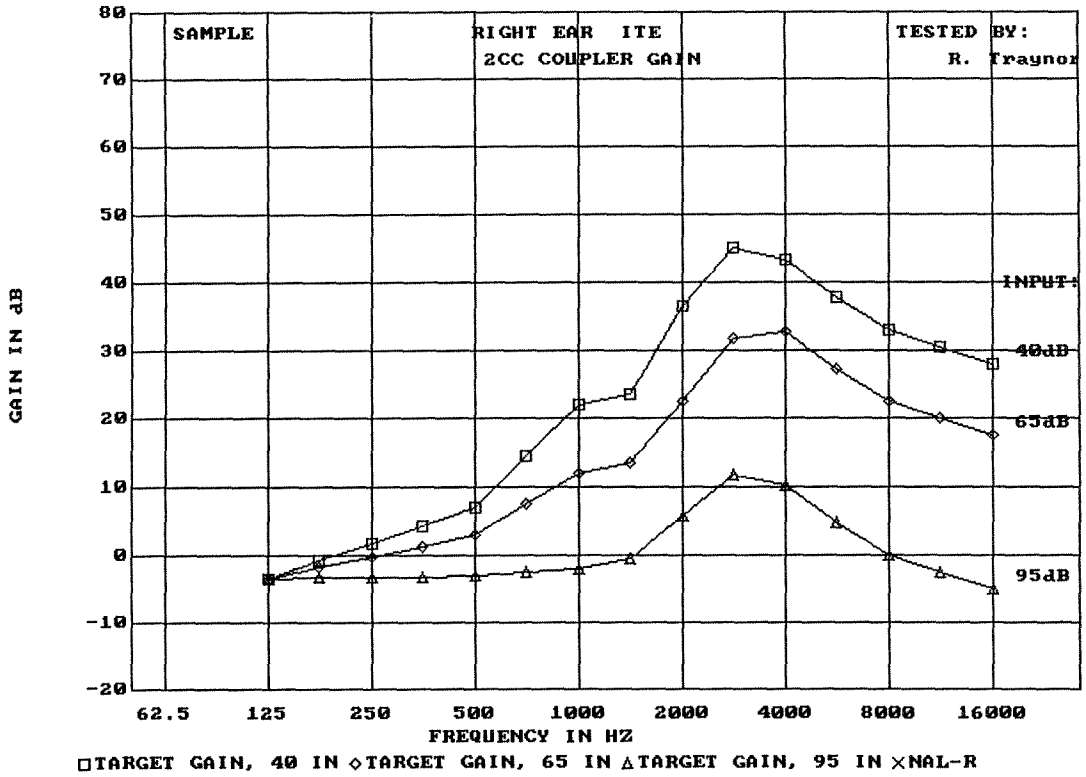


**Figure 8.**
Target gain vs. input level: Right ear ITE 2 cc coupler gain.

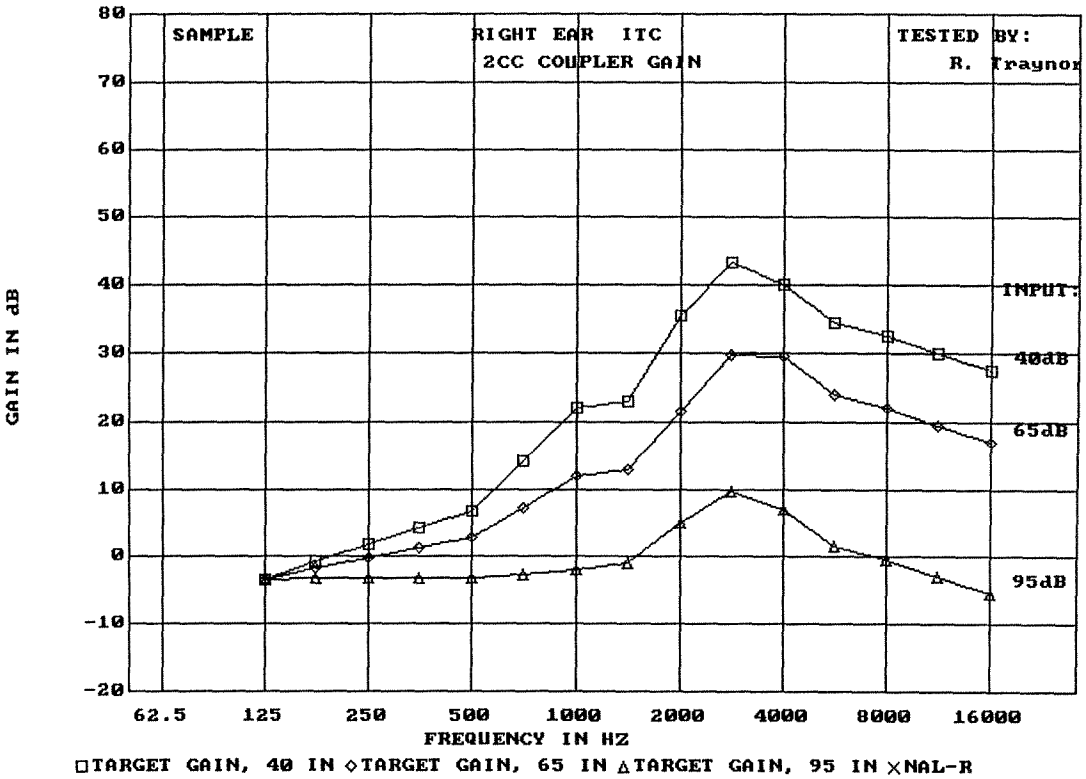


**Figure 9.**
Target gain vs. input level: Right ear ITC.

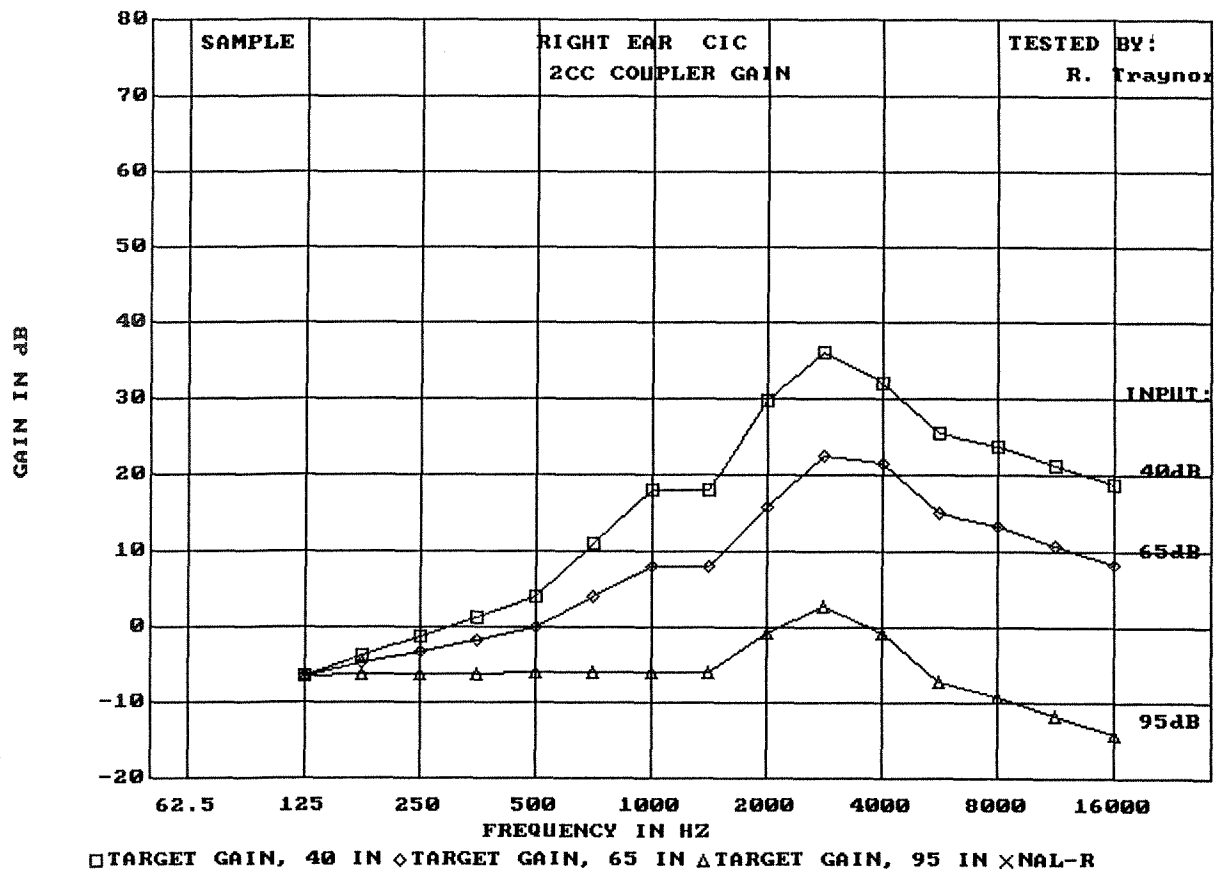


**Figure 10.**
Target gain vs. input level: Right ear CIC.

all, or the most possible, matrices available in one hearing aid.

## Once I Have the Hearing Aid, How Do I Evaluate It?

Functional gain and probe-microphone measurements are the most often utilized procedures for this purpose. 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 (38). The insertion gain of a hearing aid at a given frequency, also expressed in dB, is the difference between the aided and unaided eardrum SPL at that frequency. Both functional gain and insertion gain are real-ear measurements that require the user 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 responsive, whereas insertion gain is an objective, probe-microphone acoustic measurement that can be performed on a live subject or a manikin, such as KEMAR.

Particularly for linear hearing aid measurements, either procedure may be utilized, as both assess the parameter. The differences are the known behavioral testing variables, ±5 dB reliability, and subject variability. Also, functional gain takes longer and requires a bit more interpretation. Additionally, when using a functional gain procedure to evaluate many nonlinear circuits, caution must be exercised to present the signal below saturation of the instrument. In contrast, an insertion-gain measurement will be more reliable, probably within 1 or 2 dB at most frequencies, and takes less than half the time to obtain the same information, as well as be easier to interpret than functional gain measurement. Detailed information on the variables of evaluating prescribed instruments may be obtained from numerous sources (22,37,38).

## What Are the Variables of Using Prescriptive Procedure?

Killion and Revit (38) caution that even accurate calculations, carefully computed coupler transfer func-

tions (CORFIG/GIFROC), and rigid standards of manufacturing the hearing aid may still not yield the perfect result when measured on an individual probe-microphone system. Prescriptive techniques provide a scientifically based standard to begin the hearing aid fitting process; Killion and Revit discuss these limitations as nagging problems for both the manufacturer and the dispenser. Even when 2 cc coupler response of the hearing aid exactly matches the target 2 cc coupler response at the manufacturing facility, the measured insertion response (REIR) may not match the desired (target) REIR. They further describe the three main reasons for this difficulty. First, the effect of both the 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 the 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 upon average data. Third, measurement error, both random and not-so-random, often causes a significant difference that disappears with repeated measurements or is peculiar to the measurement setup.

## SUMMARY

This chapter attempts to present a practical discussion of the most popular prescriptive procedures for linear hearing aids and the emerging procedures for nonlinear instruments. It presents where these formulae come from, what they do, how to use them, and some of the pitfalls that can be present even when they are applied carefully.

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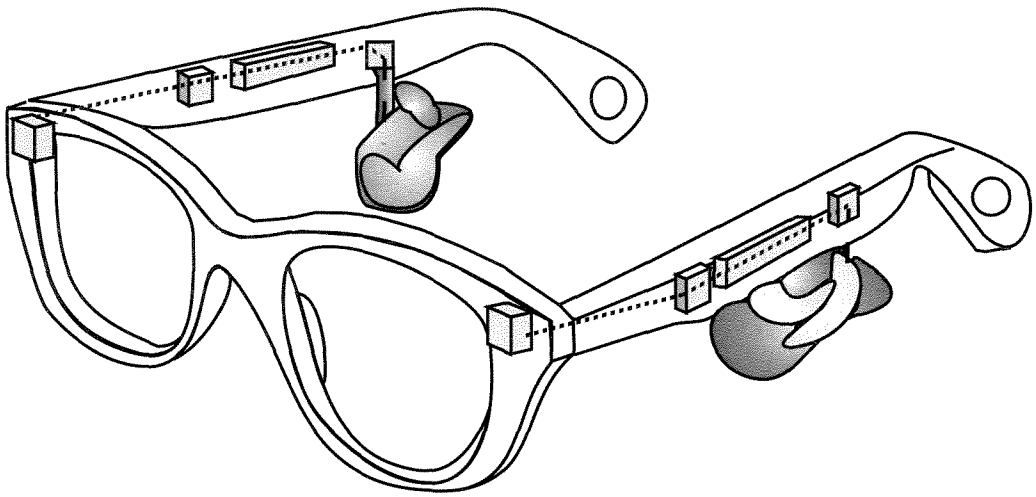
**ROBERT M. TRAYNOR, EdD,** in addition to being Clinical Professor at the University of Northern Colorado, is in private practice in Greeley, Colorado, and is a consultant to Bernafon-Maico. Dr. Traynor received his doctorate from the University of Northern Colorado in Greeley, Colorado. His major interest is in rehabilitative audiology.