Chapter Two

Circuitry Options: Signal Processing at Average Conversational Levels

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INTRODUCTION

One of the greatest challenges facing the hearing health care professional is the selection of the proper amplification device with the appropriate characteristics for the given individual. Recent technological advances have made this selection process even more difficult by increasing the number and the variety of parameter settings under the control of the audiologist. The challenge continues to grow because these advances occur at a rate that exceeds our ability to fully assess or evaluate them clinically (1). As a result, one is often called upon to make decisions regarding circuitry options for an individual without the benefit of an extensive clinical research base. In order to provide the most effective amplification services to the hearing-impaired population, the audiologist must be well versed in the available clinical literature and must be able to integrate information from engineers, the hearing aid industry, psychoacousticians, and other audiologists.

Throughout this chapter, a number of models or decision matrices will be utilized in terms of selecting circuitry options for potential hearing aid users. Before one can consider circuitry options, however, it is important to keep in mind that hearing aid selection and fitting need to be a part of the overall audiologic rehabilitation process. It is for this reason that the first model, taken from the work of Seewald (2), is presented in **Figure 1.**

In **Figure 1,** the fitting process is schematically represented as a series of sequential stages taking place within the broad context of audiological rehabilitation. The first step is a general assessment of the communicative problems and needs of the potential user. A plan for intervention is developed on the basis of this assessment. When that plan includes the use of a hearing aid, the next step is the assessment for fitting. In this figure, the dotted arrows show the interrelated nature of the four stages of that process (2). Although the remainder of this section of this chapter will be primarily concerned with hearing aid selection and verification of hearing aid performance, it is important to keep in mind that these steps constitute portions of the overall audiological rehabilitation process. The most advanced circuitry available will be of little benefit if it does not help meet the communicative needs of the individual.

PRESELECTION DECISIONS

Before one can begin the process of selecting circuitry options, there are a number of preselection decisions regarding amplification that must be made jointly by the audiologist and the prospective user. **Figure 2** lists two of the more important of these decisions: whether to pursue binaural or monaural amplification, and what hearing aid style to choose.

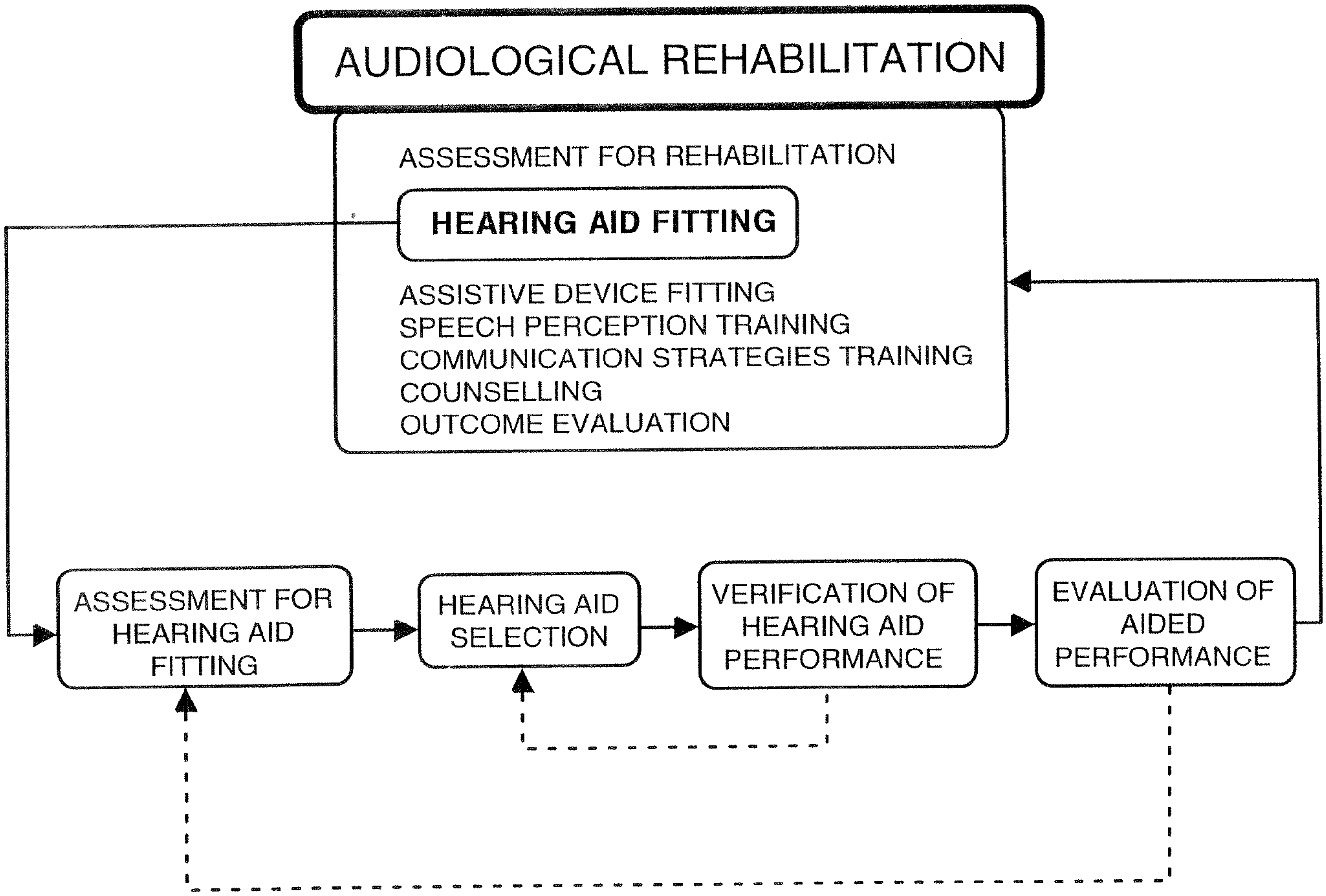

Figure 1.
Hearing aid fitting process.

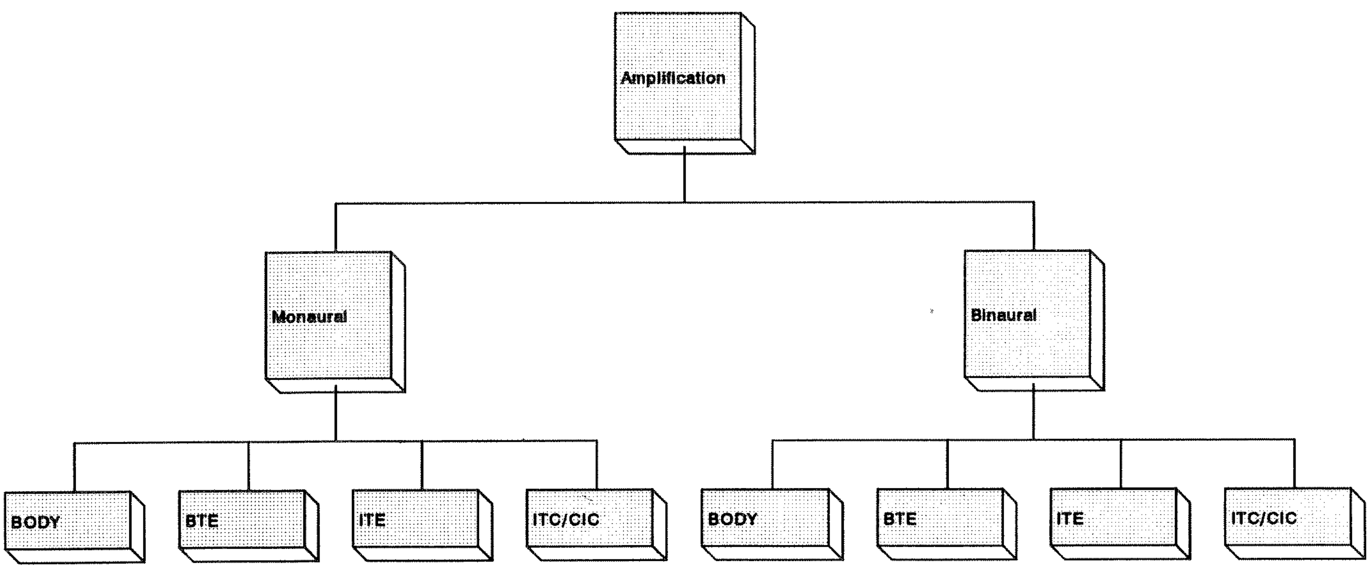

Figure 2.
Hearing aid fitting decision: Binaural vs. monaural.

Binaural Versus Monaural

Although there has been heated debate for years over the benefits provided by two hearing aids versus one, it is generally agreed that, unless there are significant contraindications, binaural amplification is the arrangement of choice. Its advantages are improved localization and improved speech recognition ability in noise, though it should be noted that these advantages are not easily demonstrated clinically (3,4), and many users may require a period of adjustment before experiencing them (5). There are also legal issues to consider: recently, some individuals have litigated against dispensing audiologists for failing to inform them of the possible benefits of binaural amplification during monaural fitting.

When contraindications, such as a large asymmetry in audiometric configuration, exist, it is sometimes necessary to amplify monaurally. Then one must decide which ear to amplify. A better approach than "better threshold" versus "poorer threshold" is to consider fitting the ear with the wider dynamic range, the ear with the greater word recognition ability, and the ear preferred by the client. When all three considerations point to the same ear, the decision is straightforward. When there is no clear-cut winner, the decision must be made by balancing or weighing the importance of these issues.

Hearing Aid Style

Decisions also need to be made regarding style. Body-worn aids represent a very small percentage of those sold annually. Ear-level amplification devices include behind-the-ear (BTE), in-the-ear (ITE), in-the-canal (ITC) and completely in-the-canal (CIC) hearing aids, and the styles vary in terms of the practical and/or acoustical advantages each provides. For example, although users tend to pursue smaller aids for cosmetic reasons, there are numerous acoustical advantages to CIC instruments (increased high frequency gain, increased ear canal SPL, and reduction of the occlusion effect) that must be considered when choosing a style appropriate for an individual.

Linear Amplification

Once binaural/monaural and style choices have been made, the next decision regards circuitry: whether to select a linear or a nonlinear amplifier. **Figure 3** is a simple schematic of this step in the selection process. Although there is a wealth of literature suggesting that the vast majority of clients should be fit with nonlinear amplifiers (6,7) the fact remains that approximately 80 percent of the hearing aids dispensed in the United States are linear (8).

A linear circuit provides a consistent amount of gain across a wide range of input levels. That is, the increase in amplified output in the ear canal is equal to the decibel increase in the input signal until the amplifier reaches the point of saturation. This 1:1 relationship between input and output is sometimes referred to as "unity gain" or "unity slope." The output of the linear hearing aid cannot exceed its saturation level, which we measure clinically as the Saturation Sound Pressure Level (SSPL-90) when we do an electroacoustical evaluation of the instrument.

Once the amplifier is saturated, the limitation in output is provided via some form of peak clipping, which occurs when the signal drives the amplifier or receiver beyond its limits. This results in a cutting or clipping off of the peaks of the amplified signal, converting the sine wave into a squared wave. Since information present in the original signal has been clipped and will not be present in the reproduced signal, a certain amount of distortion is unavoidable. Peak clipping not only removes some of the original signal, it also introduces increases in harmonic and intermodulation distortion. Distortion is often defined in terms of how faithfully a system reproduces a signal. In the case of peak clipping, energy ends up in the output signal that is different from, or was not present in, the input signal. Increases in distortion can have an impact on both sound quality and amplified speech recognition ability.

Many of the least expensive linear hearing aids are manufactured with Class A amplifiers, the several limitations of which are described by Killion (6). They are generally limited to low gain, low power applications, and while it is possible to get high fidelity response with one, its current requirements are so high that battery life is severely reduced, in some cases to less than 10 hours. In order to increase battery life, most hearing aid manufacturers use a "starved" Class A circuit, which results in significant high frequency distortion problems due to peak clipping. Class B push-pull amplifiers can produce more gain and better high frequency fidelity while using battery power more efficiently, but their size continues to be a problem: they are often too large to fit in today's smaller hearing aids.

When recommending linear hearing aids, a good option to consider is the Class D amplifier, which has made it possible to provide high fidelity amplification with increased high frequency response, higher output

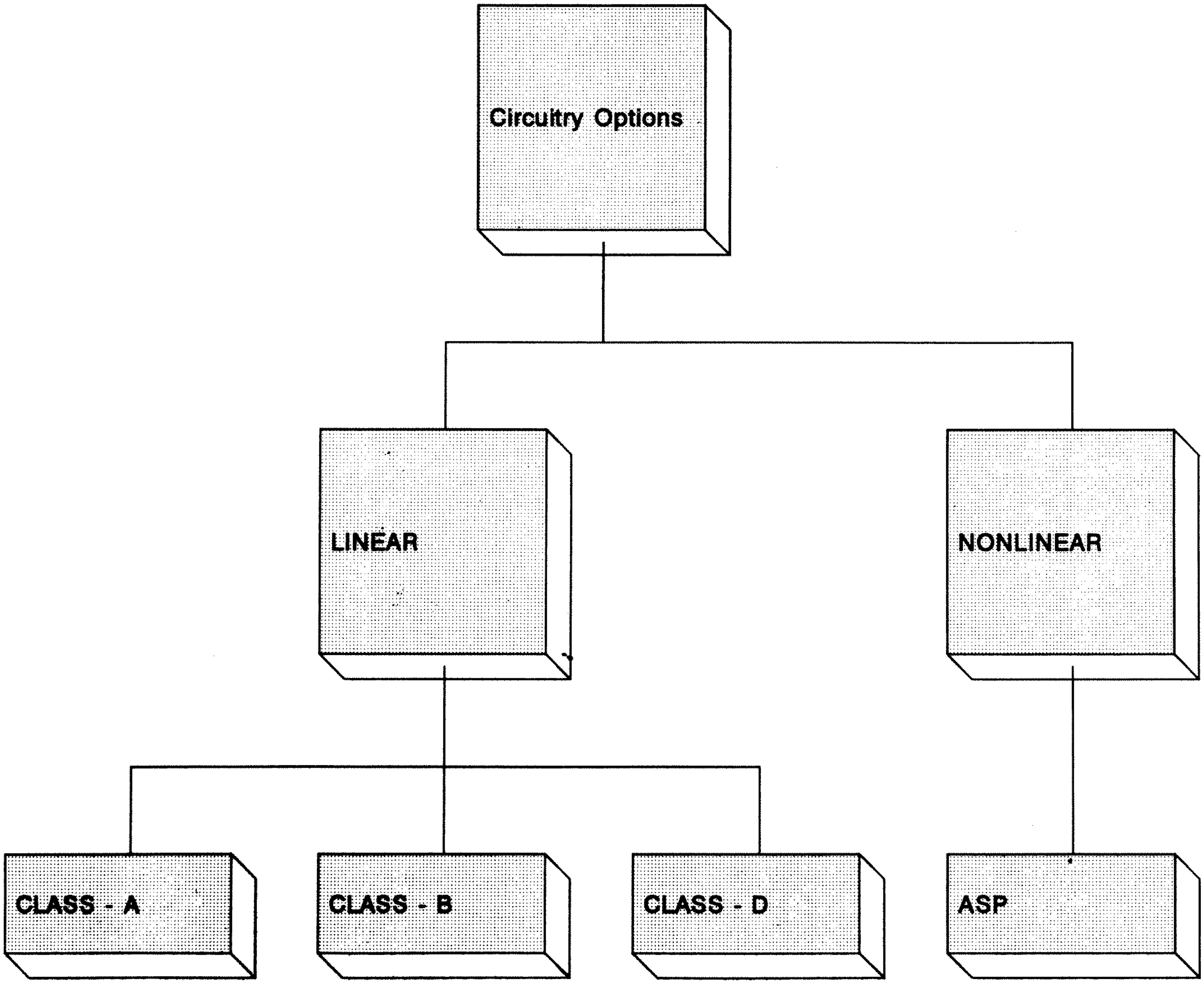

Figure 3.
Hearing aid circuitry decision.

saturation levels, and increased signal headroom while maintaining low battery drain. Space is saved with the Class D because the integrated output amplifier chip is built inside the receiver. A small preamplifier is added to the hearing aid circuitry, minimizing space requirements. The result is a smaller hearing aid capable of greater gain and maximum output with less distortion due to increased headroom (9). Killion states that this circuitry roughly doubles the cost of the receiver, an increase of about $50.00 to the wholesale price of the hearing aid (6). This option is well worth the additional expense when working with linear instruments.

NONLINEAR AMPLIFICATION

Compression

Most nonlinear amplification systems consist of some form of compression circuitry. Traditional compression instruments use a level-detecting device to monitor voltage at some location on the circuit (10); this location, relative to the volume control of the hearing aid, determines the classification of compression being utilized. The signal level is monitored if, prior to the volume control, the instrument is said to be an input compression hearing aid; if not, it is said to be an output compression hearing aid. With input compression, the input level in dB SPL at which compression is activated is independent of the volume dial setting; with output compression, the volume dial setting will affect the activation of the compression circuit.

Compression circuits can be described in terms of three primary characteristics: threshold or kneepoint of compression, compression ratio, and attack and release times.

Threshold of compression is the level of incoming signal just intense enough to trigger the compression function. **Figures 4a** and **4b** show the threshold of compression for two different types of compression hearing aids, at 80 and 65 dB SPL, respectively.

The input/output functions shown in **Figure 4** do not clearly demonstrate whether either device uses input or output compression. In order to make this distinction, it is necessary to determine the "kneepoint" of compression as opposed to the traditional "threshold" of compression. Compression kneepoint is the position on the input/output curve at which the function departs from linear (1:1) gain. That point contains both X and Y coordinates and allows the compression function to be described in terms of signal input and output (10). The kneepoint can be demonstrated graphically by obtaining input/output functions with a number of volume control settings as shown in **Figure 5. Figure 5a** depicts input/output functions of a compression device at three different settings with kneepoint coordinates of 75/95, 75/105, and 75/115. As volume control setting is changed, X remains the same, indicating that this is an input compression hearing aid. **Figure 5b** depicts input/output functions of a compression device at three different settings with kneepoint coordinates of 65/100,

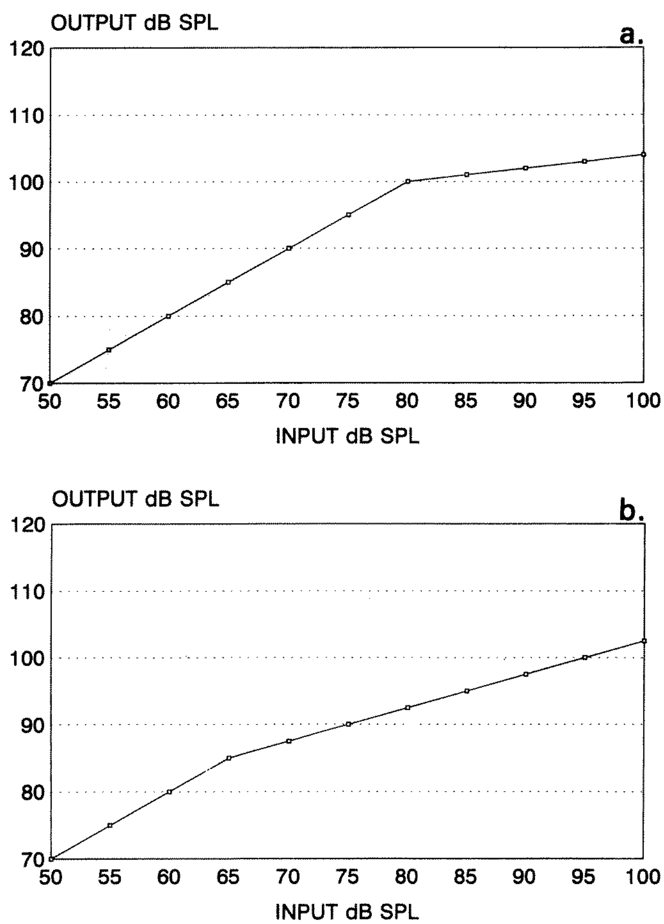

Figure 4.
Compression threshold of a) 80 dB SPL and b) 65 dB SPL.

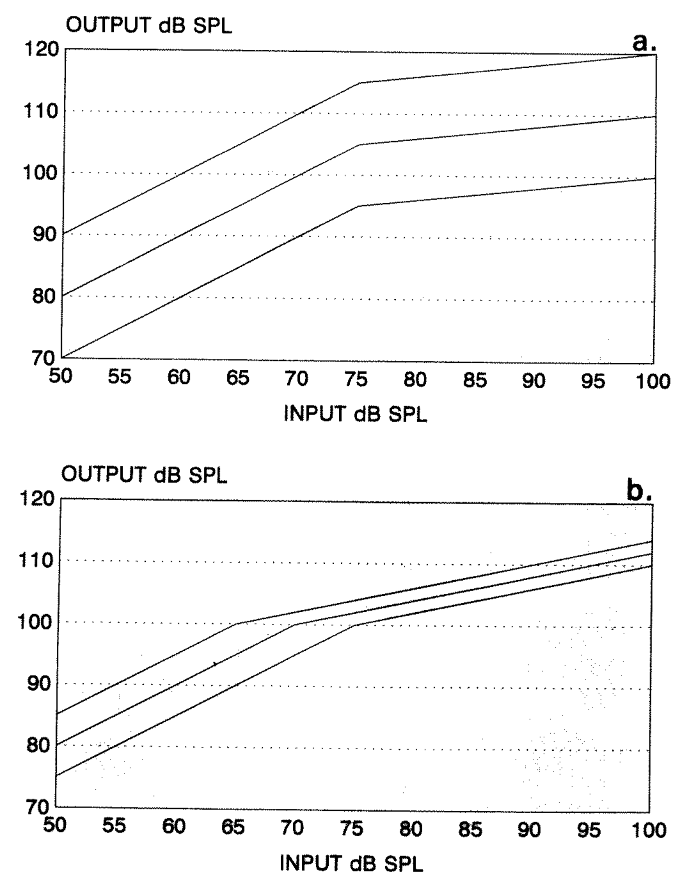

Figure 5.
Input/output functions of a) an input compression aid and b) an output compression aid—at three volume control settings.

70/100, and 75/100. As the volume control setting is changed, Y remains the same, indicating that this is an output compression hearing aid.

The compression ratio, determined by dividing the change in input level by the change in output level over a specified range of inputs, indicates the degree to which the signal will be compressed. For example, if an input signal changes by 10 dB and the output signal changes by 1 dB over the same range, the compression ratio is 10:1. **Figures 4a** and **4b** show compression ratios of 5:1 and 2:l, respectively. Less than 5:1 is generally considered to be a low compression ratio and 5:1 or greater is considered to be a high one.

Compression attack and release times refer to the time in milliseconds required for the compression circuit to be activated or deactivated. It is generally preferable for the attack time to be very short, usually <5 ms. The shorter the attack time, the more effectively the hearing aid wearer is protected from high output levels. Preferred release times are more difficult to estimate; if too short (<50 ms), an audible signal (pumping) may be created as the instrument goes in and out of compression. If too long (>150 ms), the user may miss important input signals during the recovery phase. Target values for ideal release times are, therefore, between 75 and 150 ms.

Many of the newer hearing aid designs allow the dispenser to adjust some or all of these characteristics. In addition, some hearing aid circuits incorporate release times that vary as a result of qualities of the input signal that drove the instrument into compression.

When to Pursue Nonlinear Amplification

There are a number of reasons for pursuing nonlinear amplification for a particular individual. Historically, nonlinear amplification, usually some form of compression, was selected when the person demonstrated significant tolerance problems: if the person exhibited high levels of recruitment, some form of automatic gain control (AGC) was chosen. If significant recruitment was not exhibited, linear amplification was recommended. Although it is still important to identify high recruitment levels, there is a growing body of evidence to suggest that nonlinear amplification may be appropriate for most, if not all, of our clients.

Berlin recently summarized research differentiating the functions of the outer hair cells versus inner hair cells. It appears that the majority of the sensory nerve endings in the mammalian ear come from the inner hair cells; the outer hair cells are primarily motor units (7). Dallos and Martin suggest that weak vibrations stimulate the outer hair cells, which amplify the vibrations sufficiently to stimulate the sensory inner hair cells, improving their sensitivity to soft sounds but not loud sounds (11).

Many of our clients will present with outer hair cell damage that can be verified through otoacoustic emissions, acoustic reflex testing, and measurements of recruitment. It is logical to attempt to fit these people with hearing aids that mimic, to some extent, the function of the outer hair cells. Providing amplification that varies as a function of input level can only be done with nonlinear amplification. This function can be approximated with K-Amp circuits (Etymotic Research, Inc., Elk Grove Village, IL), ReSound, and other programmable hearing aids that often employ Telex Adaptive Compression (variable release time) and have low kneepoints of compression at or below 50 dB SPL (7).

The amount of gain provided by a linear instrument may be appropriate for low level inputs but may prove to be too loud for more moderate level inputs and may produce excessive levels of distortion for high level inputs. Recently, there has been increased attention by researchers on incorporating individual loudness growth function data into the fitting process. Although the notion that matching hearing aid processing to a user's loudness growth function will result in greater benefits is still unproved (12), it appears to be a logical starting point for improving fitting. One such protocol has been proposed by the Independent Hearing Aid Fitting Forum (IHAFF) with a goal of setting amplification to normalize the relationship between environmental sounds and loudness perception (13). These procedures attempt to provide amplification that renders soft sounds audible, conversational level sounds comfortable, and loud sounds tolerable. In order to provide varying levels of amplification for different input levels, one must pursue nonlinear or automatic signal processing (ASP) amplification. The term ASP has been used by manufacturers and researchers alike to refer to many different functions: here it will be used to refer to any circuit that automatically changes the way incoming signals are processed as a result of their characteristics. **Figure 6** lists a classification system of ASP instruments suggested by Killion, Staab, and Preves (14).

As shown in **Figure 6,** most ASP instruments can be classified as either fixed frequency response (FFR) or level dependent frequency response (LDFR) systems. FFR systems maintain a constant frequency response shape regardless of input level, since all frequencies are compressed more or less equally. Traditional compres-

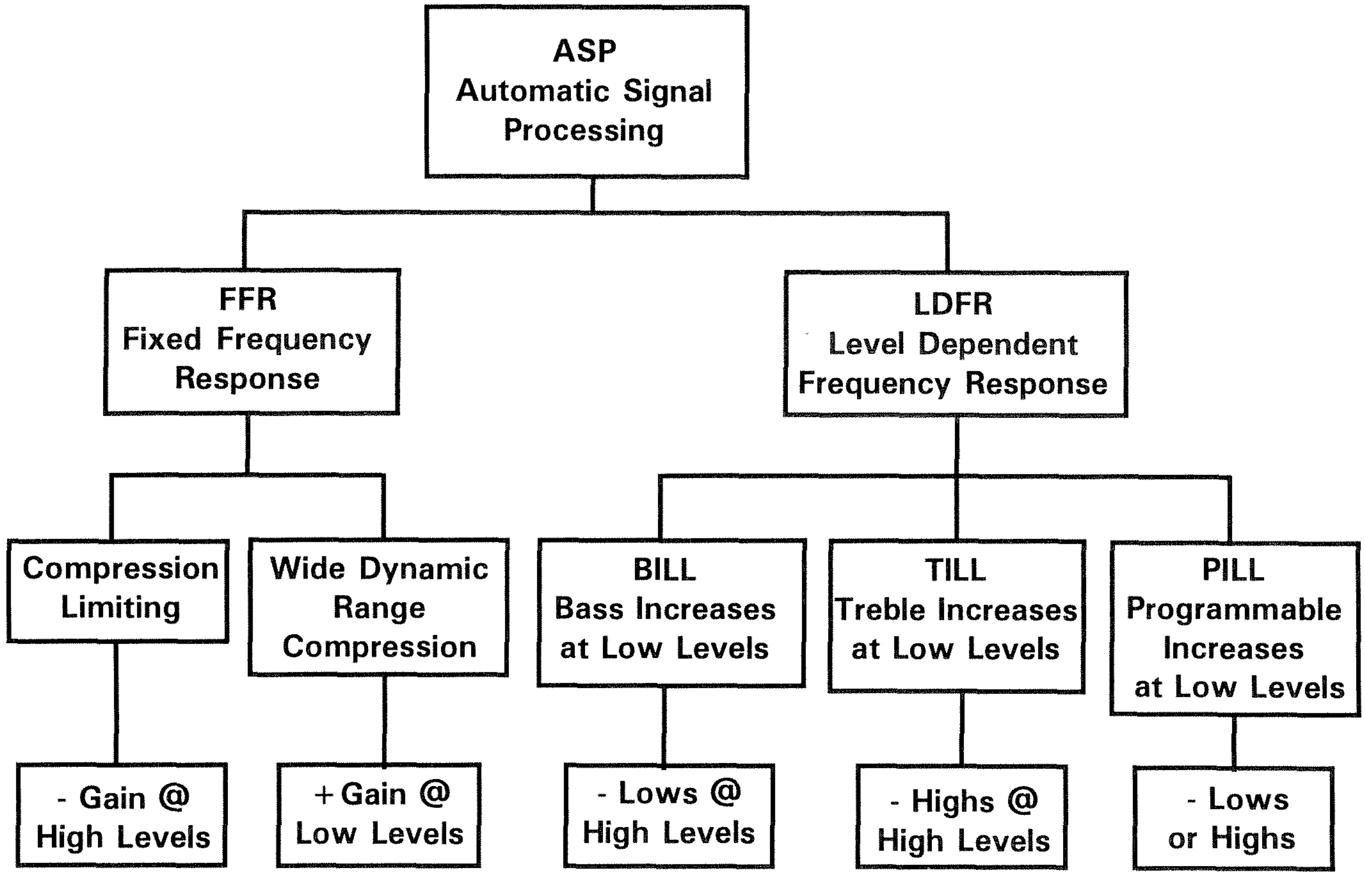

Figure 6.
Classification system of ASP instruments suggested by, and used with permission of, Killion, Staab, and Preves.

sion or AGC instruments fall under this category. LDFR systems provide frequency response shapes that vary depending on the input level as a result of a combination of compression and filtering.

Fixed Frequency Response Systems

Although all FFR systems maintain a fairly constant frequency response shape as compression is activated, there are several different compression circuits that can be chosen depending on the primary goal of compression. As shown in **Figure 6,** the two primary types of FFR circuits are compression limiting and wide dynamic range (WDR) compression.

Compression-limiting circuits are used to avoid peak-clipping problems associated with linear instruments. Typically, these circuits provide unity gain (1:1) until the kneepoint is exceeded and compression activated; they tend to have high kneepoints of compression and high compression ratios. **Figure 4a** is an example of one such circuit with a kneepoint of 80 dB SPL and a compression ratio of 5:1. As can be observed from this example, the input/output function of a compression-limiting circuit often looks very much like that of a linear hearing aid. These circuits are chosen for those persons who have complained about distortion from peak-clipping devices in a variety of settings, such as restaurants, places of worship, and work environments. They can also be useful in keeping the SSPL-90 below the uncomfortable loudness level (UCL) for those with mild tolerance problems.

WDR compression circuits are generally used with individuals with more severe tolerance problems. The term dynamic range refers to a person's range of usable hearing: the lower limit is the threshold, and the upper limit is his or her UCL. WDR compression circuits are most effective with persons with a marked reduction in dynamic range. Unlike the compression-limiting circuits, the WDR compression circuit has a low kneepoint with a low compression ratio and is activated for all but the least intense signals. **Figure 4b** is an example of a WDR compression circuit with a kneepoint of 65 dB SPL and a compression ratio of 2:1.

The decision as to which of these FFRs to pursue depends upon the needs of the user and the goal of the audiologist or dispenser. If the goal is to provide unity gain over a wide range of inputs while avoiding distortion with high level inputs, the compression-limiting circuit is the proper choice. If the goal is to provide amplification to an individual with significant tolerance problems and a reduced dynamic range, the WDR circuit is more appropriate.

Occasionally, hearing aids use a compression circuit that appears to combine dynamic range compression for low level inputs with compression limiting for higher ones. This is called curvilinear compression and is characterized by compression ratios that increase with input levels, beginning, perhaps, with a 1:1 ratio followed by changes to 2:1, 3:1, 5:1, and 10:1 as the input level increases.

Another approach that is becoming more common is multiple-channel compression. These devices have frequency regions each of which is processed separately: each channel has its own kneepoint, compression ratio, and attack and release times that can be adjusted independently of the other channel or channels. The advantage of this is to allow more frequency-specific compression to take place. For example, low frequency background noise may activate the compression circuit in that band while the high frequency band is not affected and high frequency speech cues maintained (15). The goal of such a circuit is to improve speech recognition ability in a background of noise. Multiple-channel compression circuits are used in a number of LDFR devices.

Level Dependent Frequency Response Systems

LDFR systems provide frequency response shapes that vary depending on the input level as a result of a combination of compression and filtering. As shown in **Figure 6,** the three primary types of LDFR circuits are bass increase at low levels (BILL), treble increase at low levels (TILL), and programmable increase at low levels (PILL).

A BILL device is an ASP instrument that provides a relatively broad frequency response in quiet settings. More bass response is provided for low level inputs than for the high. When the input level increases, a low-cut filter is activated and bass response decreased. BILL devices are based on the belief that reducing low frequency energy (background noise) will improve speech recognition ability in unfavorable listening situations. The Manhattan III (Argosy Electronics, Eden Prairie, MN) is an example of a popular BILL device.

The logic employed with TILL devices is the exact opposite of that of BILL devices. TILL circuits provide greater treble responses for low level inputs than for the high. That is, these instruments provide significant high frequency gain for soft sounds with minimal high frequency gain for loud sounds. At high intensity levels, the instrument utilizes a WDR circuit and becomes transparent (0 gain) for high frequencies. The K-Amp is the most commonly used TILL circuit (14).

The most versatile type of ASP comes from PILL circuitry that can provide either a BILL or a TILL response. In order to accomplish this, the instrument must have a minimum of two-channel compression. **Figures 7a, 7b,** and **7c** contain examples of BILL, TILL, and PILL circuitry, respectively.

FACTORS TO CONSIDER IN SELECTING CIRCUITRY

As shown schematically in **Figures 2, 3,** and **6,** hearing aid selection involves a series of decisions. **Figure 2** represents two of the earliest: one aid or two, and the style to be tried. In **Figure 3** it is linear or nonlinear amplification. If the decision is to pursue nonlinear, the possible options are shown in **Figure 6.** When pursuing ASP amplification, the first decision to be made is whether to employ an FFR or LDFR system, leading to a further decision regarding the type of circuit to employ (compression limiting versus WDR for the FFR instruments and BILL, TILL, or PILL for the LDFR instruments). There are a number of important factors to consider in making this decision.

Severity of Hearing Loss

The severity of the hearing loss may affect style as well as circuitry options. Although great strides have been made in incorporating more power into smaller aids, it is still true that individuals with severe and profound hearing losses may not be able to achieve adequate gain with smaller devices without running into acoustical feedback problems.

In terms of compression-circuitry options, persons with mild to moderate losses will often be able to use either compression-limiting or WDR compression circuits successfully. Individuals with moderate to severe losses require more gain for moderate and high level inputs and will, therefore, be more likely to be successful with a compression-limiting system. The WDR compression systems tend to provide too little gain for higher level inputs.

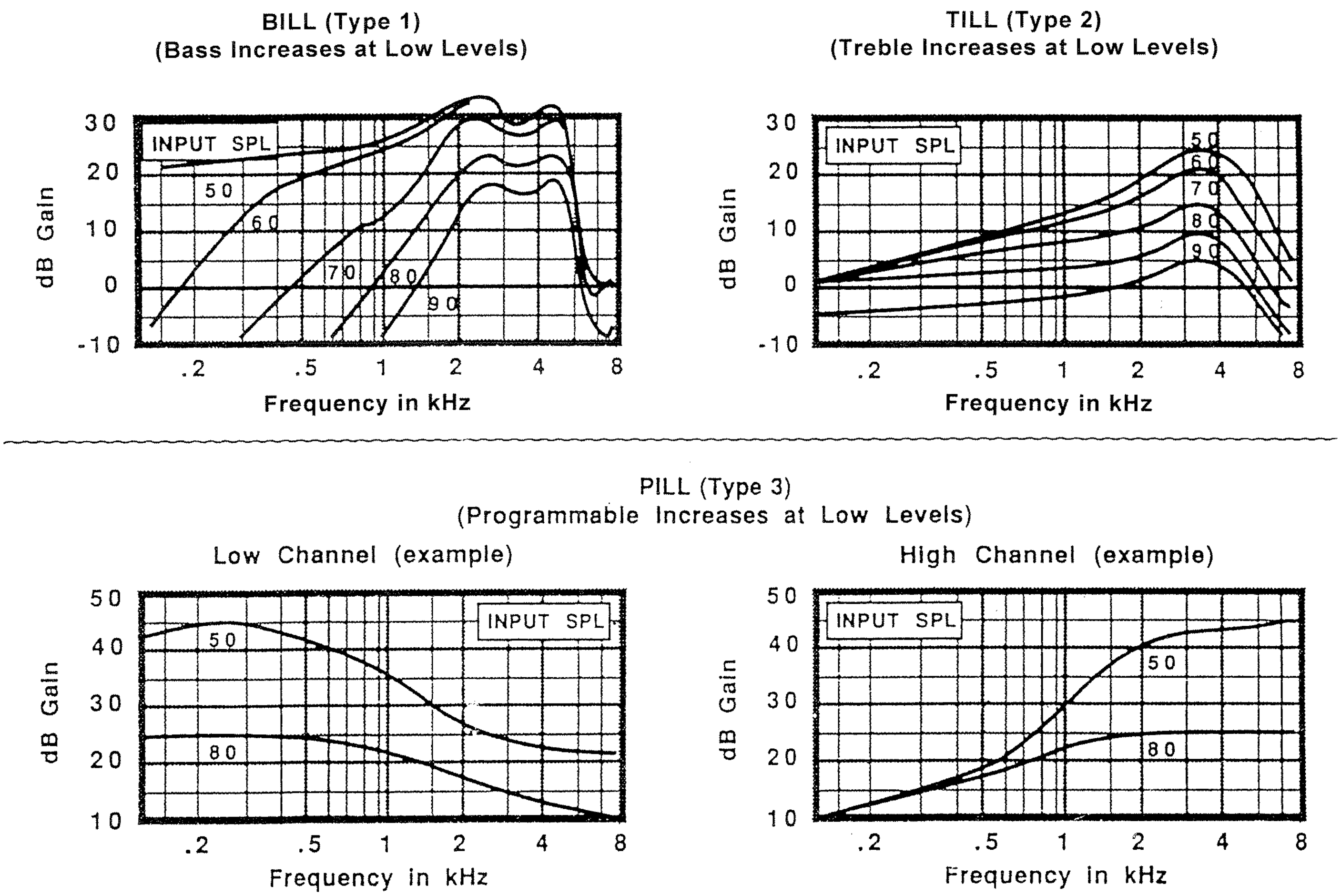

Figure 7.
Examples of BILL, TILL, and PILL circuitry.

Audiometric Configuration

Audiometric configuration can be an important factor, particularly when attempting to select a nonlinear circuit for a client. For example, many audiologists and hearing aid specialists have found that individuals with flat hearing losses perform very well with FFR systems. Of the LDFR systems, BILL circuits appear to provide the greatest benefit for these people. TILL circuits, such as the K-Amp, appear to be most successfully fit to those with mildly sloping high frequency hearing losses. Persons with precipitously sloping audiograms will probably perform most effectively with a compression-limiting circuit.

Dynamic Range

The user with a fairly wide dynamic range will most likely perform best with a compression-limiting system that functions similar to a linear instrument while avoiding its distortion at high level inputs. Despite the apparent contradiction in user and circuit descriptor, the person with a narrow dynamic range will probably perform best with a WDR compression unit. In this scenario, the instrument is operating in compression for the majority of inputs, keeping the output of the instrument within the user's narrow dynamic range.

Prior Hearing-Aid Experience

Previous hearing-aid experience often plays a major role in the success of a fitting. A wise first step in selecting and fitting an experienced user is to determine what he or she liked and did not like about the previous instrument. Reactions to style, lack of power, too much power, feedback, and the like can be most helpful in selecting instruments as well as in the counseling that will accompany and follow the actual hearing aid fitting.

If the individual was generally unhappy with the previous instrument, it is advisable to try to ascertain the source of that dissatisfaction and try a different approach with the new instrument. For example, complaints about high levels of distortion with an old, linear hearing aid, may mean a switch to a compression-limiting or WDR device may prove successful.

People who were generally happy with their previous hearing aids usually seek an instrument that performs in a similar fashion. If the user is replacing an old, linear hearing aid with which he/she was fairly happy, a new linear or compression-limiting device will probably produce positive results. WDR compression devices are often viewed by experienced users as providing too little gain. For this reason, many of those for whom a K-Amp may be theoretically appropriate will complain that the instrument is not powerful enough.

Another problem the audiologist may encounter is user resistance to an instrument with no volume control wheel. Most CIC instruments and some other compression systems, such as the MultiFocus (Oticon Corp., Somerset, NJ) do not have them. While these devices may provide many benefits to users, adjusting to the lack of direct control is often difficult for them and may result in rejection of that particular instrument.

Listening Environments

This article focuses on signal processing at average conversational levels; adverse listening conditions are detailed in the following article by Elizabeth Kennedy, Ph.D. However, it is worth noting that the variety of listening environments will affect the selection of circuitry options. For anyone with a wide array of listening environments, an LDFR system should be considered. BILL or TILL circuits can be chosen for listeners who occasionally find themselves in poor listening environments. The decision as to BILL versus TILL will be based upon audiometric configuration, severity of loss, and theoretical preference on the part of the dispenser. If the user reports frequent change in listening environment, a PILL system with multiple memories is a logical choice: these come with 2, 3, 4, or 8 memories and produce dramatic variations in frequency response at the push of a button located on the hearing aid or on a remote.

SELECTING GAIN CHARACTERISTICS

There has been a recent resurgence in prescriptive hearing aid selection techniques based on the notion that frequency response and gain characteristics of aids should attempt to compensate for the characteristics of a given hearing loss. The purpose of all of the prescriptive formulae is to provide the clinician with frequency-specific target gain values that represent a starting point for finding the ultimate amplification to be provided. From this point, finer adjustments can and should be made to the frequency response and gain characteristics of the hearing aid according to the needs of the user. The formulae, therefore, provide a structured approach to arriving at, or approximating, initial parameter settings.

Prescriptive procedures have been designed with linear hearing aids in mind, seeking to maximize speech intelligibility through linear amplification for one particular situation, that of average conversational speech in relative quiet (16). Prescriptive formulae, such as the NAL-R (17), can be used effectively for selection of frequency response characteristics for linear hearing aids and also for compression-limiting amplifiers that act very much like linear ones over the majority of the input/output function. A significant limitation, however, is that these procedures do not incorporate loudness-based data, such as that described in the IHAFF approach. That is, a particular formula may be appropriate for average conversational level input, but may also render low level inputs inaudible and high level inputs too loud (16).

Killion and Fikret-Pasa have developed a triple-target method called FIG6 for selecting gain for a particular WDR compressor, the K-Amp, that is loudness-based and appears to be particularly effective with listeners with mild to moderate impairment (18). The goals of the FIG6 approach are in concert with those of the IHAFF. The desired gain in the real ear is determined for three different input levels: soft (40–50 dB SPL), moderate (65–70 dB SPL), and high input (90 dB SPL).

The goal for low level input signals is audibility. FIG6 attempts to provide the amount of gain necessary to produce 20 dB HL sound-field thresholds for mild to moderate losses. For losses exceeding 60 dB HL, an approximate one-half gain rule is used, resulting in gains of 45 dB for a 70 dB HL loss and 50 dB of gain for an 80 dB HL loss, and so forth.

Moderate input levels are assumed to be representative of conversational-level speech. Therefore, the goal is for these signals to be perceived as comfortable for the listener with hearing impairment. The FIG6 formulae provide target gain values for moderate level inputs based upon Pascoe's measures of Most Comfortable Loudness (MCL) as a function of hearing loss (19).

The goal for high level input is perception of the signals by the wearer of a hearing aid as loud without exceeding loudness discomfort levels (LDL). The required gain was estimated from the loudness-growth data of Lyregaard

(20) and Lippman et al. (21). Software to compute FIG6 target gain values is available from Etymotic Research.

Cornelisse, Seewald, and Jamieson (22) recently described the desired sensation level [input/output] (DSL[i/o]), which is another procedure for prescriptively fitting nonlinear hearing aids. Similar in format and implementation to the DSL 3.0 (23) and the DSL 3.1 (24), which have been used for years in fitting amplification devices to children, the DSL[i/o] is specifically designed for fitting WDR compression hearing aids. The DSL[i/o] procedure uses measures of, or estimates of, the sound-field to ear-canal transfer function for a given user, along with the standard threshold of audibility, and the thresholds and upper level of comfort of the individual with hearing impairment, in order to arrive at target desired sensation levels. The DSL[i/o] formula provides a series of targets for a variety of input levels ranging from 50 to 90 dB SPL. The logic of this procedure is in concert with the previously described IHAFF approach. On this basis, the DSL[i/o] provides the clinician with theoretically sound first-order approximations of amplification settings when working with WDR compression hearing aids.

Using a Matrix

In order to select circuitry options and to specify electroacoustic characteristics as measured in a 2 cc coupler, most manufacturers provide some type of matrix, generally consisting of a set of three numbers representing, respectively: maximum output, gain, and slope. The maximum output is the maximum peak from the SSPL-90 curve and it is usually observed in the 2,000–3,000 Hz region. Depending upon manufacturer, the gain value from the matrix is either the high frequency average gain (HFA) from the full-on gain curve or the peak gain value from the same curve. Slope is most often determined by taking the decibel difference between the peak gain (also usually observed in the 2,000–3,000 Hz region) and the gain at 500 Hz (representing a low frequency). The larger the difference between these two measures, the greater the high frequency emphasis of the circuit. A matrix of 107/28/15 would, therefore, indicate a maximum SSPL-90 of 107 dB SPL, a peak gain or HFA gain of 28 dB, and a slope of 15 dB.

VERIFICATION OF AIDED PERFORMANCE

Speech Audiometry

A number of tests employing speech stimuli have been utilized in hearing aid evaluations. For example, aided and unaided speech reception tests have often been compared to arrive at a single number representing speech gain. Speech signals have also been used at high intensity levels in order to obtain aided and unaided LDLs. Historically, however, there has been a decided emphasis on word recognition scores (25).

Word recognition scores have been widely criticized in the literature for the past 30 years, beginning with the article by Shore, et al. (26). The feeling of a significant number of researchers is that no particular speech test or stimulus has been demonstrated to be a valid or reliable predictor of performance with a hearing aid. Harford states that word recognition scores, as obtained traditionally in the audiology clinic, do not predict with any degree of accuracy how an individual will function with a hearing aid in everyday situations (27). The reason for this poor predictability is the large array of uncontrollable variables, including the acoustic environment, the alertness of the individual, his or her ability to use visual cues, and the clarity of the speech produced by the speaker.

Despite these criticisms, some researchers continue to support the use of speech signals for limited purposes. Humes, for example, acknowledges the limitations of speech stimuli for selecting from among several instruments, but supports the use of speech materials in assessing aided performance of a particular device (28).

Functional Gain

Functional gain, defined as the difference between unaided and aided sound field thresholds for warble tones or narrow bands of noise, has been demonstrated to be an effective alternative to tests relying upon speech stimuli (29). The use of warble tones or narrow bands of noise allows frequency-specific values to be obtained, most often from 250 to 6,000 Hz. These measures serve the important function of verifying the electroacoustic parameters derived from the various prescriptive formulae.

Several limitations of functional gain measurement should be noted. Obviously, just as in speech audiometry, this is a behavioral measurement and subject to variables related to client participation. Also, gain values are obtained only at discrete frequencies with no information provided about what occurs between these frequencies. Finally, obtaining functional gain can be a fairly time-consuming process.

Real-Ear Measurements

Techniques are presently available that can eliminate or greatly reduce these problems with functional gain measurements. The development of probe-tube microphone systems has enabled us to make accurate measurements of aided performance in the ear (real-ear measurements), overcoming the well-documented inadequacies of previously available coupling systems at representing the frequency response characteristics of a hearing aid in the human ear (30,31).

In order to discuss real-ear measurements, one must understand the associated terminology; four of the most useful definitions may be summarized as:

1. Real-ear unaided response (REUR), which represents the open ear canal measurement in dB minus the input signal level in dB.
2. Real-ear aided response (REAR), which represents the aided ear canal measurement in dB minus the input signal level in dB.
3. Real-ear insertion response (REIR), which represents REAR in dB minus the REUR in dB across a range of frequencies. When looking at a specific frequency, this measure is referred to as real-ear insertion gain (REIG).
4. Real-ear saturation response (RESR), which represents the dB SPL generated in the ear canal with a hearing aid turned on and receiving an input stimulus sufficiently intense to operate the hearing aid at its maximum output level (32).

REUR represents the combined effects of ear canal resonance, external ear resonance, and head diffraction effects of an individual. The REAR represents the total response of the hearing aid system, including the REUR, and is equivalent to an *in-situ* gain measure. REIG represents the amount of amplification provided by the hearing aid system alone and, as such, is an electroacoustic estimate of the functional gain (33). It is not surprising, therefore, that REIG and functional gain have been demonstrated to be highly correlated (34).

These real-ear measurements, REIR in particular, have proven to be useful in terms of verifying the electroacoustic parameters derived from the various prescriptive formulae. Verification of aided performance has come to mean comparing the obtained REIR with target gain values determined via a prescriptive formula, such as the NAL-R. A common goal utilized in many clinics is to match the two within ±5 dB for the frequency range of 250 to 4,000 Hz (35). When working with linear, peak-clipping instruments, there will be minimal differences between functional and insertion gain and this verification strategy works very efficiently. However, when working with nonlinear compression instruments, several limitations must be kept in mind.

Fabry points out that results of functional gain testing and real-ear measurements will vary substantially when evaluating compression hearing aids with low thresholds or kneepoints. As mentioned previously, functional gain values are typically obtained by comparing aided versus unaided thresholds to warble tones or narrow bands of noise. These aided responses are generally obtained to low level stimuli, often in the range of 20–30 dB HL. These input levels are below the thresholds for compression instruments and, therefore, represent results obtained with significantly more gain than will be evidenced with higher input levels. It is for this reason that the functional gain value is said to overestimate the amount of gain provided by nonlinear hearing aids at conversational speech levels (36).

On the other hand, real-ear measurements are generally obtained with input levels of 65–70 dB SPL, levels that exceed the thresholds of compression for many instruments. This is particularly true for WDR compression circuits. The result will be a significant reduction in REIG, compared with functional gain. Fabry states that neither measure is right or wrong and suggests that functional gain is an effective estimate of amplification of soft sounds and REIG more accurately estimates the gain for more typical input level signals (36).

Real-ear measurements to verify aided performance with nonlinear circuits must, therefore, be employed cautiously. The manner in which real-ear measurements are obtained with nonlinear circuits will vary depending upon the goal of the audiologist. If the goal is to determine how accurately the REIR can match target gain, the input level of the broadband noise should be below the kneepoint of compression (40–50 dB SPL), based upon the fact that the prescriptive formulae were developed with linear amplification in mind. The best match with a compression circuit is likely to be obtained when it is functioning in a linear fashion (before compression is activated). On the other hand, if the goal of the audiologist is to determine insertion gain for standard conversational level speech, the input level of the broadband noise should be between 60–70 dB SPL. The clinician must keep in mind that the compression circuit may be activated with this input level and, if so, the obtained REIR will not provide a good match with the target gain from the prescriptive formula.

A good practice to develop when measuring REIR with nonlinear aids is to employ a variety of input levels (i.e., 40, 60, and 80 dB SPL). This allows you to achieve both of the aforementioned goals as well as obtaining verification of the compression action. For example, with a WDR compression aid having a low kneepoint of compression, the best match to target gain may be obtained with the 40 dB SPL input. If this instrument also happens, as does the K-Amp, to utilize a TILL circuit, a high frequency emphasis will be observed for the lower level inputs (40 and 60 dB SPL) and the response will become flatter for the higher level input (80 dB SPL). In the specific case of the K-Amp, it can be observed that the instrument becomes transparent (0 dB gain) for higher level inputs. For instruments using BILL circuits, once again the best match between target and insertion gain may take place with the lower level inputs (40 or 60 dB SPL). As the input level is increased, the frequency response of the BILL instruments shifts to a higher frequency emphasis (or de-emphasis of low frequencies). Thus, the performance of the various nonlinear circuits can be verified effectively through real-ear measurements if the time is taken to use a variety of input levels in measuring REIR.

Although it has not received the attention of the REIR measure in the past, most clinicians are becoming increasingly aware of the importance of measuring RESR with each user or prospective user. By measuring the RESR and making sure that it occurs below the individual's LDL, we can more accurately provide a comfortable and safe amount of amplification, The RESR is measured in basically the same way as the REAR, except that the hearing aid is in saturation, accomplished by using a 90 dB SPL input signal and adjusting the hearing aid volume to a point just below acoustical feedback (37). Although a broadband complex noise is generally recommended for most real-ear measures, it has been shown for measures of hearing aid maximum output that the use of pure-tone signals is the most conservative approach (38).

The RESR and the user's LDL values for specific frequencies can be used in a fashion that is analogous to the REIR and target gain matching procedure described above. The first step is to use the real-ear system to determine the SPLs generated in the ear canal that correspond to the LDLs for specific frequencies. These SPL values then become the target values for the RESR adjustment. That is, the instrument should be set so that the RESR does not exceed, or is about 5 dB less than, the LDL values at any frequency. If the output curve of the hearing aid and the LDL values follow a similar pattern across frequencies, this adjustment will be effective and relatively simple. In many cases, however, a sharp peak in the RESR will require adjustments that markedly reduce gain well below the RESR at other frequencies. As multichannel programmable hearing aids become more widely used, however, the match between RESR and user LDL across frequency will become easier to achieve (37).

RECENT ADVANCES IN HEARING AID OPTIONS

Although the topic of hearing aid styles has been covered briefly in this chapter and more extensively in another chapter in this book, there are two recently available types that deserve further discussion with regard to selecting hearing aid circuitry: CIC and programmable instruments.

Completely in the Canal

Most professionals would agree that when it comes to selecting a hearing aid for our clients, particularly first-time users, their preference will be the smaller the better, a preference based almost exclusively on cosmetics. Although cosmetic issues will always play a major role in hearing aid selection, the CIC aids yield a number of additional significant advantages.

CIC aids are often defined as instruments whose lateral portion terminates 1 to 2 mm inside the meatal opening (39). However, it is important to define the medial portion of the instrument as well: the most significant acoustical advantages will be demonstrated when the CIC is placed deeply in the canal and its medial portion terminates within 5 mm of the tympanic membrane. The deep placement allows the natural resonant characteristics of the pinna and concha to occur, resulting in increased high frequency input to the microphone and thus increased high-frequency output in the ear canal. Overall amplified output is also increased, because the length of the instrument results in a reduced residual ear canal volume between the instrument and the membrane. A smaller residual volume results in increased sound pressure levels being generated. Saturation-induced distortion can also be reduced because greater real-ear output is achieved with less amplifier gain (40).

Another significant advantage of CIC instruments is a reduction of the hearing aid occlusion effect. The occlusion effect is often defined as an enhancement of bone conduction sensitivity resulting from a closing off or occluding of the ear canal. The negative impact of the occlusion effect is often seen in the clinic when a new user complains about the sound of his/her own voice, saying "I sound like I'm talking with my head in a barrel," or "My voice has an echo." The key factor in reducing this effect with CICs appears to be the length of the instrument (41). The longer the CIC, and the further into the osseous portion of the ear canal it reaches, the greater will be the reduction in the occlusion effect.

There are numerous practical advantages to CICs as well: deep-fitting CICs are barely noticeable when being worn, and this factor alone will provide significant user satisfaction. In addition, these instruments tend to be comfortable to wear, very easy to insert and remove, and actually present fewer wax problems than ITCs and ITEs because the canal opening terminates beyond the major areas of cerumen production in the external auditory meatus. Improved localization ability as a consequence of microphone location as well as reduction in wind noise are two other possible advantages.

In order to work successfully with CIC instruments, it is critical to obtain accurate, deep impressions of the external auditory meatus. Staab and Martin (42) present detailed instructions on how to properly and safely obtain deep-canal impressions. In my opinion, a video-otoscope is a mandatory piece of equipment in this area: it allows the clinician to observe the condition of the external auditory meatus most effectively prior to taking the impression, as well as to view the position of the oto-block once it has been placed in close proximity to the tympanic membrane. The video-otoscope provides both the greatest efficiency and the most safety for this procedure.

There are a number of circuitry options available with CIC instruments. Many can be obtained with linear circuits, with some form of AGC or with an LDFR circuit (such as a K-Amp). Selecting a circuit with a CIC instrument is of particular importance, because the majority of these devices do not have volume control wheels, in which case it is advisable to select some form of nonlinear circuit so that the user will be able to achieve comfortable listening levels for a variety of input levels. The K-Amp has become a particularly common circuit choice in CIC fittings and comes standard on several manufacturer's models.

The lack of a volume control wheel may be problematic, particularly with experienced users accustomed to a certain amount of control. Several CICs now meet this need: one provides a magnetic volume control option, while several others have incorporated a control option into the string trimmer. A linear circuit may be a more feasible option in conjunction with some form of volume control.

Programmable Hearing Aids

It is beyond the scope of this chapter to discuss all of the programmable hearing aid devices currently available. It is important, however, to provide a cursory discussion of them, because they will constitute an increasing percentage of fittings in the future. In addition, programmable hearing aids represent an efficient way, and in some cases the only way, to implement some of the LDFR circuitry (BILL, TILL, or PILL) described above.

Most programmable hearing aids on the market today perform all processing functions via analog components but can be digitally programmed by the dispenser. For this reason, these instruments were originally referred to as digitally controlled analog (DCA) hearing aids or quasi-digital hearing aids. Today, they are simply referred to as programmable hearing aids.

There are many advantages to programmable hearing aids over analog instruments. Digital programming allows for more efficient and precise electroacoustic setting in terms of matching target gain and target SSPL-90. In those instruments with more than one channel, the audiologist is able to set the frequency/gain response in more than one frequency region independently of adjacent frequency regions. In addition, gain and output can be adjusted with significant flexibility over time if necessary (e.g., as with a dramatic change in hearing sensitivity). Some ASP, LDFR circuits, such as PILL, are available that are not possible in nonprogrammable instruments. Also, with programmable aids having multiple memories, the user has instant access to different types of signal processing and frequency response characteristics for different listening conditions otherwise impossible to achieve.

Programmable hearing aids are often desirable not because they are programmable but because they offer a better way to accommodate the amplification needs of a given user (45). In addition, because needs vary so dramatically, it is important to realize that no one programmable model, no matter how flexible in terms of acoustical performance, will meet them all (44). In order to match a particular user's needs to a given programmable instrument, of which there are currently more than 25 available, it is helpful to divide the aids into groups or tiers based upon the number of channels and memories each has. In his classification scheme, Mueller refers to all programmables with a single channel as Tier I instruments and those with two or three as Tier II (43). Using this model, the dispenser ends up with four categories: Tier I with single/multiple memories; Tier II with single/multiple memories. Bray's similar four-category classification scheme of specific instruments (45) is reproduced in **Table 1.**

The number of channels and memories can, therefore, serve as first order approximations in selecting the appropriate programmable instrument. More than one channel provides more frequency-specific control of amplification and compression parameters. A person with a precipitously sloping audiogram would probably benefit from a device with at least two channels in order to provide high frequency emphasis. A multiple-memory device provides instant access to dramatically different signal processing and frequency response characteristics. This type of system is particularly useful with those who frequently find themselves in a variety of listening environments.

The scheme shown in **Table 1** is particularly useful in reducing the pool of the most sophisticated (Class IV) instruments. Of the 25 plus programmable aids available, only 5 have multiple channels and multiple memories. The choices are, therefore, limited if multiple channels and multiple memories are required for a given individual.

Once the number of possible selections has been reduced, several other factors need to be considered in selecting the appropriate programmable instrument. One is the type of signal processing used: the ASP circuitry can be primarily compression-limiting or WDR compression. The same decisions needed for nonprogrammable aids and their types of compression are necessary for programmable instruments as well. To summarize, a person with a fairly wide dynamic range who requires a significant amount of amplification will probably perform best with a compression-limiting ASP circuit, and a person with significant recruitment and a reduced dynamic range will probably perform best with a WDR compression circuit.

Most, but not all, programmable instruments are available in ITE and/or ITC styles, and this may be a critical issue for some users. Several programmable instruments will only operate via a remote control, and while many people like remotes and realize some dexterity advantages with them, others prefer not to carry around yet another piece of equipment (46).

ADDITIONAL OPTIONS

Directional Microphones

The directional microphone hearing aid is designed to amplify signals originating from in front of the user to a greater extent than those from the rear. These devices gained significant popularity during the late 1970s but

Table 1.
Programmable hearing aid classification system summary.

	Single Memory (b=1)	Multiple Memories (b>1)
Single Channel (a=1)	**Class I Systems** Audiotone 2000 Bosch EUROSTAR Ensoniq SOUND SELECTOR Maico/Bernafon PHOX Qualitone CYBER I Rexton HORIZON Siemens INFINITI Starkey RESOLUTION Telex ARTISAN	**Class II Systems** Maico/Bernafon GAMMA [1/2] Phillips FARO [1/3] Phonak PICS [1/3] Qualitone CYBER II [1/2] Siemens INFINITI 3 [1/3] Starkey TRILOGY II [1/3] Unitron SIGMA [1/2] Widex DUO [1/2] Widex LOGO [1/2] Widex QUATTRO [1/4]
Multiple Channels (a>1)	**Class III Systems** Oticon MULTIFOCUS (2/1) Phillips GALAXY (2/1) ReSound ENCORE (2/1) 3M SINGLE-PRO (2/1)	**Class IV Systems** Danavox AURA (3/4) ReSound PHS (2/2) Siemens 2004 (2/4) Siemens 3004 (3/4) 3M MULTI-PRO (2/8)

are recommended much less frequently today, due in part to the high percentage of ITE and ITC instruments being dispensed. Directional microphones have been shown to perform better than omnidirectional ones in many studies using BTE instruments (47). Many audiologists have not considered them for ITE and ITC models because the microphone location on these styles provides a directional effect, rendering the directional microphone unnecessary. Mueller and Hawkins have recently demonstrated significant directional effects of directional versus omnidirectional microphones in ITEs (48). Although the amount of rear attenuation is less than that observed with BTEs, the directional effects, particularly in the low frequencies, may provide a positive impact on signal-to-noise ratio for a person in a real-world listening environment. Directional microphones are presently available on many BTE models, some ITE models, and as options on several programmable instruments.

Active Tone Control

Active tone controls are screwdriver-controlled potentiometers that allow the audiologist to adjust the shape of the instrument's frequency response. This is particularly useful when fitting a hearing aid and attempting to match measured insertion gain with predicted target gain. Active tone controls have been popular in BTE models for years, and the combination of a Class D amplifier, with its higher frequency-resonant peak, with active tone control has made it possible to achieve good agreement between insertion and target gains in ITE and ITC instruments.

Telecoil

It is beyond the scope of this chapter to discuss in depth the properties of a telephone induction pick-up coil or telecoil. The telecoil consists of a core of metal around which is wound a coil of wire. This system is used to pick up and amplify magnetic leakage from telephones and many assistive listening devices, such as inductive loop systems. The need of a given individual to use the telephone or other assistive devices will be critical in selecting a telecoil option or in selecting a particular instrument. For example, several programmable hearing aids have no telecoil option, which might eliminate the device for a person who uses the telephone or an assistive device often.

PROS AND CONS IN SELECTION CHOICE

In considering the pros and cons of the circuitry selection choice, it is necessary to return to the model shown in **Figure 1,** stressing that hearing aid selection and fitting needs to be viewed as a part of the overall audiologic rehabilitation process. The individual needs of the user must be determined first and foremost. The pros and cons in our selection will be based upon how well we have met the needs of a particular hearing aid user.

By way of a general summary, a number of scenarios will be stated for the categories of circuitry selection options covered in this chapter. Possible pros and cons (or advantages and disadvantages) of each type of hearing aid fitting will be listed for each category. The assumption is made that all fittings are binaural and that the person has been able to use the hearing aid style with which he/she is happiest.

Linear

A 60-year-old individual with a long-standing moderate to severe, bilaterally symmetrical, sensorineural hearing loss decided to pursue amplification. Measures of LDL revealed minimal tolerance problems and a fairly wide dynamic range. This person was fitted with binaural ITEs with linear amplifiers. The instruments included Class D amplifiers and active tone controls which allowed the audiologist to obtain an excellent match between target gain based on the NAL-R formula and measured insertion gain. The person uses the instruments in what he describes as difficult listening situations and is very pleased with his relatively inexpensive solution to his communicative difficulties. The downside of this circuitry choice includes the probability of significant distortion with high level inputs with a peak-clipping device. In addition, these linear instruments will provide unity gain for this user and will not vary the amount of amplification based upon input level until the instruments saturate. Although the person appears satisfied, it is certainly possible that his level of satisfaction would be even better with an instrument with some type of ASP circuitry.

Automatic Signal Processing: Fixed Frequency Response

A 65-year-old female client with a moderate, flat, bilaterally symmetrical sensorineural hearing loss had been using linear hearing aids for 8 years. She stated that she was reasonably happy with her aids, but they were occasionally noisy and had been sent for repair three times. Testing of LDL revealed some minor tolerance problems and a somewhat reduced dynamic range. This woman was fitted with binaural ITEs with compression-limiting circuits. In choosing between compression limiting and WDR compression FFR circuits, compression-limiting was the logical choice because of her experience with linear amplification and only somewhat reduced dynamic range. She reported significant satisfaction with the instruments. It is likely that a WDR compressor would have provided too little amplification for this experienced user. If she had a more significant tolerance problem and a more dramatically reduced dynamic range, the WDR compressor would have become a more logical choice.

Automatic Signal Processing: Level Dependent Frequency Response

A 52-year-old male attorney decided to pursue amplification for the first time. Audiometric testing revealed a bilaterally symmetrical, mild to moderate, gently sloping high frequency hearing loss. Test of LDL revealed some tolerance problems with a marked reduction in dynamic range in the high frequencies. This individual was fitted with binaural CIC instruments with K-Amp II circuits and is extremely happy with the instruments. High on the list of advantages he cites is the cosmetic acceptability: these instruments are barely visible when worn. In addition, the K-Amp circuitry provides peak gain around 2,800 Hz and extended high frequency response to 16,000 Hz (46). This TILL circuit also provides maximum gain for low level inputs and becomes transparent for higher level inputs, thus mimicking the function of the outer hair cells. Possible disadvantages are that the circuitry can be somewhat noisy for individuals with very good low frequency hearing. Although it is not a disadvantage for this case, many clinicians report that this type of TILL circuit does not provide enough gain for experienced users. If given enough time, many will acclimate to this form of amplification but many give up prior to acclimatization.

Programmable Instruments

A 70-year-old woman with a long-standing bilaterally symmetrical, moderate to severe, sensorineural hearing loss wished to replace her old linear ITE hearing aids. She reported that she had been saving her money for several years in order to obtain new aids with memories. She also stated that she spent several nights per week working in her son's noisy restaurant, one night per week playing bridge with her quiet friends, and one night per week playing poker with her rowdy friends. She specifically requested some form of amplification that would provide her with instant access to different frequency responses. Measures of LDL revealed minimal tolerance problems and a fairly wide dynamic range. The client was fitted with a binaural programmable hearing aid with a single channel and four memories (Class II in **Table 1**). Several weeks later she reported significant satisfaction with the new instruments, citing the instant accessibility of the four frequency responses as their greatest advantage. Additional advantages included a directional microphone and a powerful telecoil that she utilized often. Although she was quite happy, possible disadvantages with this system include the use of a remote control, a pro or a con depending upon the individual user. In addition, this system is quite expensive, literally requiring this client to save up for a significant length of time. Also, this instrument, although programmable, provides linear amplification and will certainly be inappropriate for individuals with reduced dynamic ranges.

Once again, the pros and cons of all of these circuitry decisions will vary from person to person and our measure of success of fit should be based upon meeting the needs of each individual user.

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