THE COMPRESSION HANDBOOK

THIRD EDITION

An overview of the characteristics and applications of compression amplification

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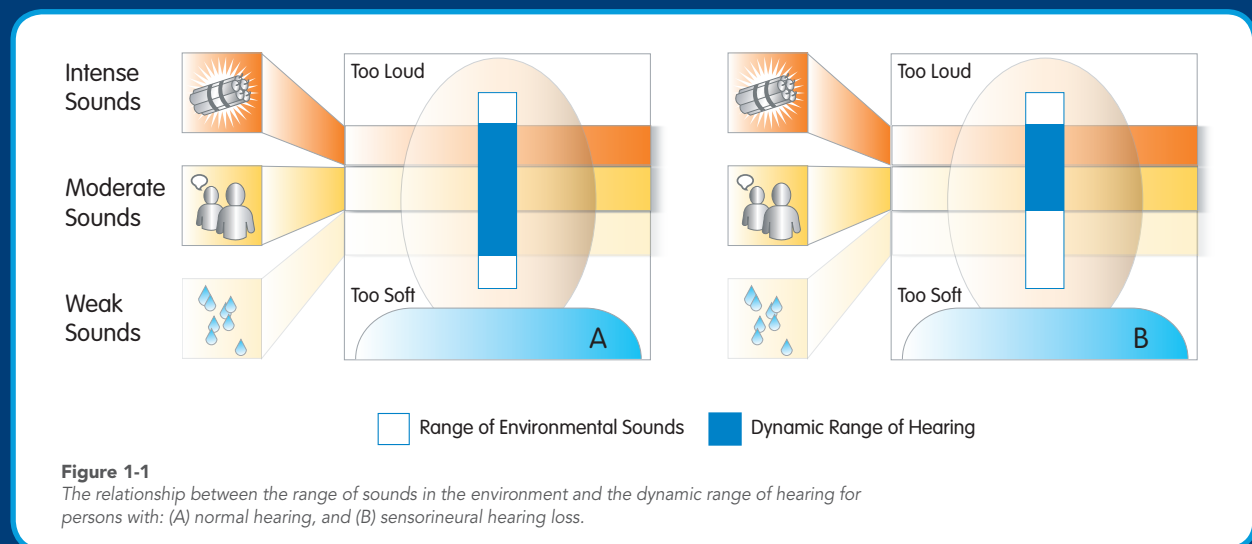
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CHAPTER 1: INTRODUCTION

According to the World Health Organization (WHO), hearing loss is one of the leading health concerns around the world (WHO, 2005). At the present time, hearing aids are the most common first step in [re] habilitation. On the surface, this seems fairly straightforward because much is known about the physiology of the auditory system and the psychoacoustics of perception. However, hearing loss impacts an individual in numerous ways, making the fitting of a hearing aid a complex process.

The intact auditory system is capable of perceiving a wide range of sounds, from the quiet pitter-patter of rain to the loud boom of explosives. In Figure 1-1A, the white bar represents the entire range of sounds, from extremely weak to extremely intense, that may occur in an individual's environment. The weakest sounds that are audible lie at the threshold of hearing sensitivity. At the opposite end is the loudness discomfort level (LDL), representing the most intense sounds that can be tolerated without pain. In between these two extremes is the dynamic range of hearing (shown by the blue bar).

For an individual with normal hearing, average conversational speech falls approximately midway within the dynamic range of hearing and coincides with the **most comfortable loudness level (MCL).**

The most common complaint associated with hearing loss is the inability to hear; specifically, the inability to hear soft sounds. Figure 1-1B depicts a person with sensorineural hearing loss. Once again the white bar is the range of sounds in the environment, while the blue bar represents the individual's dynamic range of hearing. You notice three things immediately. First, average conversational speech is now barely audible to the individual. Second, weak sounds (for example, gentle rain) are below the threshold of hearing and, therefore, too soft to be heard. And, finally, intense sounds (for example, the boom of an explosion) are still perceived as being loud. As a result of the threshold increasing and the LDL remaining the same, the dynamic range of hearing is considerably reduced compared to that of a person with normal hearing.

Intense Sounds — Too Loud
Moderate Sounds
Weak Sounds — Too Soft

A

B

☐ Range of Environmental Sounds   ■ Dynamic Range of Hearing

Figure 1-1
The relationship between the range of sounds in the environment and the dynamic range of hearing for persons with: (A) normal hearing, and (B) sensorineural hearing loss.

Human communication is arguably the single most important function of the auditory system. Indeed, reduced ability to hear speech is a major reason for seeking remediation. In addition to a reduced dynamic range, the communication difficulties of a person with hearing impairment are further complicated by the dynamic nature of speech itself. As shown in Figure 1-2, average conversational speech spans a range of 30dB (Beranek, 1947). Note that, in general, vowel sounds (for example /a/, /u/, and /i/) are low-pitched, relatively intense, and primarily responsible for making speech audible. On the other hand, consonants (especially unvoiced sounds such as /th/, /f/ and /s/) are high-pitched, relatively weak and carry most of the information that aids in speech understanding.

Figure 1-2
Frequency and level of various speech sounds at average conversational levels.
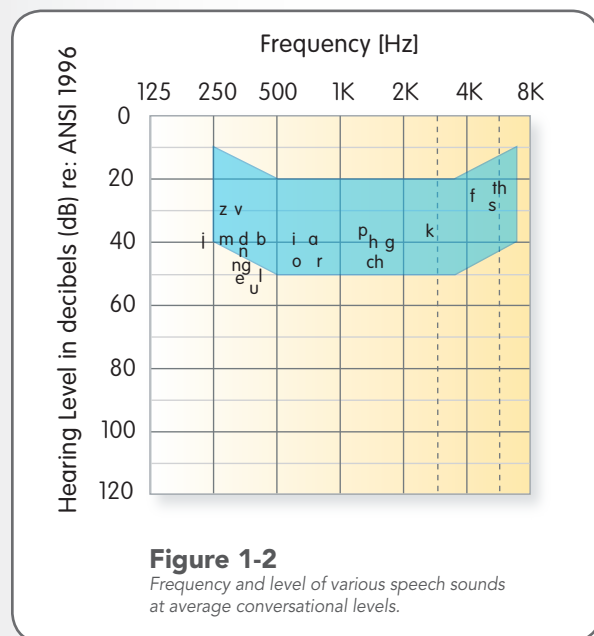
In light of these considerations, the hearing healthcare professional is faced with squeezing an elephant into a suitcase when fitting hearing aids – soft sounds must be made audible without loud sounds becoming uncomfortable, and speech should remain at a comfortable level. Compression amplification is a means for fitting the world of sound (the elephant) into the dynamic range of the individual with hearing impairment (suitcase).

This handbook is designed to provide the reader with a working knowledge of compression amplification: what it is, how it works, and how it is applied. Also included is some discussion on the principles of fitting compression systems, and troubleshooting problems.

Essential Terminology

Before launching into the detailed workings of a compression circuit, it is important to have some general knowledge of amplification. Despite the variety available, all hearing aids have some of the same basic components: a **microphone**, an **amplifier**, a **receiver**, and a **battery** (Figure 1-3). The microphone picks up the incoming acoustic signal and converts it to an electrical signal. The amplifier then magnifies the electrical signal. Like a loudspeaker, the receiver converts the amplified electrical signal back into an acoustic signal that is delivered to the ear. Finally, the battery provides the power for the circuit.

It is essential that the reader understand the following terminology that relates to the signal entering the hearing aid, the amplification, and the sound that is delivered to the ear.

Input

Input refers to the acoustic signal entering the hearing aid. Specifically, the **American National Standards Institute** (ANSI, 2003) defines input level as the **sound pressure level (SPL)** at the microphone opening of a hearing aid. Input level is expressed in dBSPL.

Figure 1-3
Basics components of a hearing aid: microphone, amplifier, receiver and battery.
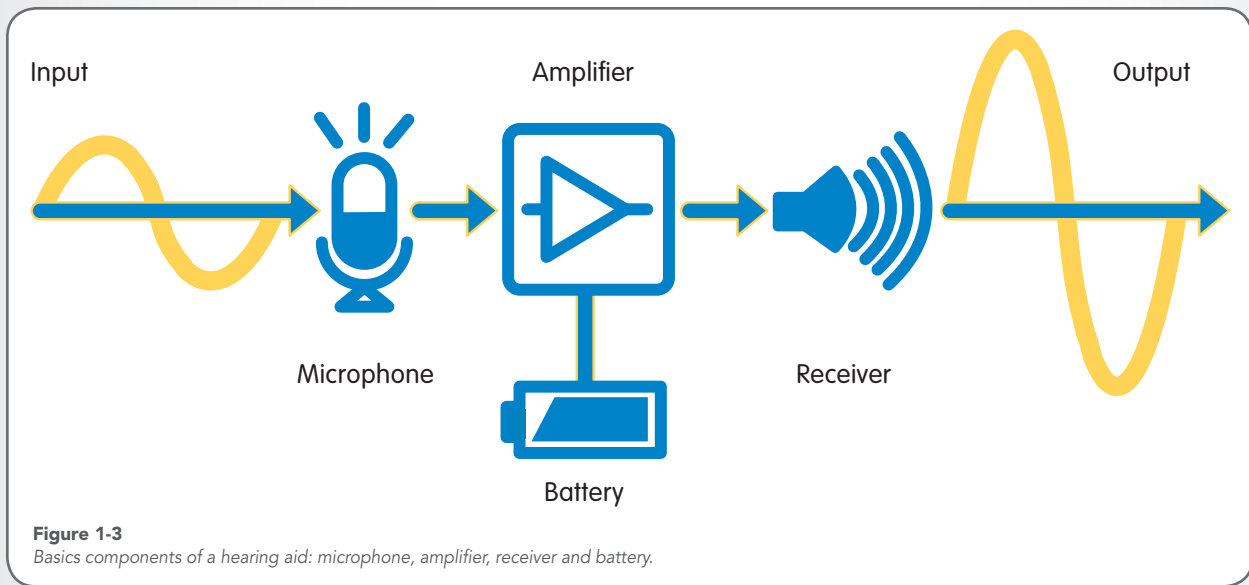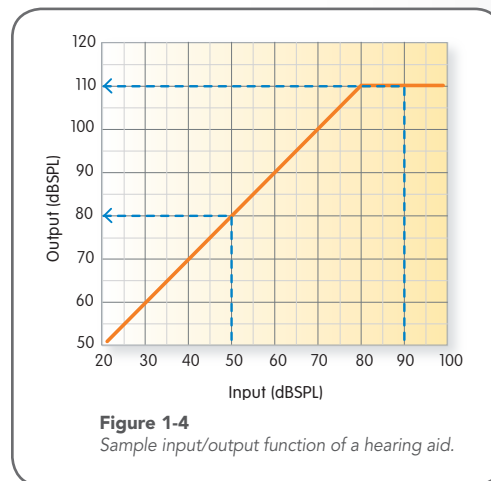
Output

Output refers to the amplified signal that is delivered to the ear. The output level is expressed in dBSPL.

Input/Output Function

An input/output (I/O) function is a graphical representation of the output of a hearing aid at various input levels. ANSI (2003) defines it as a single-frequency plot of the coupler SPL on the ordinate (Y-axis) as a function of the input SPL on the abscissa (X-axis) with equal decibel divisions on each axis; a similar definition is used by the **International Electrotechnical Commission** (IEC, 1983a). Figure 1-4 shows a sample I/O function of a hearing aid. In this example, an input of 50 dBSPL results in an output of 80 dBSPL, while an input of 90 dBSPL results in an output of 110 dBSPL. It can also be seen that the output of the hearing aid does not exceed 110 dBSPL. Figure 1-5 (page 6) shows sample I/O functions for three hearing aids. Notice that they do not all behave in the same way – inputs of 50 and 90 dBSPL result in different outputs for the three hearing aids.

Gain

Gain refers to the amount of amplification applied to the input signal. Specifically, ANSI (2003) defines gain as the difference between the output SPL in a coupler and the input SPL. Gain is expressed in dB. The mathematical relationship between input, gain and output is given by the simple formula:

Figure 1-4
Sample input/output function of a hearing aid.

Gain = Output – Input

Thus, in Figure 1-4, if an input of 50 dBSPL results in an output of 80 dBSPL, the gain of the hearing aid is:

Gain = Output – Input = 80 – 50 = 30dB

Figure 1-5
Sample input/output functions for three different hearing aids.

Figure 1-6
Sample input/gain function of a hearing aid.

Figure 1-7
Sample input/gain functions for three different hearing aids.
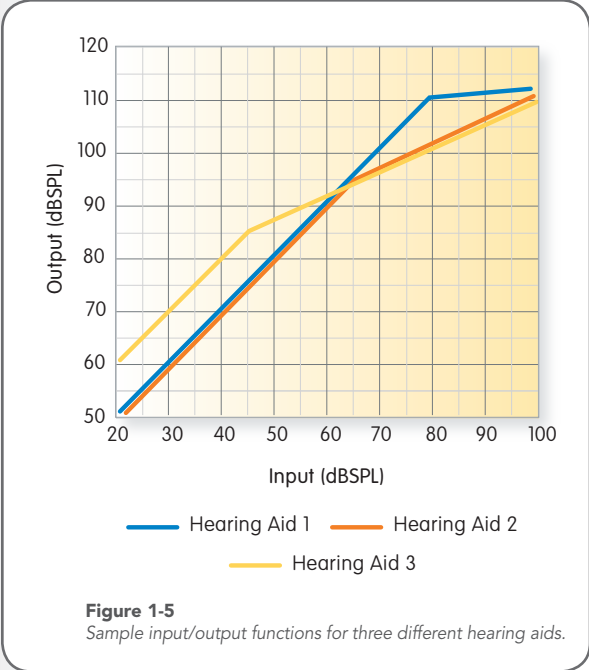
Similarly, with an output of 105 dBSPL for an input level of 90 dBSPL, Hearing Aid 3 (in Figure 1-5) has a gain of 15dB.

Input/Gain Function
An input/gain (I/G) function is a graphical representation of the gain of a hearing aid at various input levels. Figure 1-6 shows a sample I/G function of a hearing aid. In this example, the hearing aid provides 30dB of gain for an input of 50 dBSPL, but only 20dB of gain for an input of 90 dBSPL. Figure 1-7 shows sample I/G functions for three hearing aids. Notice that they do not all behave in the same way – different amounts of gain are applied to inputs of 50 dBSPL and 90 dBSPL.
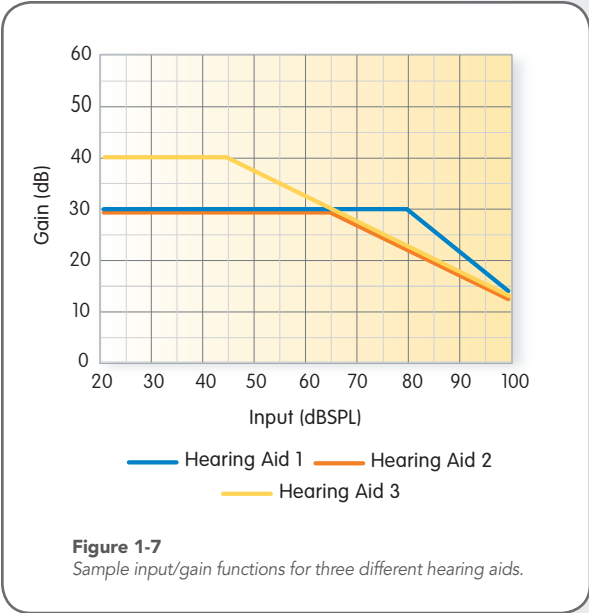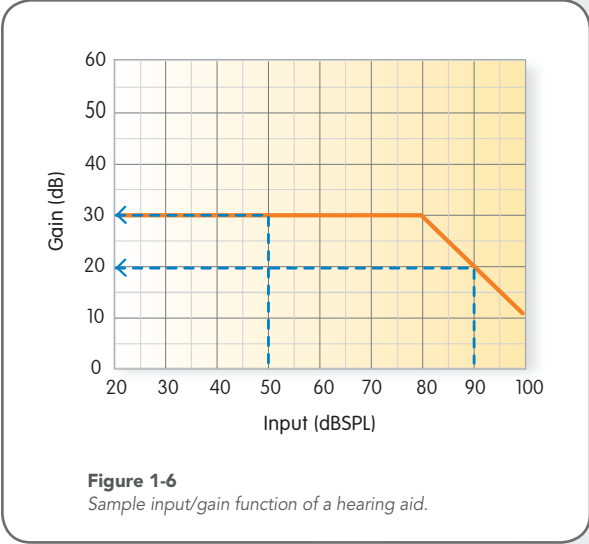
Just as gain can be calculated from the input and output, output is calculated by rearranging the formula for calculating gain as follows:

$$Output = Input + Gain$$

Thus, in Figure 1-6, if 30dB of gain is applied to an input of 50 dBSPL, the output of the hearing aid is:

$$Output = Input + Gain = 50 + 30 = 80 \text{ dBSPL}$$

Similarly, with gain of 18dB applied to an input level of 90 dBSPL, Hearing Aid 3 (in Figure 1-7) has an output of 108 dBSPL.

An obvious, but frequently overlooked, point is that even though the gain of a device decreases with increasing input level, the output continues to increase. This occurs because the decrease in gain is less than the increase in input level. Figures 1-8A and 1-8B, which show the I/O and I/G functions for a hearing aid, illustrate this point. 30dB of gain applied to an input of 50 dBSPL results in an output of 80 dBSPL. On the other hand, 20dB

of gain applied to an input of 90 dBSPL results in an output of 110 dBSPL. In this example, the input level increases by 40dB, while the gain decreases only by 10dB. Thus, although less gain is applied to the 90 dBSPL input than to the 50 dBSPL input, the output SPL is still greater for the 90 dBSPL input.

Figure 1-8A
Sample input/output function for a hearing aid.

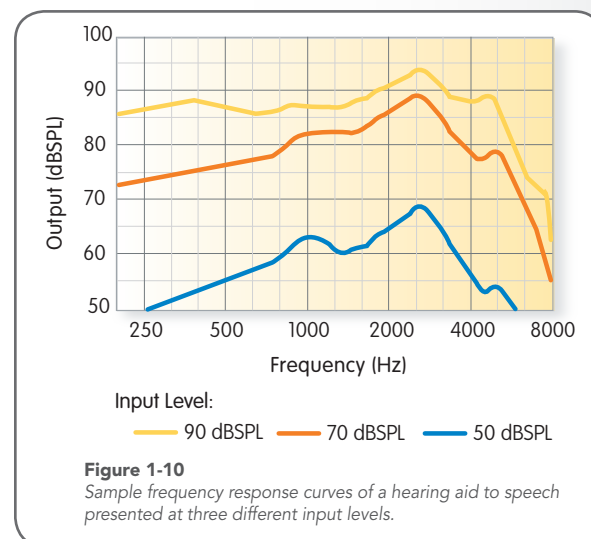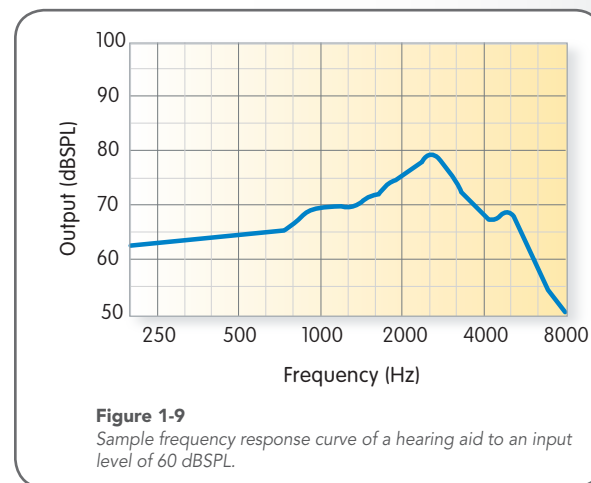**Figure 1-8B**
Sample input/gain (B) function for a hearing aid.
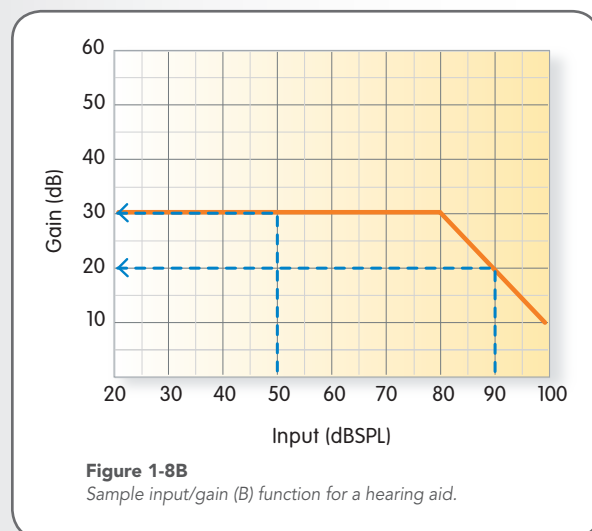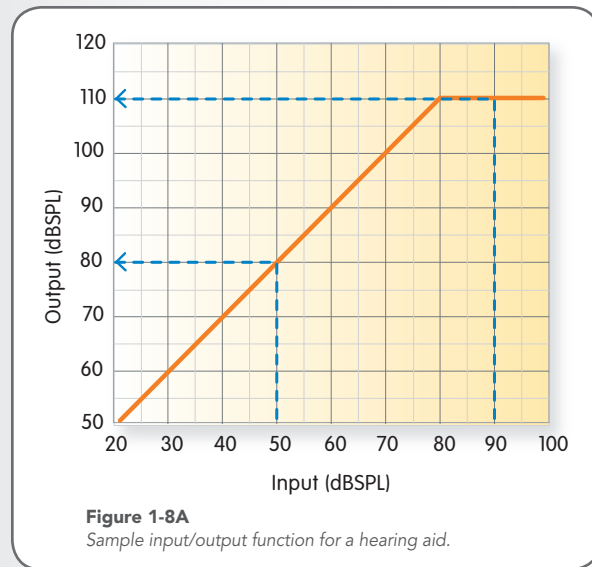
Frequency Response Curve

A frequency response curve is a graphical representation of the hearing aid output as a function of frequency. Specifically, the IEC (1983a) defines it as the SPL developed by a hearing aid in the ear simulator expressed as a function of frequency under specified test conditions. The input level and overall gain of

the hearing aid are fixed when measuring a frequency response curve. Figure 1-9 depicts a sample frequency response curve of a hearing aid for an input of 60 dBSPL. It can be seen that the output of the hearing aid varies across frequencies. Figure 1-10 shows sample frequency response curves for speech presented at three different input levels. It is not uncommon for the shape of the curve to change as the input level increases.

Figure 1-9
Sample frequency response curve of a hearing aid to an input level of 60 dBSPL.

Input Level:
— 90 dBSPL — 70 dBSPL — 50 dBSPL

Figure 1-10
Sample frequency response curves of a hearing aid to speech presented at three different input levels.

A **frequency-gain curve** is a graph showing the gain of a hearing aid as a function of frequency (Figure 1-11) under specified test conditions.

Figure 1-11
Sample frequency-gain curve of a hearing aid to an input level of 60 dBSPL.
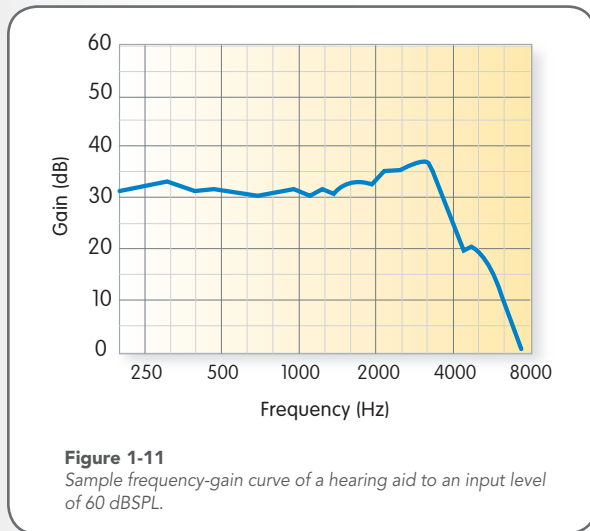
Peak Clipping

In general, the output of a hearing aid increases as the input level increases. However, once the output reaches a certain level, the hearing aid is incapable of producing a louder signal. **Maximum output** is the highest possible signal that a hearing aid is capable of delivering, regardless of the input level or the gain of the hearing aid. The maximum output of a hearing aid is determined by the characteristics of the microphone, amplifier and receiver. That is, the maximum output of the hearing aid is only as high as the weakest component in the chain. When the input level and gain exceed the maximum output, the hearing aid is said to be in **saturation**.

As long as the output of the hearing aid remains below the maximum output, the output signal is similar to the input signal, only larger in amplitude (Figure 1-12A). When the sum of the input level and gain exceed the maximum output of the hearing aid, however, the peaks of the output signal are clipped at the maximum output (Figure 1-12B). This is referred to as peak clipping. Note that the shape of the output signal is quite different from that of the input signal once the peaks are clipped.

Peak clipping is one method of controlling or limiting the maximum output of a hearing aid. [Alternate methods of output limiting will be discussed in subsequent chapters.] It is virtually impossible to determine whether or not peak clipping is occurring merely by examining an I/O or I/G function. The hallmark of peak clipping is that it produces an output signal that is distorted, often described as sounding "scratchy."

Distortion

Distortion refers to the presence of frequency components in the output of a hearing aid that were not present in the input signal. There are two types of distortion – harmonic and intermodulation.
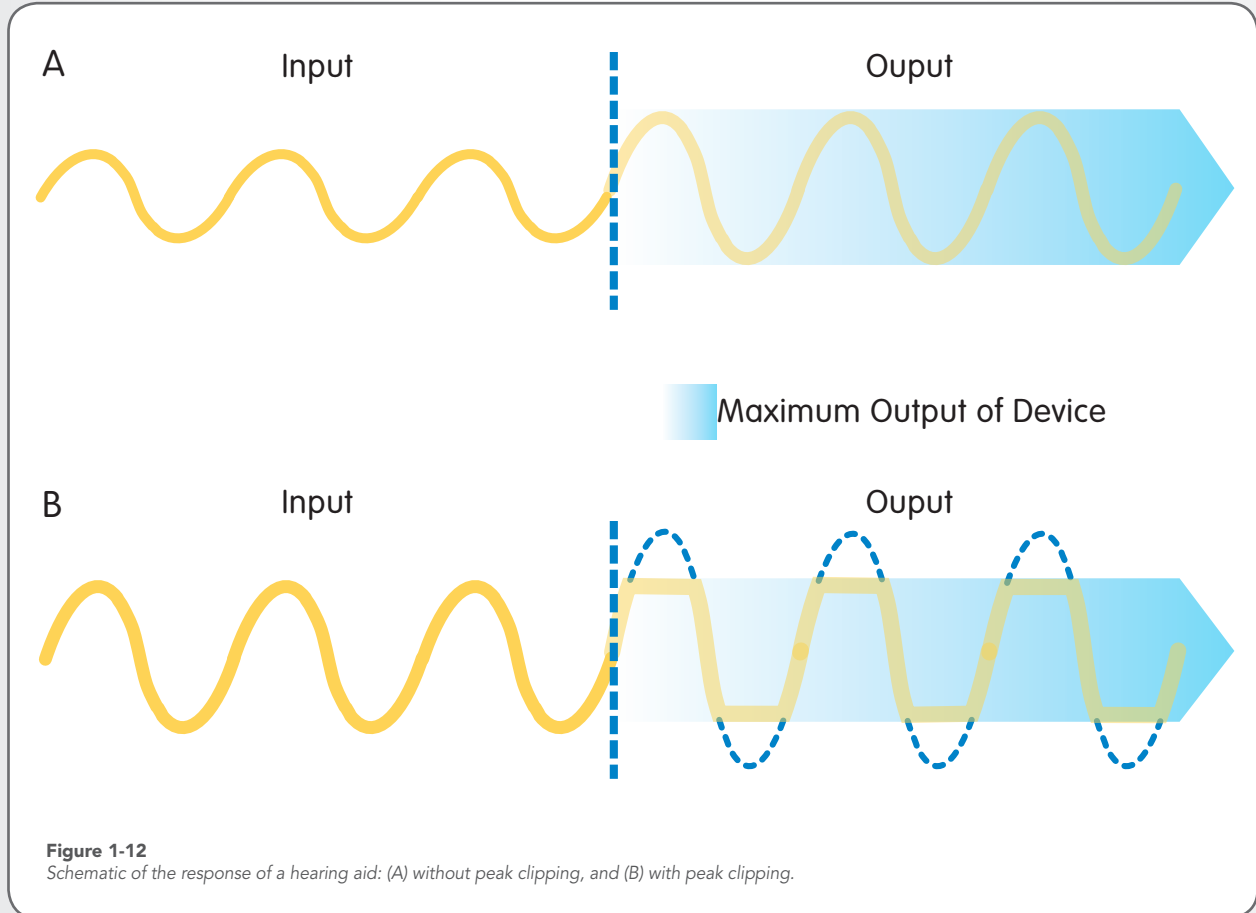
Harmonic distortion is said to occur when the output contains frequency components that are integer multiples of the input signal frequency (IEC, 1983a). For example, harmonics of a 500 Hz input signal would occur at 1000 Hz, 1500 Hz, 2000 Hz, 2500 Hz, and so on. Thus, harmonic distortion only occurs at frequencies greater than the input signal frequency. **Total harmonic distortion (THD)** is the summed power of all the harmonic distortion products relative to the power of the original input signal. THD is typically expressed as a percentage.

Intermodulation distortion is generated by the interaction of at least two signals of different frequencies in the input (IEC, 1983a). It occurs when the frequency components of a complex input signal combine to generate additional frequency components in the output signal. For example, if F1 and F2 represent two different frequencies, intermodulation distortion may occur at frequencies corresponding to F2-F1, 2F1-F2, 2F2-F1, 3F1-2F2, and so on. Thus, intermodulation distortion may occur at frequencies above and below the input frequencies.

Both types of distortion may occur simultaneously in a hearing aid. Distortion of any type results in unpleasant sound quality and may adversely affect speech intelligibility.

Now that the reader is familiar with some general hearing aid terminology, the next few chapters will introduce and describe the topic of compression.

Figure 1-12
Schematic of the response of a hearing aid: (A) without peak clipping, and (B) with peak clipping.

As discussed in Chapter 1, the intact auditory system has an exquisite ability to hear a wide range of sounds in the environment, from weak to intense (Figure 2-1A); average conversational speech falls comfortably in the middle of this dynamic range. A sensorineural hearing loss reduces the dynamic range of hearing available, also known as the **residual dynamic range**. That is, the individual is unable to hear weak sounds, average conversational speech is barely audible, and intense sounds are heard as loudly as the normal-hearing ear (Figure 2-1B).
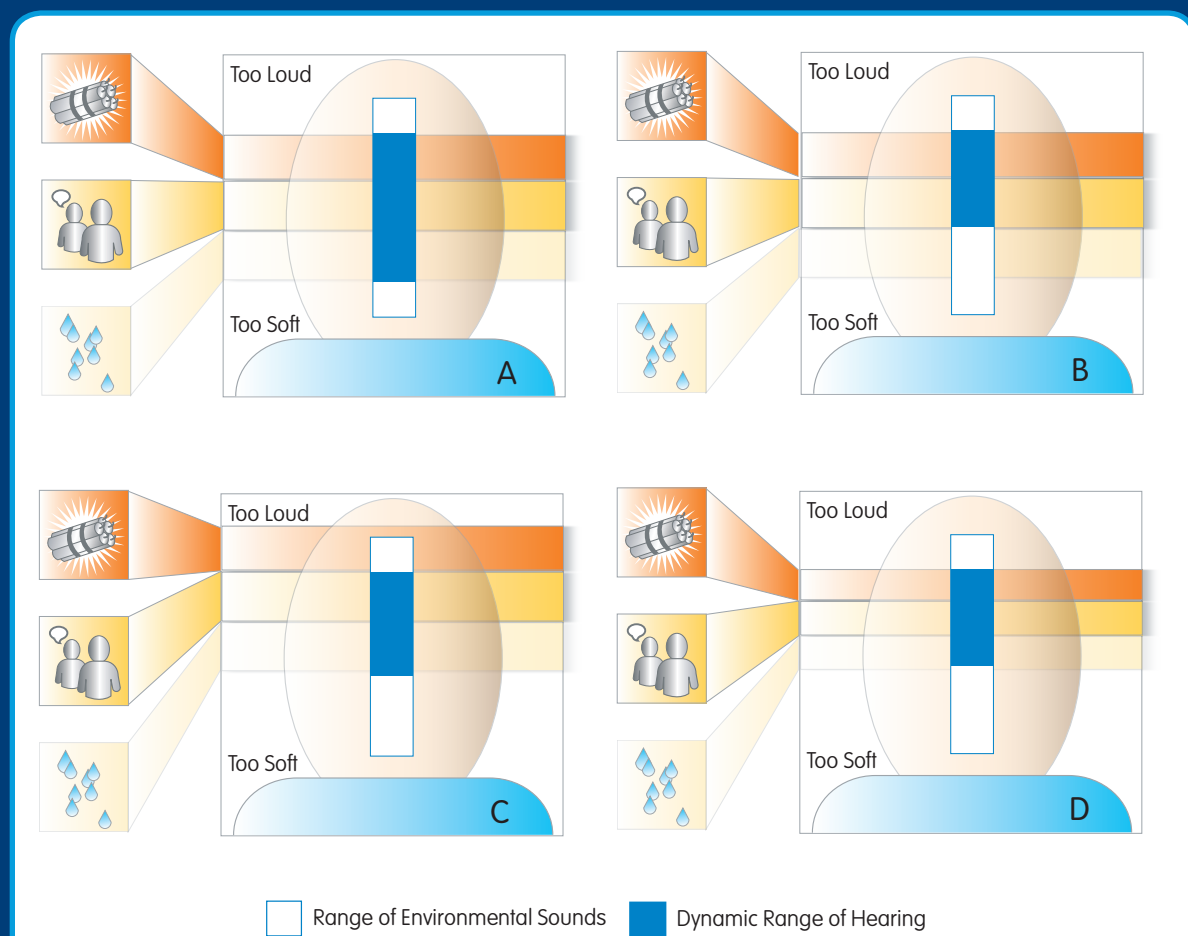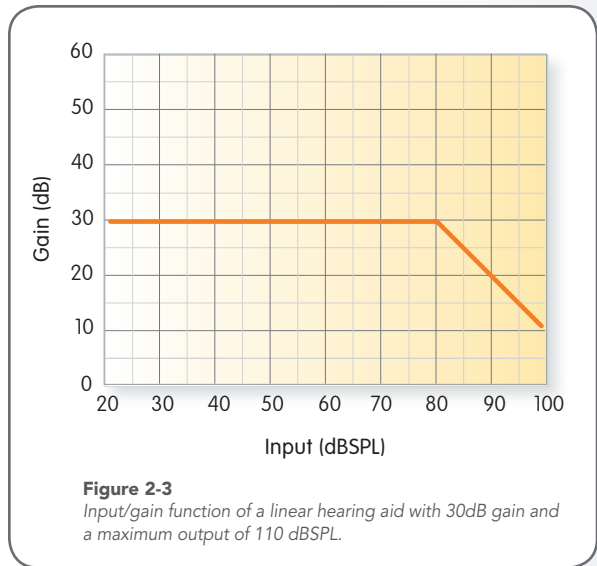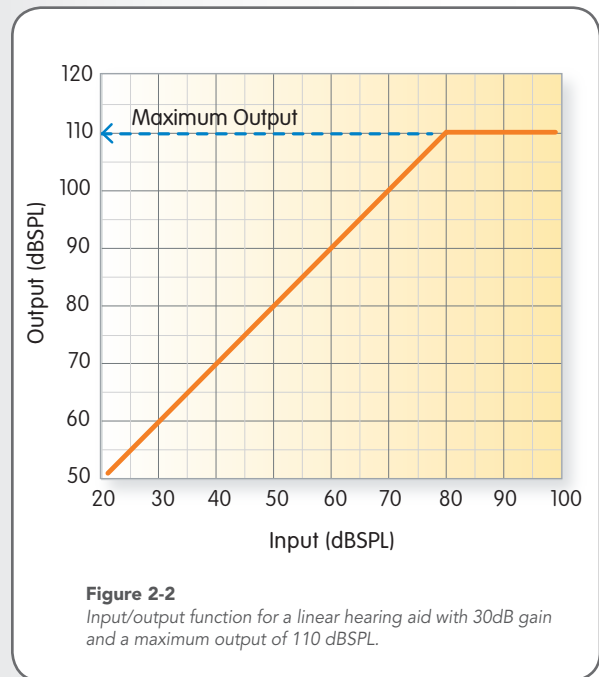

Figure 2-1
The relationship between the range of sounds in the environment and the dynamic range of hearing for persons with: (A) normal hearing, (B) sensorineural hearing loss, (C) sensorineural hearing loss with linear amplification, and (D) sensorineural hearing loss with compression amplification.

Because the most common complaint associated with hearing impairment is an inability to hear, it is tempting to alleviate the problem simply by making sounds louder. Figure 2-1C shows the sensorineural hearing loss to which a fixed amount of gain is applied to all sounds – i.e., amplification is **linear**. This makes weak sounds audible. However, average conversational speech is now loud, and intense sounds are amplified beyond the upper end of the dynamic range making them uncomfortably, or even painfully, loud. Ouch! In contrast, in Figure 2-1D, different amounts of gain are applied to weak, moderate and intense sounds – i.e., amplification is **non-linear**. This squeezes the range of environmental sounds to fit within the reduced dynamic range of the person with sensorineural hearing loss. Weak sounds are made audible, moderate sounds are comfortable, while intense sounds are loud without being uncomfortable. The result is that the hearing aid user perceives the world of sounds in much the same way as a person with normal hearing.

The exact manner in which non-linear amplification, or compression, is applied depends on the goal of the hearing aid fitting as well as on the characteristics of the circuit.

Figure 2-3
Input/gain function of a linear hearing aid with 30dB gain and a maximum output of 110 dBSPL.

Figure 2-2
Input/output function for a linear hearing aid with 30dB gain and a maximum output of 110 dBSPL.

Before we begin discussing compression, it may be helpful to lay the groundwork with a brief description of linear amplification, the simplest and most basic amplification system. Figure 2-2 shows the I/O function for a linear hearing aid. As expected, the output increases as the input to the hearing aid increases. More importantly, for every 10dB increase in the input level, the output also increases by 10dB. For example, when the input level increases from 50 dBSPL to 60 dBSPL, the output level increases from 80 dBSPL to 90 dBSPL. This 1:1 relationship continues until the maximum output of the hearing aid is reached. Notice also, that the gain of the hearing aid is constant at 30dB until the maximum output is reached. This is more easily seen in Figure 2-3, which shows the I/G function for the same linear hearing aid. The reduction in gain for input levels greater than 80 dBSPL occurs because the maximum output of the hearing aid cannot exceed 110 dBSPL. Thus, the hallmark of a linear hearing aid is that it applies a fixed amount of gain regardless of the level of the input signal, until the maximum output of the hearing aid is reached. In contrast, a non-linear hearing aid applies different amounts of gain to weak, moderate and intense sounds.

Characteristics of a Compressor

The function and application of compression circuits are defined by their static and dynamic features. Static features, such as compression threshold and compression ratio, indicate the behavior of the circuit in response to steady input signals (e.g., a running vacuum cleaner or the constant hub-bub of a noisy restaurant). On the other hand, dynamic features, such as attack time and release time, describe the length of time required for the circuit to respond to a changing input signal (e.g., speech in a one-on-one conversation).

Compression Threshold

Compression threshold (CT) is defined as the input SPL which, when applied to the hearing aid, gives a reduction in the gain of 2 (±0.5)dB with respect to the gain in the linear mode (IEC, 1983b). In other words, it is the point on the I/O function at which the output level is 2dB lower than it would be if no compression had occurred (i.e., if the processing were linear). Thus, in Figure 2-4, the CT is 54 dBSPL. Because it looks like the knee of a bent leg, the point at which the slope of the I/O function changes is referred to as the threshold kneepoint (TK). The hearing aid shown in Figure 2-4 has a TK of 50 dBSPL. Because the CT and TK are within a few dB of each other, the two terms are often used interchangeably, and will be used as such here.

Depending on the purpose, a compression system may have high or low TKs. A high TK, 60 dBSPL or greater, is used to limit the output of a hearing aid so that it does not exceed the individual's loudness discomfort levels and to maximize listening comfort (for example, Hearing Aids 1 and 2, respectively, in Figure 2-5). On the other hand, a low TK, typically set below 60 dBSPL, may be used to improve audibility of the softer components of speech and/or to restore loudness perception (for example, Hearing Aid 3 in Figure 2-5). Finally, like Hearing Aid 4 in Figure 2-5, a device may have a high and a low TK to achieve all of these goals.
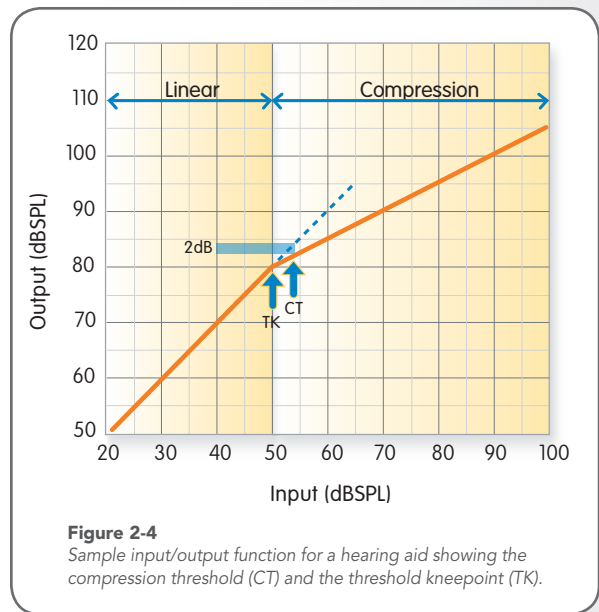
Figure 2-4
Sample input/output function for a hearing aid showing the compression threshold (CT) and the threshold kneepoint (TK).
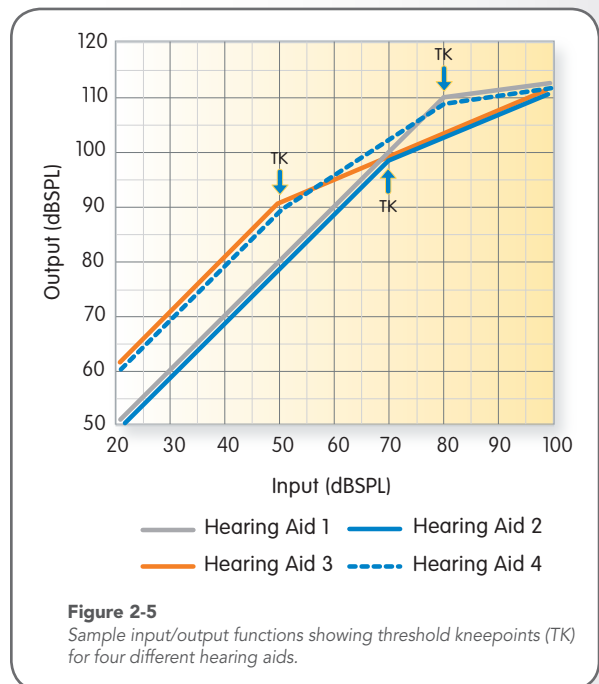
Figure 2-5
Sample input/output functions showing threshold kneepoints (TK) for four different hearing aids.

Compression Ratio

Once the input signal is loud enough to activate compression (i.e., when the input level exceeds the TK), the compression ratio (CR) determines how much the signal will be compressed. Specifically, under steady-state conditions, it is the ratio of an input SPL difference to the corresponding output SPL difference (IEC, 1983b). Thus, CR relates a change in the input level (ΔInput) to a change in the output level (ΔOutput). [The symbol Δ is pronounced "delta"

and stands for a change in the target quantity.] CR is calculated using the formula:

$$CR = \frac{\Delta Input}{\Delta Output}$$

For the hearing aid shown in Figure 2-6, increasing the input from 30 to 50 dBSPL (ΔInput = 20dB) increased the output from 60 to 80 dBSPL (ΔOutput = 20dB). Using the above formula, the CR of the hearing aid is:

$$CR = \frac{\Delta Input}{\Delta Output} = \frac{20}{20} = 1{:}1$$

Similarly, when the input increases from 70 to 90 dBSPL and the output increases from 90 to 100 dBSPL (Figure 2-6), the CR is 2:1.

CRs are generally expressed in terms of the number of dB by which the input must change in order to effect a 1-dB change in the output. For example, a CR of 2:1 indicates that a 2-dB change in input results in a 1-dB change in output. Because the reference condition is always a 1-dB change in output, the latter part of the ratio may be dropped. Thus, a CR of 2:1 can be simply expressed as 2.

Note that, in Figure 2-6, 30dB of gain is applied at and below the TK regardless of the input level, resulting in linear amplification. Thus, another way to look at linear amplification is that it has a CR of 1:1; compression is applied only above the TK and when the CR is greater than 1:1.

Depending on the purpose, a compression system may have high or low CRs. A high CR, 5.0 or greater, is used to limit the output of a hearing aid so that it does not exceed the individual's loudness discomfort levels (for example, Hearing Aids 1 and 2 in Figure 2-7). On the other hand, a low CR, typically set between 1.0 and 5.0, may be used to improve audibility of the softer components of speech and/or to restore loudness perception (for

example, Hearing Aid 3 in Figure 2-7). Finally, like Hearing Aid 4 in Figure 2-7, a device may have a high and a low CR to achieve all of these goals. Although necessary for some applications, high CRs are known to adversely affect the clarity and pleasantness of the amplified sound (Neuman et al, 1998).

Figure 2-6
Calculating compression ratios (CRs) from an input/output function. ΔI = Change in input, ΔO = Change in output.

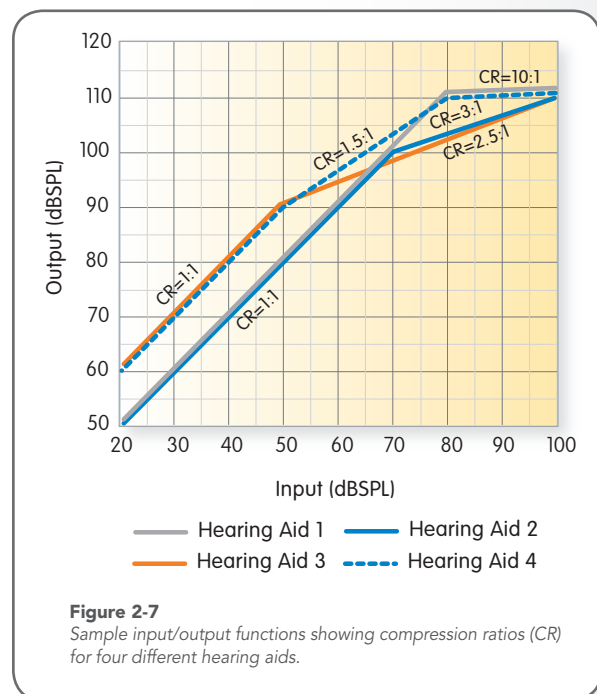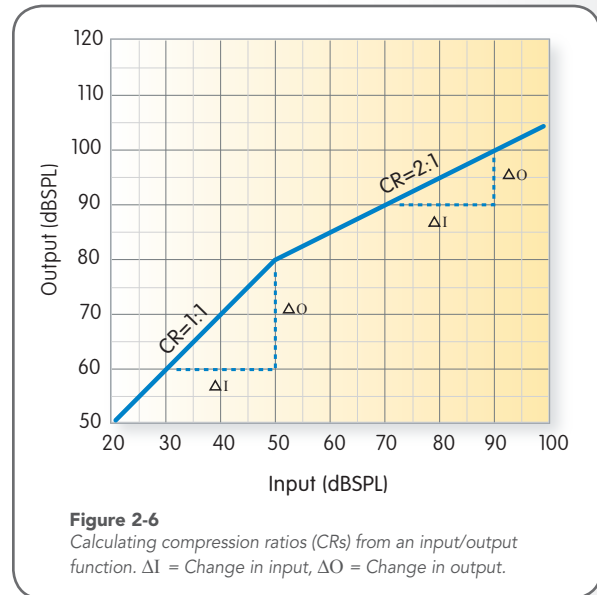**Figure 2-7**
Sample input/output functions showing compression ratios (CR) for four different hearing aids.

Attack Time and Release Time

When the incoming signal changes abruptly in level from below the TK to above it, the compressor is unable to change the gain instantaneously. The dynamic characteristics of a compressor refer to the length of time required for the compression circuit to respond to a sudden change in the input. Figure 2-8 is a schematic representation of how changes in the input level produce variations in gain and output over time.

Attack time (AT) is the time delay that occurs between the onset of an input signal loud enough to activate compression (i.e., input signal exceeds the TK) and the resulting reduction of gain to its target value. Specifically, ANSI (2003) defines AT as the time between the abrupt increase in input level from 55 to 90 dBSPL and the point where the output level has stabilized to within 3dB of the steady value for an input of 90 dBSPL. [IEC (1983b) defines AT as the time interval between the moment when

Figure 2-8
Schematic representation of the behavior of a hearing aid associated with changes in input. The input level, gain and output of the hearing aid are shown in panels A, B and C, respectively. The attack time (AT) and release time (RT) are indicated by the shaded areas. TK = Threshold kneepoint.
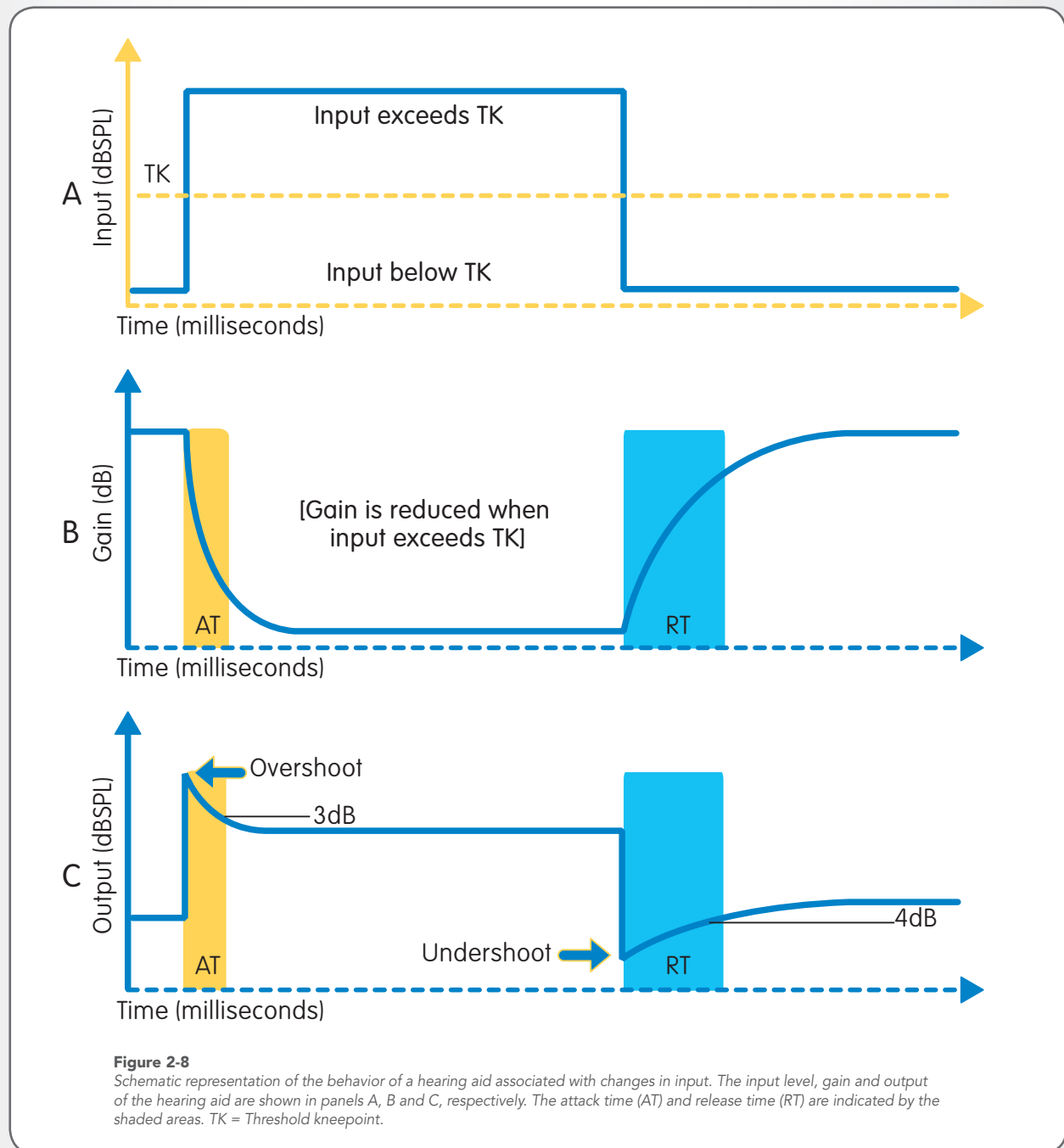
the input signal level is increased abruptly by a stated number of decibels and the moment when the output SPL from the hearing aid stabilizes at the elevated steady-state level within ±2dB. The AT for the normal dynamic range of speech in computed between input levels of 55 and 80 dBSPL, whereas the high-level AT is computed between 60 and 100 dBSPL.] In Figure 2-8, when the input increases to a level above the TK (Figure 2-8A), the gain of the hearing aid does not change immediately (Figure 2-8B). The result is an overshoot in the output (Figure 2-8C). As the gain approaches its target, so too does the output of the hearing aid – i.e., reaches within 3dB of its final value.

Release (or recovery) time (RT) is the time delay that occurs between the offset of an input signal sufficiently loud to activate compression (i.e., input signal falls below the TK) and the resulting increase of gain to its target value. Specifically, ANSI (2003) defines RT as the interval between the abrupt drop in input level from 90 to 55 dBSPL and the point where the output level has stabilized to within 4dB of the steady value for an input of 55 dBSPL. [IEC (1983b) defines RT as the time interval between the moment when the input signal level is decreased abruptly by a stated number of decibels and the moment when the output SPL from the hearing aid stabilizes at the lower steady-state level within ±2dB. The RT for the normal dynamic range of speech in computed between input levels of 80 and 55 dBSPL; the high-level RT is computed between 100 and 60 dBSPL.] In Figure 2-8, the gain of the hearing aid does not change immediately (Figure 2-8B) when the input decreases to a level below the threshold kneepoint (Figure 2-8A). The result is an undershoot in the output (Figure 2-8C), until both the gain and output reach their target – i.e., to within 4dB of their final values.

Depending on the purpose, a compression system may have fast or slow attack and release times. The faster the AT, the shorter the duration of the overshoot and the shorter the period of

time that the individual has to listen to sounds louder than desired. Indeed, ATs as fast as 5 ms are especially desirable when compression is used to limit the maximum output of a hearing aid. Although RTs may also be a few milliseconds in duration, the consequences when combined with a fast AT may be undesirable because the gain will vary in response to each cycle of the incoming signal resulting in a distorted waveform. Thus, the RT is generally longer than the AT. An RT of 20 ms is considered fast. The disadvantage of ATs and RTs between 100 ms and 2 s is that it causes the compressor to respond to brief sounds, or lack thereof, in the environment. For example, although gain may be reduced in the presence of speech, it will increase during pauses in the utterance. This results in a **pumping** sensation where the level of the background (or ambient) noise increases audibly during pauses and decreases when speech is present. This pumping sensation is less problematic for attack and release times faster than 100 ms because the gain changes occur too quickly to be perceived. A fast AT coupled with a slow RT (greater than or equal to 2 s) will adversely affect the audibility of speech that follows immediately after the gain reduction in response to a finger snap or the click of a pen. Finally, attack and release times of 2 s or slower respond to changes in the overall level of sound in the environment rather than to individual events.

Some compression circuits incorporate **adaptive** or **variable release times** – that is, the RT is adjusted based upon the duration of the triggering signal. Figure 2-9 (page 16) is a schematic representation of the changes in gain and output associated with changes in the input level of different durations. Thus, for example, if the input is a brief, transient sound such as a door slam, the RT is fast to have the least possible effect on the audibility of the speech that follows. On the other hand, an input that is sustained is indicative of a change in the overall level of sound in the environment. In this instance, a slower RT acts in much the same way as a manual adjustment of the volume control.
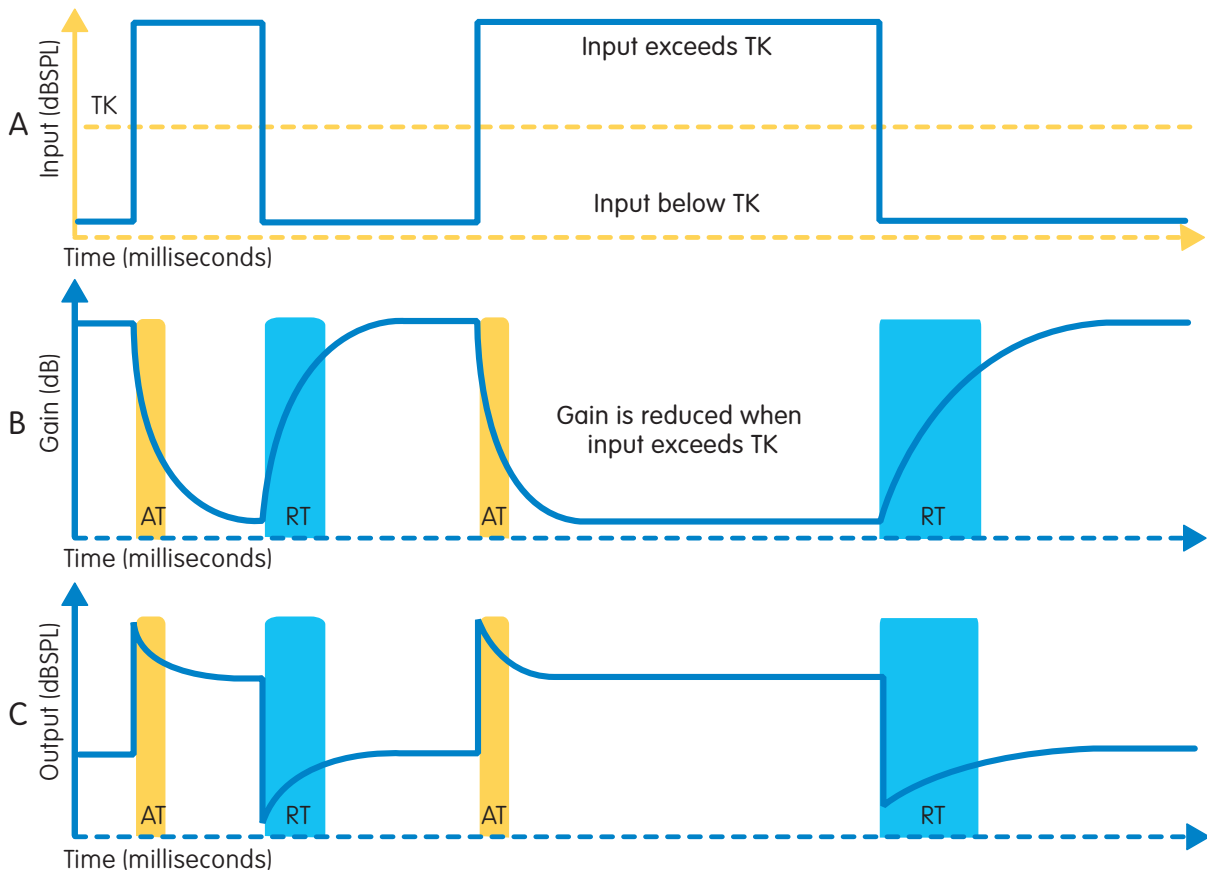
Figure 2-9
Schematic representation of variable release time. The input level, gain and output of the hearing aid are shown in panels A, B, and C, respectively. The attack time (AT) and release time (RT) are indicated by the shaded areas. TK = Threshold kneepoint.

Although it is convenient to discuss the static and dynamic characteristics of compression as discrete entities when learning about them, it is important to understand that they interact with each other in systematic ways. For example, CRs are determined from the response of the hearing aid to relatively steady signals, such as pure tones or speech-shaped noise. For time-varying inputs, such as speech, the effective CR is significantly affected by the AT and RT. When the AT and RT are fast – i.e., shorter than the duration of a phoneme or syllable – the gain changes sufficiently quickly to amplify softer components more than the louder components. The result is an effective CR for speech that is similar to that specified on the basis of steady signals. On the other hand, when the AT and RT are slow – i.e., longer than the duration of a typical word or utterance – the gain does not change much between softer and louder phonemes. As a result, the effective CR for speech is much lower than would be expected for steady signals. Figure 2-10 shows the effects of fast versus slow AT and RT on average conversational speech. Specifically, note that the range between the upper and lower limits of speech is smaller with the fast AT and RT, indicating more compression of the signal, than with the slow AT and RT.

Another example of the interaction between compression parameters is the observation that fast ATs and RTs are more detrimental to the perceived sound quality at high CRs than at low CRs (Woods et al, 1999).

Input-Versus Output-Controlled Compression

The term **automatic gain control (AGC)** is often used to describe compression circuits because the amount of gain applied is automatically determined by the signal level. Thus, a level detector is an essential component of any compression circuit. The position of this level detector relative to the volume control influences the operation of the circuit.

With **input-controlled compression (AGC-I)**, the level detector is located before the volume control (Figure 2-11) and compression acts on the input to the hearing aid. T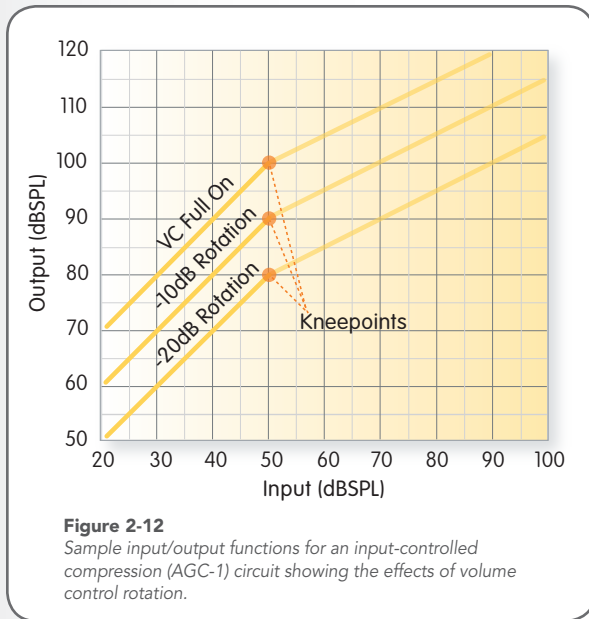hat is, once the input exceeds the TK, the compressor is activated and gain is reduced at the pre-amplifier. Therefore, the volume control setting has no impact on the compression parameters. As shown in Figure 2-12 (page 18), rotating the volume control from full-on to -20dB (relative to full-on) decreases the gain of the hearing aid above and below the threshold kneepoint. However, neither the TK nor the CR changes as a result.

Figure 2-10
Effects of fast and slow attack and release times on the response of a hearing aid to a speech signal.

Figure 2-11
Schematic representation of an input-controlled compression (AGC-1) circuit.

Figure 2-12
Sample input/output functions for an input-controlled compression (AGC-1) circuit showing the effects of volume control rotation.
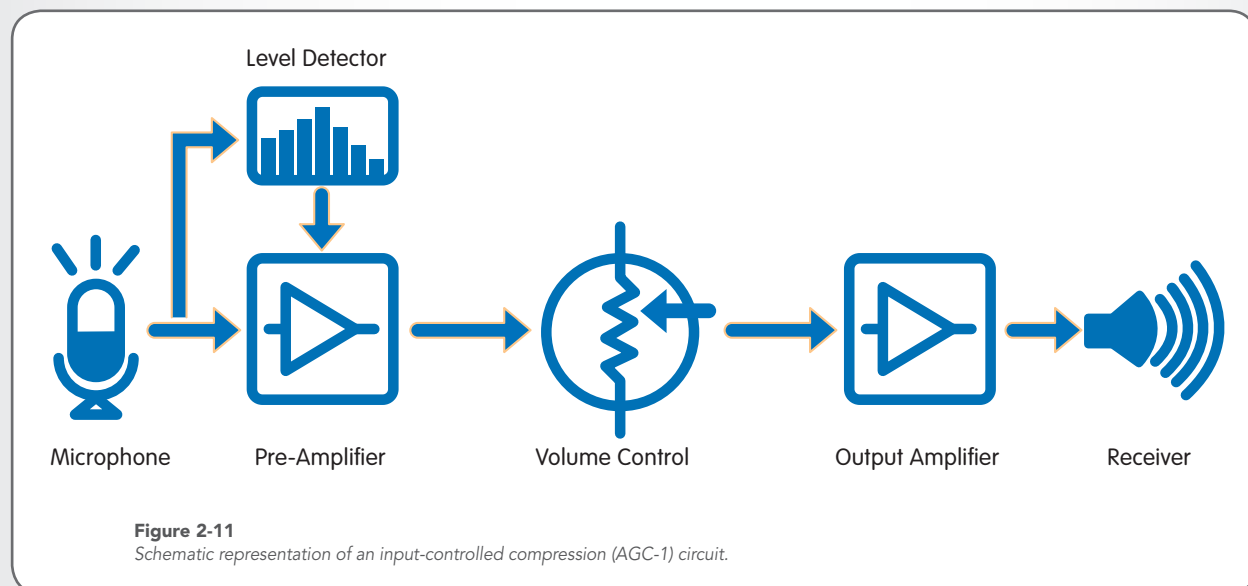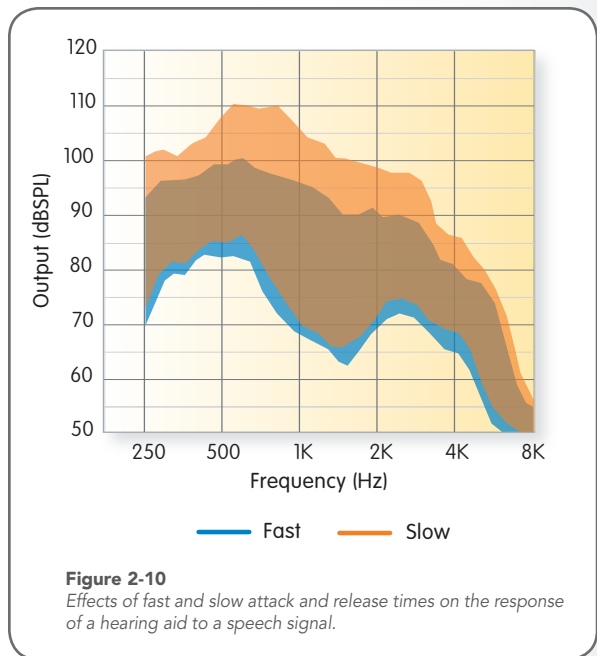
In **output-controlled compression (AGC-O)**, the level detector is located after the volume control (Figure 2-13) and compression acts on the output of the hearing aid. That is, the compressor is activated once the output exceeds the TK. As shown in Figure 2-14, rotating the volume control from full-on to -20dB (relative to full-on) decreases the gain of the hearing aid. More importantly, however, it results in an increase in the TK.

It is impossible to determine whether a compression circuit uses AGC-I or AGC-O simply by examining a single I/O function or frequency response curve. The only way to distinguish between the two types of circuits is by measuring the response of the hearing aid at various settings of the volume control. Although gain changes are seen with both types of circuits, the TK changes only in AGC-O circuits.

Channels and Bands

Although not strictly a "characteristic" of a compressor, a brief discussion of channels and bands is in order. The terms are often used erroneously or interchangeably, but a distinction is made between the two in this book.

Frequency bands are independently controlled areas for gain adjustment only. Thus, increasing or decreasing the gain in a frequency band will equally affect the response to weak, moderate and intense sounds at frequencies within that band. The compression parameters are unaffected. In contrast, **compression channels** allow separate adjustments for weak and intense input levels;

Figure 2-13
Schematic representation of an output-controlled compression (AGC-O) circuit.
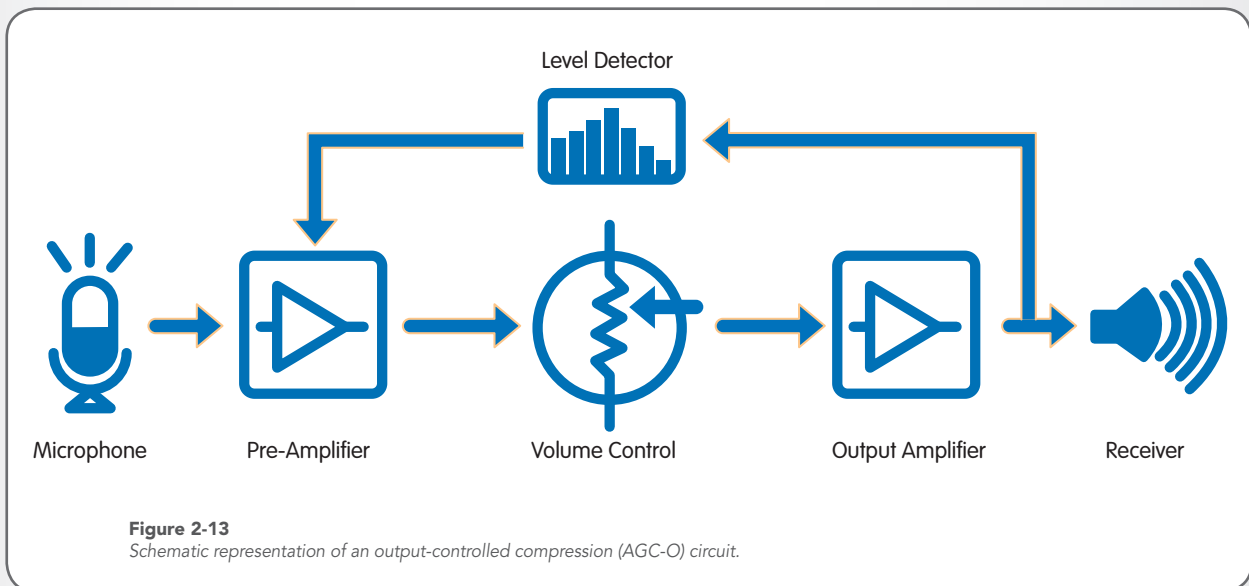
the effect on moderate level inputs depends on the compression architecture. Thus, in addition to gain, changes within a compression channel may affect the CR and/or the TK. Depending on the controls provided by the hearing aid manufacturer, it may be possible to make changes across both, frequency bands and compression channels.
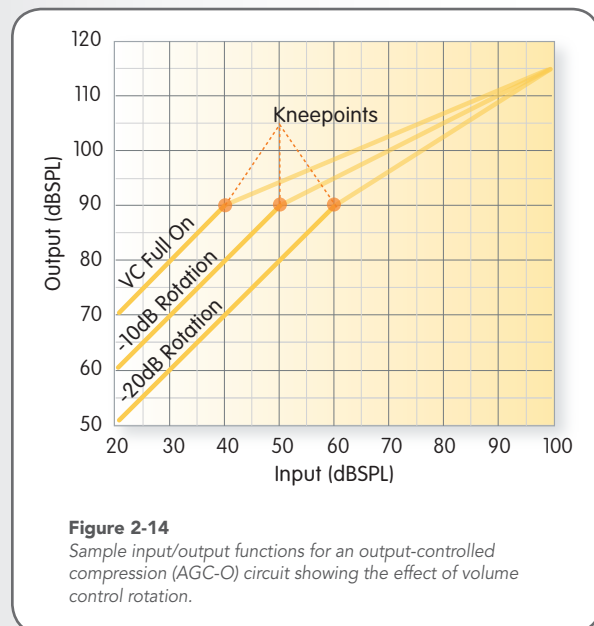

Figure 2-14
Sample input/output functions for an output-controlled compression (AGC-O) circuit showing the effect of volume control rotation.
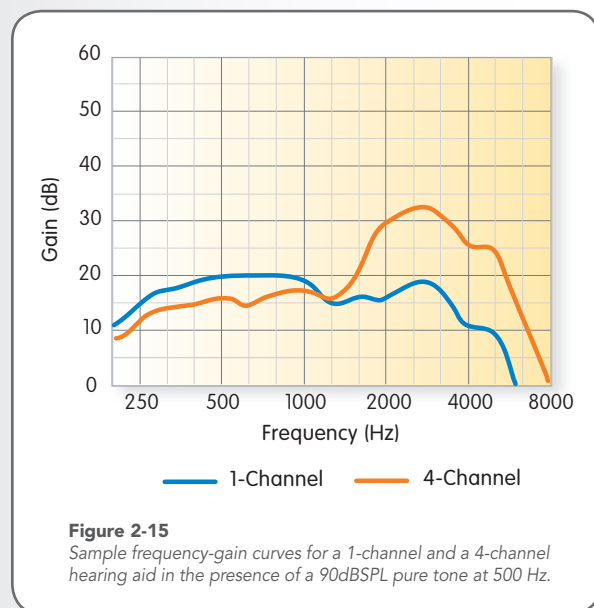

Figure 2-15
Sample frequency-gain curves for a 1-channel and a 4-channel hearing aid in the presence of a 90dBSPL pure tone at 500 Hz.

A hearing aid may have the flexibility to allow adjustments in multiple frequency bands and/or compression channels. Multiple channels are separated by **crossover frequencies**, which may or may not be adjustable. The ability to change gain individually in bands and/or channels is useful for two reasons. First, there is a wide variety of audiometric configurations for which a given hearing aid may be useful. The ability to adjust gain and compression in discrete frequency regions permits customization of amplification to the individual's needs. Another advantage of **multichannel compression** is that acoustic events in a discrete frequency region do not affect the response of the hearing aid at all frequencies. For example, Figure 2-15, shows the response of a 1-channel and 4-channel hearing aid in the presence of a 90 dBSPL tone at 500 Hz. It is clear that the high-frequency response is considerably reduced for the single-channel hearing aid, but unaffected for the multichannel hearing aid. This multichannel advantage may also extend to advanced signal processing features such as noise management.

Although on the surface it appears that a large number of channels would be advantageous, it is unclear as to whether or not this is the case. Trine and Van Tasell (2002) have shown that 3-4 channels are adequate to fit the majority of audiometric configurations, with or without the ability to further fine-tune the frequency response within each channel (i.e. in frequency bands), provided that the crossover frequencies are adjustable. One argument against the use of multichannel compression is that it obliterates the relative intensity relationships between various speech sounds, which are an important cue for speech understanding (Yund and Buckles, 1995). Finally, the sounds and noises that occur in our environments are usually not discrete in frequency. Thus, gain would be affected in fairly broad frequency regions even for a hearing aid with an infinite number of channels. This problem is further exacerbated when there is considerable overlap across channels.

Visualizing Compression

In summary, consider the following analogy to visualize the basics of compression. Picture a gondolier guiding his gondola under a bridge. The level of the water represents the input level and the height of the gondolier the gain of the hearing aid. Thus, the output is the sum total of the water level and the height of the gondolier. The bridge represents the maximum output.

In Figure 2-16A, the water level is low (weak sounds) and the gondolier is able to guide the gondola under the bridge with ease (i.e., no gain reduction is necessary).

If the water level rises (moderate or intense sounds), as in Figure 2-16B, the gondolier will lose his hat when going under the bridge

(peak clipping). Thus, when he emerges on the other side of the bridge, he will look different (distorted) without his hat.

At high water levels, the gondolier can avoid losing his hat by bending his head forward (high TK), as shown in Figure 2-16C. Note that he must bend his head as far forward as possible (high CR). Also, if the water level rises further, he could still lose his hat.

Another way that the gondolier can avoid losing his hat at high water levels is by bending forward gently at the waist (low TK) (Figure 2-16D). In addition to being more comfortable (low CR), this approach also affords him greater flexibility in the event that the water level rises further.
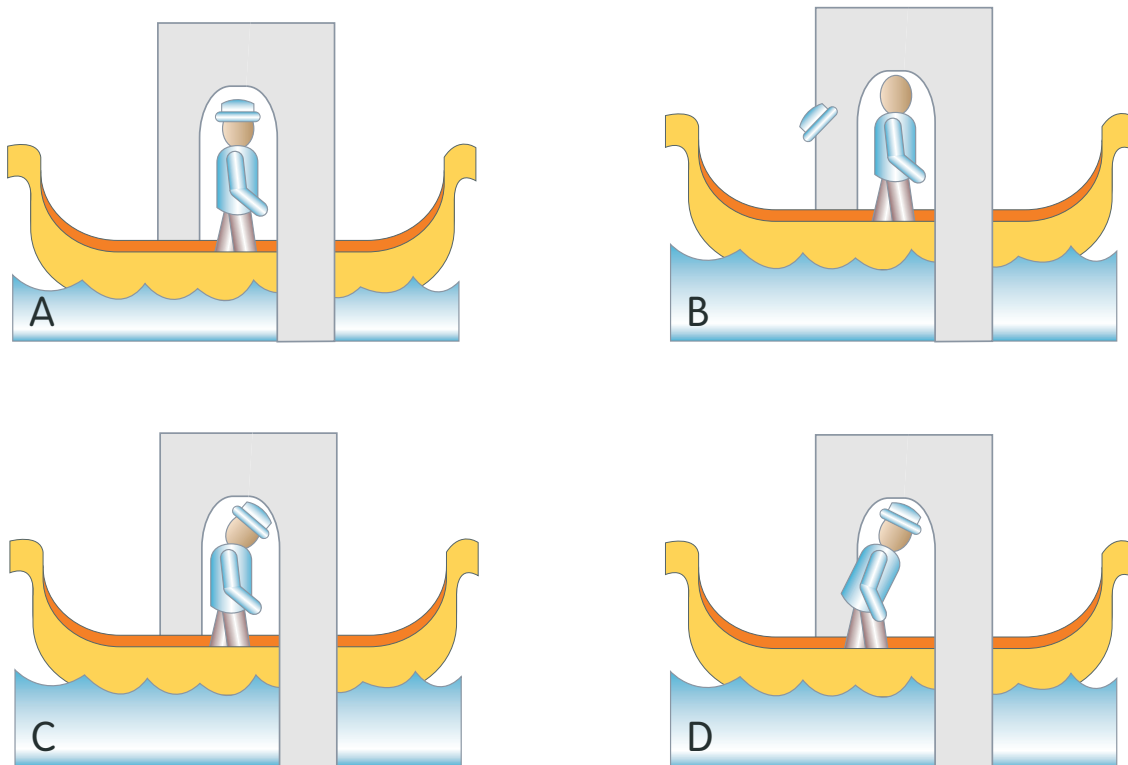

Figure 2-16
Schematic for visualizing compression: (A) at low water levels, the gondolier passes under the bridge with his hat, (B) at high water levels, the gondolier would lose his hat while passing under the bridge. At high water levels, the gondolier could avoid losing his hat by (C) bending his head as far forward as possible, or (D) bending more gently from the waist.

CHAPTER 3: APPLICATIONS OF COMPRESSION

We have already determined that compression provides non-linear amplification. That is, the gain decreases as the input level increases. But, by how much should the input signal be compressed? At what input level? And, how quickly? The answers to these questions depend upon the overall goal of the hearing aid fitting. Compression may be used to:

1. Limit the output of the hearing aid without distortion,

2. Minimize loudness discomfort,

3. Prevent further damage to the auditory system,

4. Optimize the use of the residual dynamic range,

5. Restore normal loudness perception,

6. Maintain listening comfort,

7. Maximize speech recognition ability, and

8. Reduce the adverse effects of noise.

Compression circuits are defined by their features – threshold kneepoint, compression ratio, and attack and release times. At the present time, research offers no compelling reasons for setting these parameters a certain way. Thus, each must be adjusted to achieve a desired goal. Bear in mind that, to improve usability, manufacturers may limit the adjustable parameters.

Avoiding Distortion, Discomfort and Damage

Distortion, discomfort and damage all have the same foundation – intense sounds. Intense sounds force a hearing aid into saturation causing distortion. Intense sounds may be amplified beyond the individual's LDLs causing discomfort. Finally, if left unchecked, intense sounds entering a hearing aid may cause amplification-induced hearing loss. While the latter two problems can be overcome simply by limiting the maximum output by other means, compression is the only way to prevent distortion.

This type of application is referred to as **compression limiting**. Figure 3-1 shows a sample I/O function of a hearing aid with compression limiting. The following are some desirable characteristics of a circuit designed for this purpose.
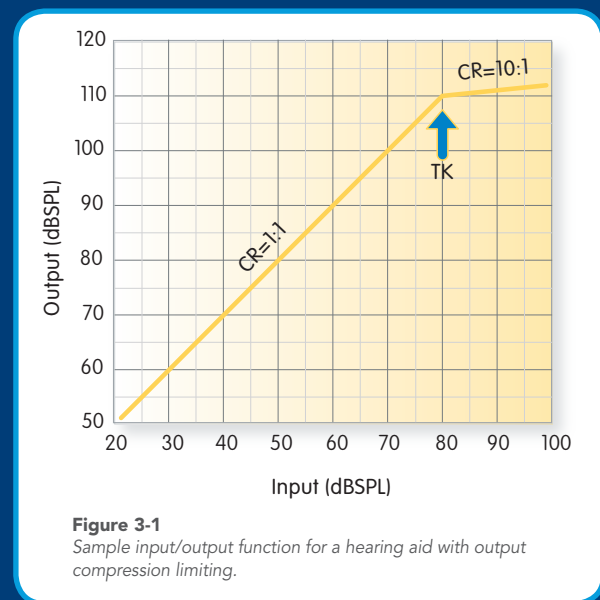
Figure 3-1
Sample input/output function for a hearing aid with output compression limiting.

- AGC-O compression is used so that volume control adjustments do not affect the maximum output of the hearing aid. For example, consider a situation where the individual increases the volume control of the hearing aid to listen to a child's relatively soft voice and a door is slammed shut. If an AGC-I circuit is used, the output of the device may exceed the LDL. In an AGC-O circuit, the maximum output could be set just below the LDL and, more importantly, volume control adjustments would have no impact on the maximum output of the hearing aid. Thus, this application of compression is often called **output compression limiting (OCL)**.

- The TK is set high. Because intense sounds are of primary concern, TKs of 70 dBSPL or greater are typically used. A high CR, greater than 8:1, is used to prevent the amplified sound from exceeding the LDL.

- A fast AT minimizes the overshoot associated with a rapid increase in input level. Thus, ATs of 10 ms or less are generally used to limit the duration for which the output of the hearing aid exceeds the LDL.

- The RT is a less critical element because it is associated with decreasing input and output levels. That said, however, consider the situation where a pot dropping on a tile floor during a conversation forces a hearing aid into compression limiting, resulting in a severe reduction in gain. If gain releases from compression limiting slowly, the speech that follows may be inaudible. Thus, an RT of 100 ms or shorter is preferred; a circuit with an adaptive release time may also be used.

- Finally, single- or multi-channel compression is suitable for this application.

Figure 3-2 shows frequency response curves for devices with and without OCL (Hearing Aids A and B, respectively). Both hearing aids have identical responses for inputs of 50 dBSPL. However, at 90 dBSPL, the output of Hearing Aid B is greater than that of Hearing Aid A. This is an important consideration because the output of Hearing Aid B could exceed the individual's LDL and, if saturation occurs, the associated distortion will be high.

Optimizing Use of the Residual Dynamic Range and Restoring Normal Loudness Perception

As indicated previously, hearing impairment results in a loss of sensitivity for weak sounds, with little or no loss of sensitivity for intense sounds. Thus, in order for the range of environmental sounds to fit within the residual dynamic range of the individual, more amplification is required for weak sounds than for intense sounds. The net result is that weak sounds are audible, moderate sounds comfortable, and intense sounds are perceived as loud without causing discomfort.

This type of application is referred to as **wide dynamic range compression (WDRC)**. Figure 3-3 shows a sample I/O function of a hearing aid with WDRC. The following are some desirable characteristics of a circuit designed for this purpose.

Figure 3-2
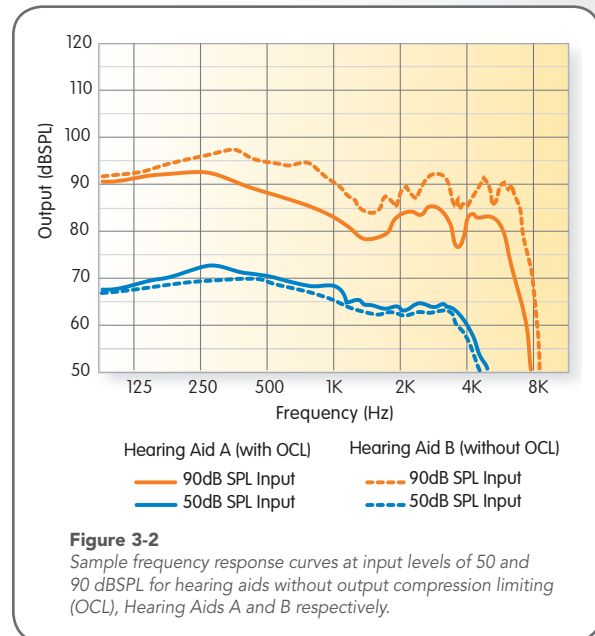*Sample frequency response curves at input levels of 50 and 90 dBSPL for hearing aids without output compression limiting (OCL), Hearing Aids A and B respectively.*
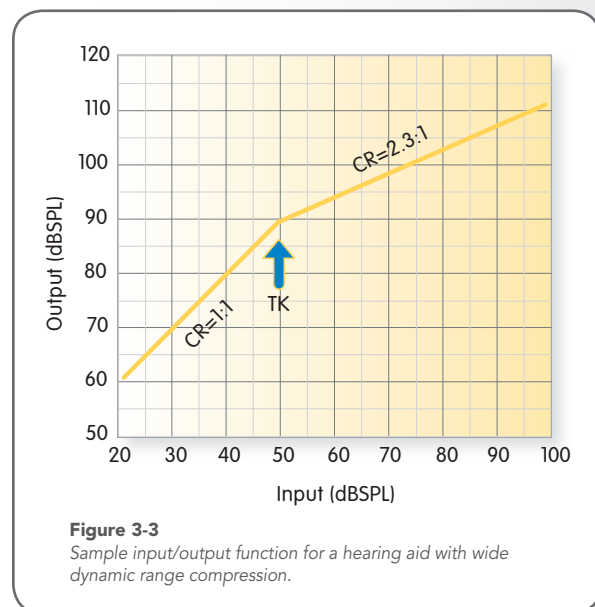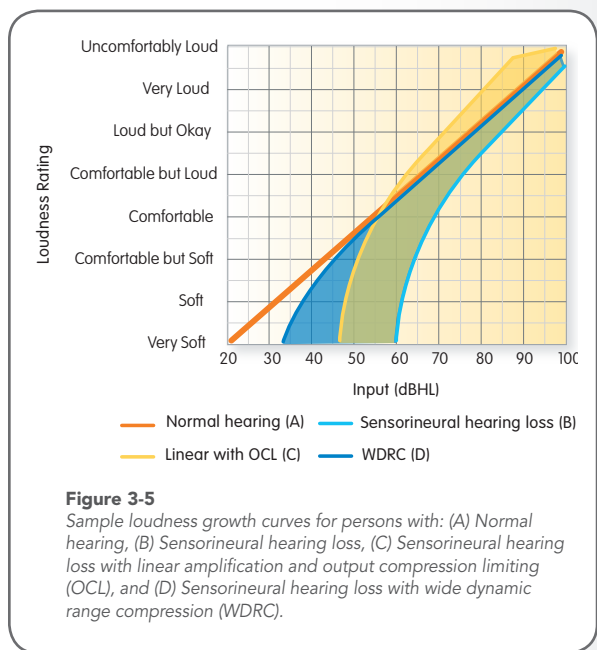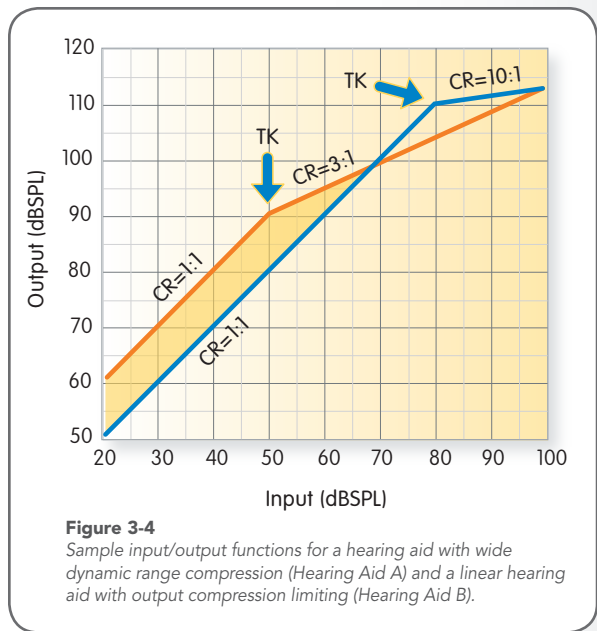

Figure 3-3
Sample input/output function for a hearing aid with wide dynamic range compression.

- AGC-I is used because the goal is to achieve a certain degree of audibility and/or loudness for incoming signals. If average, rather than individual, data are used to derive the amount of amplification necessary, the option may exist to increase or decrease the output of the hearing aid by adjusting the volume control.

- The TK is as low as possible in order to make weak sounds audible. Thus, the TK is typically set at or below 50 dBSPL. Hearing aids with TKs as low as 20 dBSPL are said to use full dynamic range compression (FDRC) because they aim to compress the entire gamut of environmental sounds in the residual dynamic range of the individual.

- Low CRs of 4:1 or less can be used because compression acts over a wider range of inputs. Mueller (2002) gives the example that WDRC is akin to gentle braking as one approaches a stop sign while driving. [By the same token, compression limiting is more like screeching to halt at the last minute!]

- The AT and RT may be fast or slow. However, as indicated earlier, several studies have shown that fast ATs and RTs significantly degrade sound quality (Neuman et al, 1998; Woods et al, 1999). Thus, the use of ATs slower than about 100 ms and RTs slower than 2 s is advisable.

- Multichannel compression is used to accommodate audiometric configurations that deviate substantially from flat. Large variations in the degree of hearing loss across the frequency range also significantly change the perception of loudness.

As shown in Figure 3-4, the I/O function of a hearing aid with WDRC (Hearing Aid A) is visibly different from that of a linear hearing aid with OCL (Hearing Aid B). In addition to the TK and CR being lower, Hearing Aid A also provides more gain than Hearing Aid B at and below the TK. This additional gain results in improved audibility of weak sounds.

Figure 3-4
Sample input/output functions for a hearing aid with wide dynamic range compression (Hearing Aid A) and a linear hearing aid with output compression limiting (Hearing Aid B).

Figure 3-5
Sample loudness growth curves for persons with: (A) Normal hearing, (B) Sensorineural hearing loss, (C) Sensorineural hearing loss with linear amplification and output compression limiting (OCL), and (D) Sensorineural hearing loss with wide dynamic range compression (WDRC).

The effects of hearing impairment on loudness perception and compensation through the use of wide dynamic range compression will be discussed at some length in the next chapter in the context of fitting hearing aids with compression. For now, consider the loudness growth functions shown in Figure 3-5 (page 23). A loudness growth function is a graph showing the perceived loudness of a sound as a function of input level. As expected, the loudness rating increases with input level. However, the shapes of the loudness growth functions are significantly different for persons with normal hearing and those with sensorineural hearing loss. In addition, note that the "loss of loudness" (as measured by the difference between the two curves) is greater for sounds rated as being soft, with little or no loss for sounds rated as loud. Linear amplification with OCL provides a fixed amount of gain regardless of the level of the incoming signal, making all sounds louder by the same amount. Thus, the shape of the loudness growth function remains the same as that for sensorineural hearing loss, but is shifted to the left. The result is that weak sounds may still be inaudible, moderate sounds are comfortable, and intense sounds are too loud. In contrast, when WDRC is used, the resulting loudness growth curve is very similar to that of a person with normal hearing. In other words, weak sounds are now audible, moderate sounds comfortable, and intense sounds are perceived as loud. Note that experienced users of linear amplification may object to the use of WDRC because they have become accustomed to intense sounds being too loud and weak sounds being inaudible.

Maintaining Listening Comfort

Even when the maximum output of a hearing aid is appropriately restricted to within the residual dynamic range, it may still be desirable to further reduce the output of the hearing aid to a level below the maximum output (or LDL)

for much of the time. There are several ways in which this problem may be addressed that do not require the user to constantly adjust the volume control.

First, as discussed in the preceding section, WDRC employs a low TK and CR. This results in intense sounds approaching the maximum output of the hearing aid more gradually than linear amplification coupled with OCL (Figure 3-5). The characteristics of a circuit designed for this purpose are much the same as those described in the previous section for optimizing use of the residual dynamic range and restoring loudness perception. That is, multichannel AGC-I compression should be used with low TK and CR, and long ATs and RTs.

The second method of maintaining listening comfort has no official name, but has been called mid-level or comfort-controlled compression (Byrne and Dillon, 1986). Figure 3-6 shows a sample I/O function of comfort-controlled compression. The following are some desirable characteristics of a circuit designed for this purpose.

- AGC-I compression is used because the goal is to achieve the desired loudness for moderate to intense inputs.

- The TK is set at approximately 60 dBSPL so that compression can operate on moderate and intense sounds.

- Low CRs of 4:1 or less can be used because compression acts over a wider range of inputs than for OCL.

- ATs and RTs may be fast or slow, but should probably be slow to preserve sound quality. Thus, the use of ATs slower than about 100 ms and RTs slower than 2 s is advisable.

- Single- or multi-channel compression is suitable for this purpose.

Figure 3-7 compares mid-level compression (Hearing Aid A), WDRC (Hearing Aid B) and linear amplification with OCL (Hearing Aid C). Note that below the kneepoint, gain for mid-level compression is similar to that for linear amplification with OCL; in contrast, gain is greater for the hearing aid with WDRC. Although it does not increase speech audibility, this alleviates the primary disadvantage of WDRC – increased likelihood of feedback resulting from the provision of more gain. Above the TK, mid-level compression takes advantage of reduced gain for intense sounds when compared to linear amplification with OCL. As a result, like WDRC, this assists in maintaining listening comfort.
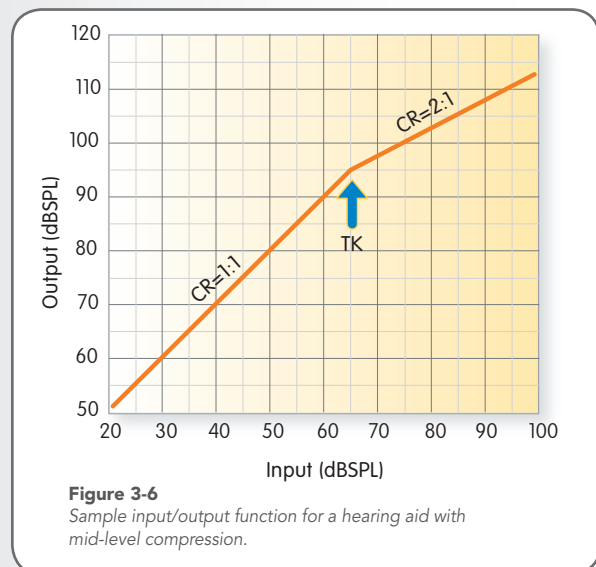

Figure 3-6
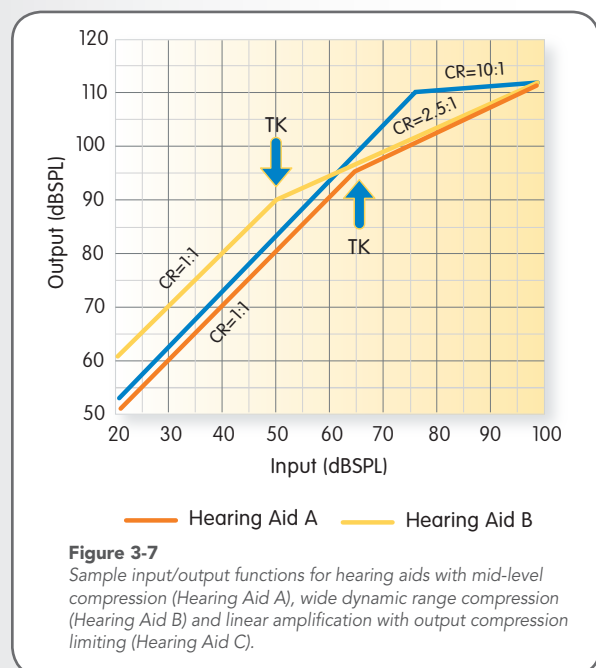*Sample input/output function for a hearing aid with mid-level compression.*

Figure 3-7
Sample input/output functions for hearing aids with mid-level compression (Hearing Aid A), wide dynamic range compression (Hearing Aid B) and linear amplification with output compression limiting (Hearing Aid C).

Maximizing Speech Intelligibility

Arguably, the single most important criterion for maximizing speech intelligibility is increased audibility. That is, speech must be audible before one can be expected to understand it.

Treble-Increase-at-Low-Levels (TILL) was introduced (Killion, 1991) as a means to improve the intelligibility of speech in single-channel devices. As the name suggests, this circuit provides a high-frequency boost for weak sounds. When the overall input level is relatively low, high-frequency consonants are amplified more than the low-frequency vowels. This additional high-frequency gain not only makes the consonants more audible, but also reduces the **upward spread of masking** by the more intense vowels. Although high-frequency emphasis is potentially useful at all input levels, it also increases the likelihood of loudness discomfort at high input levels (Skinner, 1976). Thus, for intense sounds, the frequency response curve is relatively flat. Figure 3-8 shows frequency-gain curves for a TILL circuit at inputs of 50, 70 and 90 dBSPL. Note that, for an input of 50 dBSPL, the high-frequency emphasis is in addition to the gain that would ordinarily be prescribed for the hearing loss. Further, the shape of the curve changes with the input level – i.e., it gets flatter as the input level increases. Improved customization of the frequency response through the use of multiple channels has resulted in the demise of single-channel TILL processing.
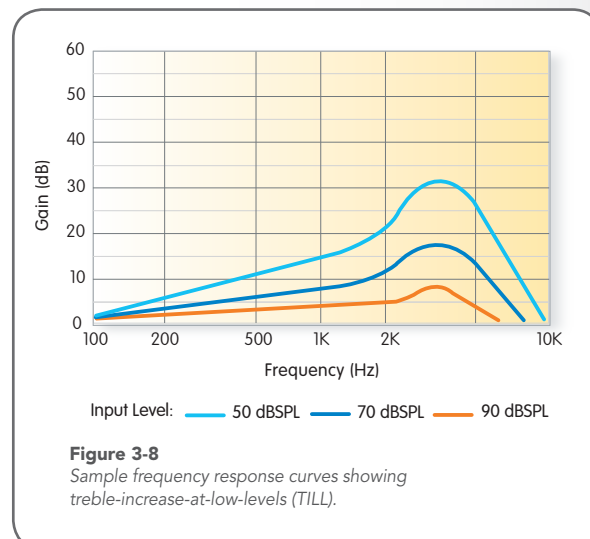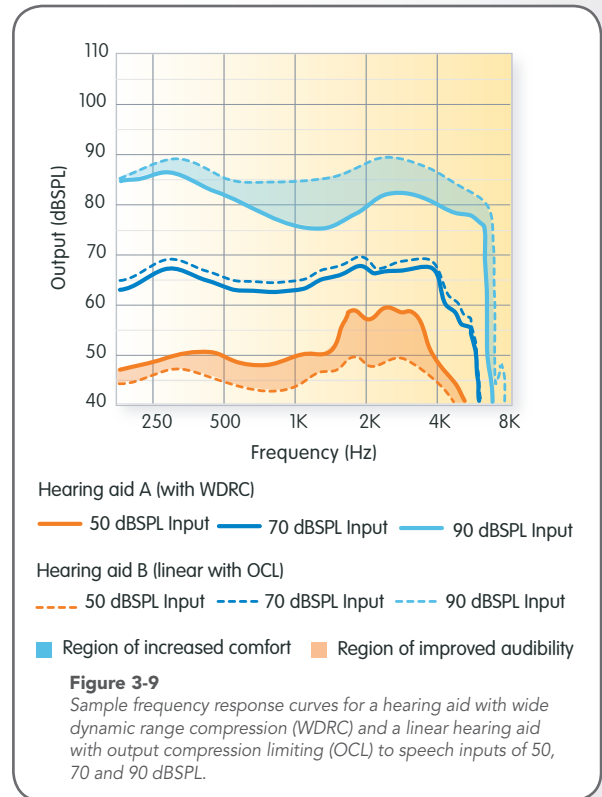

Figure 3-8
Sample frequency response curves showing treble-increase-at-low-levels (TILL).

Today, WDRC is the primary means for maximizing speech intelligibility. The following are some desirable characteristics of a circuit designed for this purpose.

- AGC-I compression is used because amount of gain applied depends on the level of the incoming sound.

- The TK is as low as possible, at or below 50 dBSPL, in order to make the weaker components of speech audible.

- CRs of 4:1 or less can be used because compression acts over a wider range of inputs.

- ATs and RTs should be faster than the duration of a typical syllable to provide more amplification for the weaker components than for the more intense components of speech.

- Multichannel compression is used so that weak consonant sounds can be amplified independently of the more intense vowel sounds. Increasing the consonant-to-vowel ratio (CVR) – i.e., the intensity of a consonant relative to that of a vowel – has been shown to improve speech understanding (Montgomery and Edge, 1988).

The advantage of WDRC over linear amplification with OCL, from the point of view of improving speech audibility, is shown in Figure 3-9. Consider two hearing aids that are set such that conversational speech is comfortable for the person with a sensorineural hearing loss. We have already seen that, for intense sounds, the output of the hearing aid approaches LDL more slowly with WDRC than with linear amplification and OCL. This is indicated in the Figure by the region of increased comfort. More importantly, weak sounds are made audible by WDRC (indicated by the region of improved audibility); the same cannot be said of the linear hearing aid with OCL. Keep in mind also that only average

levels are shown in Figure 3-9. Speech has a dynamic range of 30dB. As a result, there may be components that are even weaker with linear amplification and OCL than shown in the figure.

Hearing aid A (with WDRC)

——— 50 dBSPL Input ——— 70 dBSPL Input ——— 90 dBSPL Input

Hearing aid B (linear with OCL)

---- 50 dBSPL Input ---- 70 dBSPL Input ---- 90 dBSPL Input

■ Region of increased comfort ■ Region of improved audibility

Figure 3-9
Sample frequency response curves for a hearing aid with wide dynamic range compression (WDRC) and a linear hearing aid with output compression limiting (OCL) to speech inputs of 50, 70 and 90 dBSPL.

Reducing the Adverse Effects of Noise

A common complaint regarding hearing aids relates to the effects of noise. Although noise management schemes are beyond the scope of this handbook, the effects of compression will be discussed. The problem with noise is two-fold. First, amplification of intense sounds, such as a vacuum cleaner, may cause the listener distress even if the LDL is not exceeded. And, second, communication is challenging in noisy environments, such as at a party or restaurant.

Hearing aids with Automatic Signal Processing (ASP) were introduced in the 1990s to counter the deleterious effect of background noise. The principle is based on two assumptions: (1) the overall level of sound is relatively high in noisy environments, and (2) the hubbub of parties and/

or restaurants is dominated by energy in the low frequencies. Thus, ASP hearing aids reduce the amount of low-frequency amplification for high input levels to alleviate the loudness of the noise. Low frequency amplification is restored at low input levels. One can think of this as Bass-Increase-at-Low-Levels (BILL), which is another common name for ASP circuits. A further advantage of reduced low-frequency amplification at high input levels is the decreased likelihood of upward spread of masking. Figure 3-10 shows the frequency-gain curves of BILL circuits for speech presented at 50, 70 and 90 dBSPL. Like TILL circuits, BILL processing is seldom used nowadays because of the widespread use of multichannel hearing aids to achieve the same objective.

The compression of sound in multiple channels reduces the adverse effects of noise in several ways that provide an advantage over single-channel BILL processing. First, no assumptions are made regarding the frequency composition of the noise. Thus, gain is reduced in any frequency region where the input levels are high. Second, gain is reduced only in frequency regions where a great deal of noise is present; gain and audibility in the remaining channels are unaffected. Third, when the spectra of the signal and noise are different, multichannel compression may provide a slight improvement in the overall signal-to-noise ratio

(SNR) when the outputs of channels with poor SNR are reduced relative to the outputs of those where the SNR is good. It is important to note that the SNR is not improved within any given channel.

Although it is convenient to compartmentalize the applications of compression, more than one goal can be achieved in a single fitting. For example, multichannel WDRC may be used to optimize use of the residual dynamic range, normalize the perception of loudness, maintain listening comfort, maximize the intelligibility of speech, and reduce the adverse effects of noise. Figure 3-11 summarizes the compression characteristics associated with each application.

Figure 3-10
Sample frequency response curves showing bass-increase-at-low-levels (BILL).
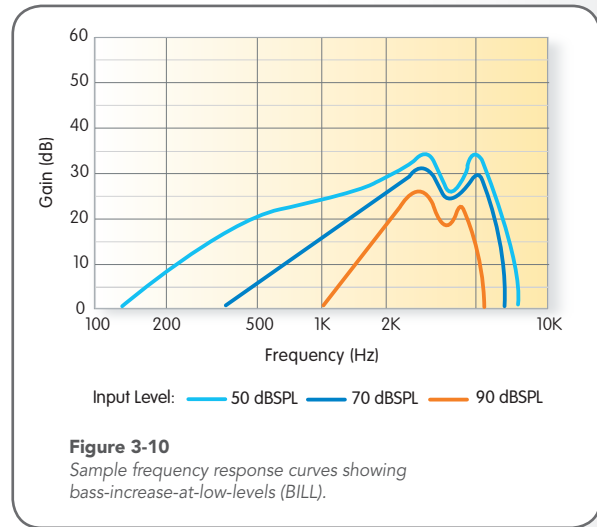
Application	Input-or-Output Control	Threshold Kneepoint	Compression Ratio	Attack & Release Times	Number of Channels
Avoid Distortion, Discomfort and Damage	Output	High	High	Fast	Single or Multiple
Optimize Use of Residual Dynamic Range & Restore Loudness Perception	Input	Low	Low	Faste or Slow (slow preferred)	Multiple
Maintain Listening Comfort	Input	WDRC: Low Mid-Level: High	WDRC: Low Mid-Level: High	Fast or Slow (slow preferred)	Single or Multiple
Maximize Speech Intelligibility	Input	Low	Low	Fast	Multiple

Figure 3-11
Summary of compression applications and associated characteristics.

CHAPTER 4: FITTING COMPRESSION

Factors to Consider

Simply knowing that an individual has a hearing impairment is not sufficient for selecting and fitting a hearing aid. At a minimum, a case history and audiometric evaluation are required. These will provide the professional with insight into audiological and non-audiological factors that must be taken into account when recommending the use of amplification.

Non-audiological considerations include perceived hearing difficulty, cosmetic concerns, cognitive capabilities, and physical limitations. In addition to assisting the professional in selecting a style of hearing aid and suitable features, knowledge of the individual's needs and lifestyle can be used to set appropriate expectations. Although important, discussion of these factors is outside the scope of this handbook.

Degree of Hearing Loss
It has been reported that the amount of perceived benefit from hearing aids is directly correlated with the degree of hearing loss (Gobalek et al, 1988) – that is, the more severe the hearing loss, the greater the benefit from amplification. However, this should not be construed as a contraindication for fitting hearing aids to mild hearing impairments.

Persons with mild to moderately severe hearing loss benefit tremendously from the improved audibility and increased listening comfort offered by WDRC, preferably used in combination with OCL. Traditionally, linear amplification, with or without OCL, has been used for individuals with severe to profound hearing losses. Contrary to the popular belief that this population cannot take advantage of increased audibility for soft sounds, the primary reason for not using WDRC has been the lack of sufficiently powerful hearing aids – recall that WDRC provides more gain at low input levels than linear amplification. Hearing aids utilizing WDRC are now available that provide the additional gain needed for severe and profound hearing losses.

Type of Hearing Loss
A hearing loss always results in a loss of sensitivity for soft sounds. However, the impact on the perception of sounds at suprathreshold levels differs dramatically depending on the type of hearing loss. Further, each presents unique challenges to the hearing aid fitting.

A conductive loss, due to pathology in the outer and/or middle ear, results in an attenuation of the incoming sound. While the degree of attenuation may vary with frequency, it applies equally to weak, moderate and intense sounds (Figure 4-1A). The effect is similar to wearing earplugs – the hearing threshold, MCL, and LDL are all elevated by the amount of conductive loss (Walker, 1997). As a first approximation, simply amplifying all sound levels by the same amount – i.e., linear amplification – is a reasonable solution (Figure 4-1B). However, compression may also be used to avoid the consequences of operating the hearing aid at or close to its maximum output for much of the time.

In contrast, a sensorineural hearing loss results in a greater loss of sensitivity for weak sounds than for moderate or intense sounds. In fact, LDLs may not even increase beyond the normal range for hearing losses of 50dBHL or less (Kamm et al, 1978). Thus, some form of compression is highly desirable for those with sensorineural hearing loss. Unless otherwise specified, the discussion in this book relates to issues surrounding sensorineural hearing loss, with or without a conductive component.
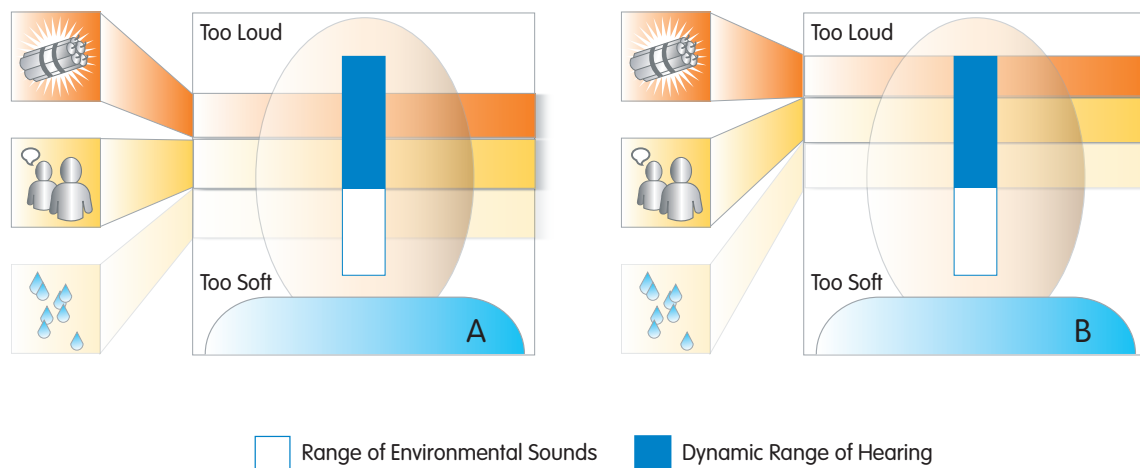
Figure 4-1
The relationship between the range of sounds in the environment and the dynamic range of hearing for persons with: (A) conductive hearing loss, and (B) conductive hearing loss with linear amplification.

Range of Environmental Sounds   *Dynamic Range of Hearing*

Audiometric Configuration

Due to variations in etiology, as well as the anatomy and physiology of the cochlea, the degree of hearing loss often changes across frequency. The typical hearing impairment manifests itself as a greater loss of sensitivity at high frequencies than at low frequencies. Atypical configurations, such as precipitous, reverse slope, and "cookie bite," are also encountered.

The amount of amplification necessary to compensate for a hearing impairment depends on the degree of hearing loss. Multichannel compression provides the professional with a way to finely tailor the frequency and gain characteristics to an individual's audiometric configuration across a broad range of frequencies. Trine and Van Tasell (2002) have shown that 3-4 channels, with adjustable cross-over frequencies, are generally sufficient for this purpose.

Residual Dynamic Range

The upper end of the range of hearing is limited by the LDL, which may or may not be affected by the hearing loss (Kamm et al, 1978). Thus, the residual dynamic range of a person with impaired hearing is considerably smaller than that for a person with normal hearing. How the range of environmental sounds is squeezed into this residual dynamic range depends on the goal of the hearing aid fitting. Regardless of the goal, compression must be applied to the incoming sound.

Loudness Growth

The type and degree of hearing loss, audiometric configuration and residual dynamic range all contribute to variations in the growth of loudness with increasing input levels. In general, more amplification is required for weak sounds than for intense sounds; the amount of amplification required for moderate sounds lies in between these extremes. The goal is for weak sounds to be perceived as soft, moderate sounds to be perceived as comfortable, and intense sounds to be perceived as loud by the hearing aid user. Varying amounts of gain depending on the input level can only be achieved through the use of compression.

Although the loudness growth function can be predicted to some extent, the use of individual loudness data may enhance customization of the hearing aid fitting. [As described in the previous chapter, a loudness growth function is a graph depicting the perceived loudness of sound as a function of input level.] These data can be obtained via testing under headphones, in a sound field, or directly through the hearing aid (a feature available in many high-end digital hearing aids). While sound in principle, the benefits of the time spent obtaining loudness growth data for individuals are debatable.

Unilateral Versus Bilateral Amplification

There are several documented advantages of fitting hearing aids bilaterally – increased speech understanding in background noise and/or reverberation (Nabalek and Pickett, 1974), better sound quality (Balfour and Hawkins, 1992), improved localization ability (Byrne et al, 1992), avoidance of auditory deprivation (Silman et al, 1984), and binaural summation of loudness. Only the last of these benefits directly impacts the hearing aid fitting.

Binaural summation refers to the phenomenon whereby a sound is louder if heard in two ears than in one. This loudness increase is to the order of 3dB at threshold (Dermody and Byrne, 1975), and approximately 6dB for moderate (Christen, 1980) and intense sounds (Scharf and Fishken, 1970). Interestingly, binaural loudness summation has little or no effect on the LDL (Hawkins et al, 1987). This implies that, in a bilateral fitting, less gain (and more compression) is required in each hearing aid (compared to a unilateral fitting) to restore loudness perception. Further, this can be achieved without causing loudness discomfort.

Previous Hearing Aid Use

WDRC can be used to restore loudness perception for new users of amplification. However, the onset of hearing loss is gradual and its impact insidious. It has been reported that the typical individual lives with hearing loss for about 10 years before seeking help (Kochkin, 1991). Such a person has generally forgotten what it was like to hear "normally," resulting in a distorted recollection of the perception of loudness. Thus, less than the prescribed amount of gain and/or more compression may be used initially, with the possibility of moving closer to the prescription as the individual acclimates to the amplified sound.

Experienced users of linear amplification present a different challenge. WDRC is often rejected with the complaints like "these hearing aids are too soft," and "I hear better with my own hearing aids." This is in spite of improved

audibility with WDRC! The problem is related to the perception of moderate-to-intense sounds. Recall that, with linear amplification, intense sounds are amplified more than necessary, even when average conversational speech is comfortable. Although undesirable, long time users of linear hearing aids have become accustomed to overamplification of moderate-to-intense sounds. Viable options include mid-level compression or easing into WDRC over a period of weeks or months.
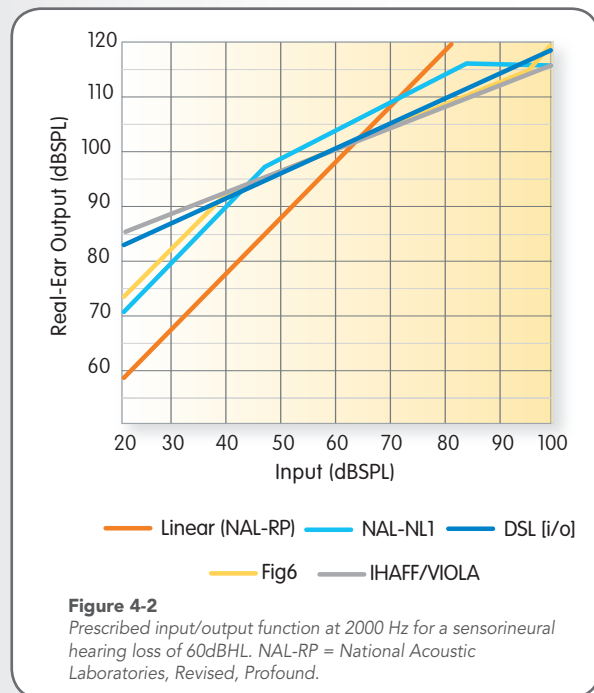
Prescribing Amplification

Given the multitude of hearing losses and the complexity of hearing aids, the only practical way to begin the hearing aid fitting process is to prescribe amplification based on some audiological characteristic of the person with impaired hearing. Prescriptive approaches use a formula to generate an amplification target, which serves as a reasonable starting point.

The earliest prescriptions were developed to fit the available linear technology. Thus, linear prescriptive formulas – like NAL-RP, POGO, Berger and Libby 1/3 gain – provide a single target for moderate sounds, because the most common and important sound to which we listen is average conversational speech. However, these do not take into consideration the level-dependent gain characteristics of modern hearing aids. This has led to the development of non-linear prescriptions aimed at restoring loudness perception and improving the intelligibility of speech.

Depending on the underlying philosophy, non-linear prescriptive approaches can be broadly divided into two categories: loudness normalization and loudness equalization. As the term suggests, loudness normalization attempts to restore normal loudness perception within each frequency band. Presumably, this also results in normal loudness perception of the overall sound – i.e., once all the frequency bands are combined. Loudness equalization,

on the other hand, aims to maximize speech intelligibility by making all frequency bands of speech contribute equally to the loudness of speech. Although not normal within each frequency band, normal loudness perception is restored for the overall sound. The four most commonly-used, widely-accepted and well-researched non-linear prescriptive approaches will be discussed – NAL-NL1,DSL [i/o],FIG6 and IHAFF/VIOLA. Variations in the core principles result in different prescriptions from each of these approaches. Figure 4-2 compares the I/O functions prescribed by each of these approaches for a sensorineural hearing loss of 60dBHL. The direction and magnitude of these differences are further related to the degree of hearing loss and audiometric configuration.

Figure 4-2
Prescribed input/output function at 2000 Hz for a sensorineural hearing loss of 60dBHL. NAL-RP = National Acoustic Laboratories, Revised, Profound.

NAL-NL1

NAL-NL1 (National Acoustic Laboratories, non-linear, version 1; Dillon, 1999) specifies insertion gain at 1/3-octave frequencies between 125 and 8000 Hz. The NAL-NL1 formula has three unique features. First, the goal is to maximize speech intelligibility while maintaining the overall loudness equivalent to that for a person with normal hearing. Thus, loudness equalization is the underlying philosophy for this approach.

Second, corrections are applied to the target to accommodate conductive and mixed hearing losses. And, third, it takes into account the limited ability of the impaired ear to extract useful information from an audible speech signal (Ching et al, 1998). This is especially true for severe-to-profound hearing losses.
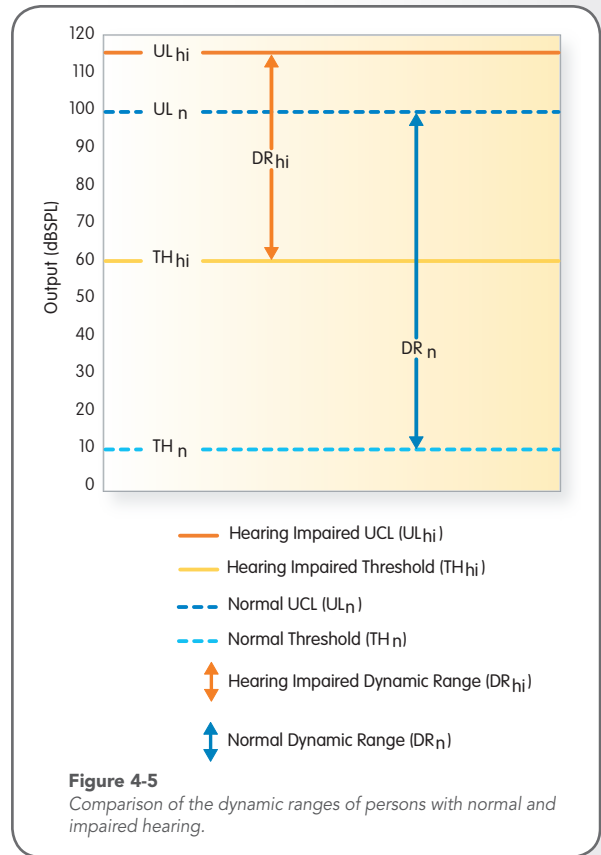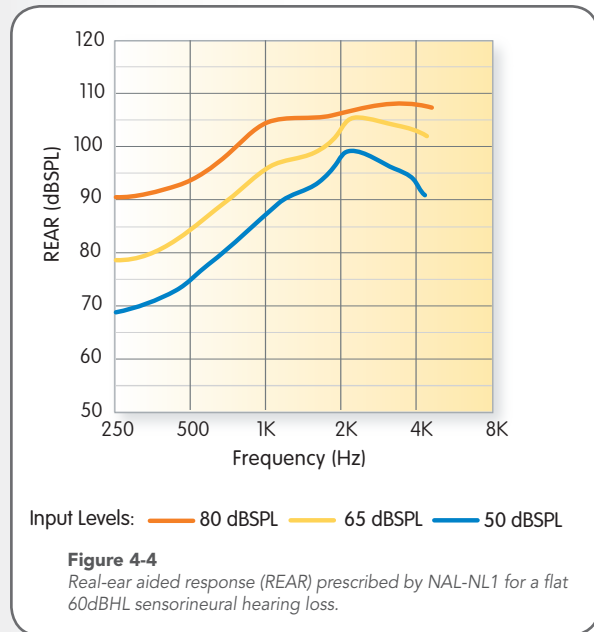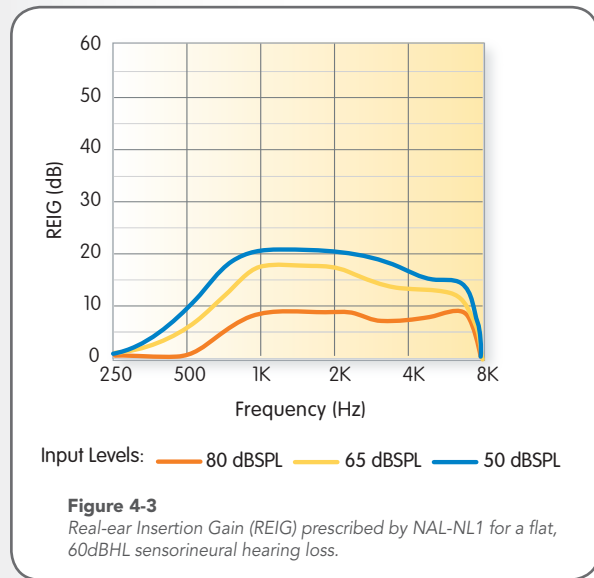
The prescription is dependent upon the threshold at a particular frequency, the 3-frequency average threshold, the slope of the audiogram between 500 and 2000 Hz, and the long-term average speech spectrum (LTASS) (Byrne et al, 1994). For a speech input of 65 dBSPL, the target generated by NAL-NL1 is the same as that prescribed by NAL-RP. Although traditionally shown as real-ear insertion gain (REIG) (Figure 4-3, page 32), targets can be displayed in various forms, including insertion gain, aided gain or I/O function measured in a real ear, 2-cc coupler or ear simulator. A customized prescription is obtained by measuring the individual real-ear unaided gain (REUG) and real ear-to-coupler difference (RECD). The number of channels in the hearing aid can be specified. Also, appropriate corrections are applied for bilateral fittings. Finally, the fitting software may provide recommendations regarding the dimensions of the vent and the type of tubing used in the earmold.

Figure 4-4 (page 32) shows the real-ear aided response (REAR) prescribed by NAL-NL1 for a flat hearing loss of 60dBHL. Amplification targets for input levels of 50, 65 and 80 dBSPL are displayed.

DSL [i/o]

Desired Sensation Level [input/output] (Cornelisse et al, 1995) is based on the premise that the desired sensation level for a variety of sounds can be predicted from threshold and LDL. Based on loudness normalization, the goal of this approach is to ensure audibility for weak sounds, provide comfort for average conversational speech, and prevent discomfort for intense sounds. There are actually two DSL [i/o] procedures. In DSL [i/o] linear, the I/O function is a straight line over a wide range of

inputs, i.e., the CR is fixed. Note that, despite the use of the term "linear," compression is still being applied over a range of input levels. DSL [i/o] curvilinear, on the other hand, has a curved I/O function where the CR is constantly changing. Because DSL [i/o] linear is more commonly used and better developed than DSL [i/o] curvilinear, only the former will be discussed here. Although developed and widely used for the fitting of children, this approach can be successfully applied to adult hearing aid fittings.

Figure 4-3
Real-ear Insertion Gain (REIG) prescribed by NAL-NL1 for a flat, 60dBHL sensorineural hearing loss.

Figure 4-4
Real-ear aided response (REAR) prescribed by NAL-NL1 for a flat 60dBHL sensorineural hearing loss.

Figure 4-5
Comparison of the dynamic ranges of persons with normal and impaired hearing.

DSL [i/o] attempts to map the normal dynamic range into the residual dynamic range of the individual with impaired hearing. As shown in Figure 4-5, the dynamic range is bounded by the threshold of audibility – TH_n and TH_{hi} for normal and hearing-impaired, respectively – and the upper limit of comfort – UL_n and UL_{hi}. An I/O function can be derived by connecting these points (Figure 4-6). Note that the I/O function has three distinct segments: a region of linear gain below the TKlow, a region of compression between TKlow and TKhigh, and a region of output limiting above TKhigh. At a minimum, threshold data are required to obtain target information. Patient-specific information such as RECD, **real ear-to-dial difference (REDD), real-ear unaided response (REUR)**, and LDL can be entered to customize the prescription. The style of hearing aid and type of compression can also be entered to provide a more accurate target.
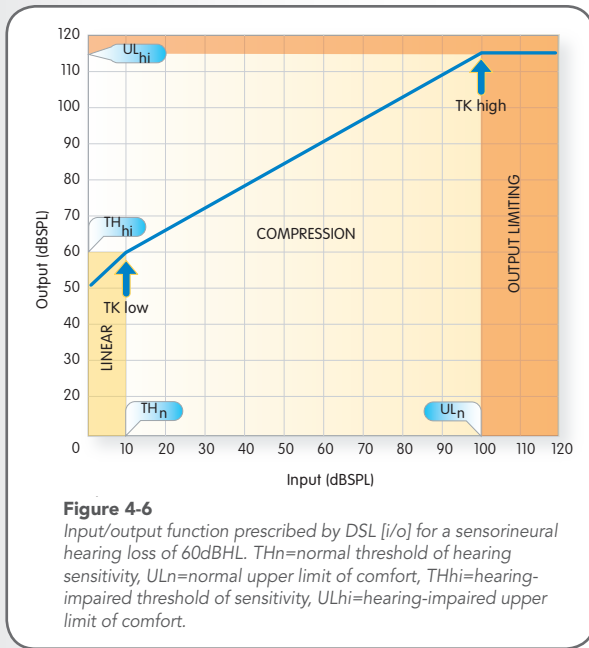
Figure 4-6
Input/output function prescribed by DSL [i/o] for a sensorineural hearing loss of 60dBHL. THn=normal threshold of hearing sensitivity, ULn=normal upper limit of comfort, THhi=hearing-impaired threshold of sensitivity, ULhi=hearing-impaired upper limit of comfort.

Figure 4-7 shows the REAR prescribed by DSL [i/o] for a flat hearing loss of 60dBHL. Amplification targets for input levels of 50, 65 and 80 dBSPL are displayed.
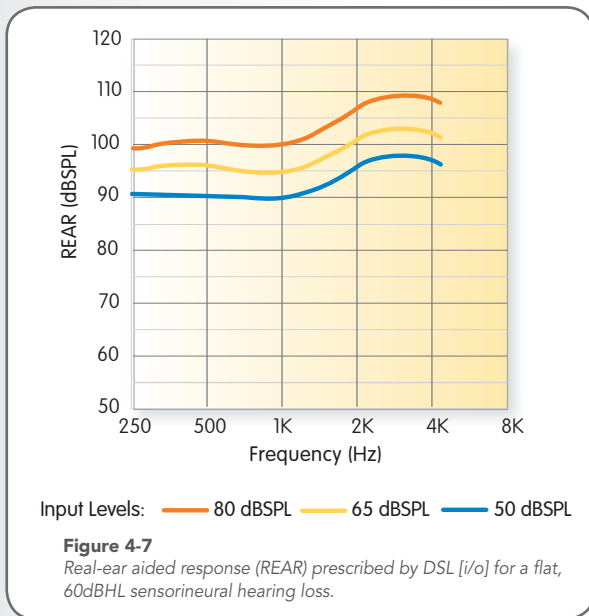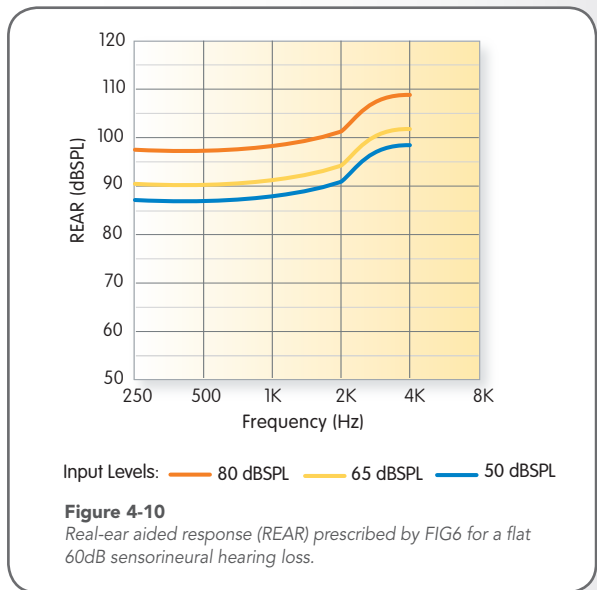

Input Levels: —— 80 dBSPL —— 65 dBSPL —— 50 dBSPL

Figure 4-7
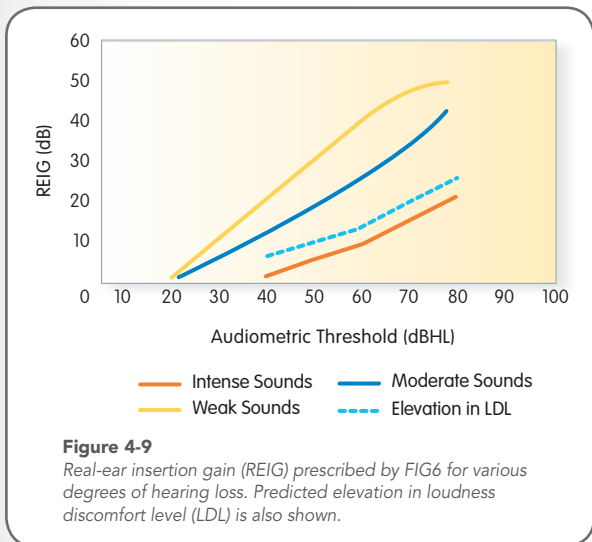*Real-ear aided response (REAR) prescribed by DSL [i/o] for a flat, 60dBHL sensorineural hearing loss.*
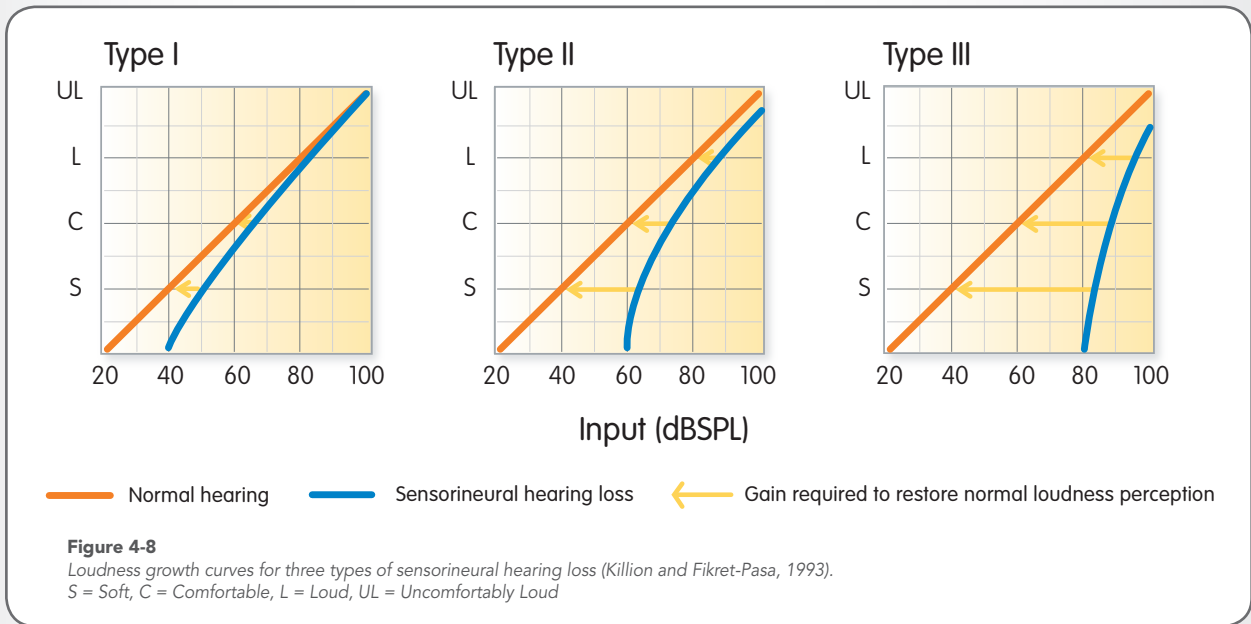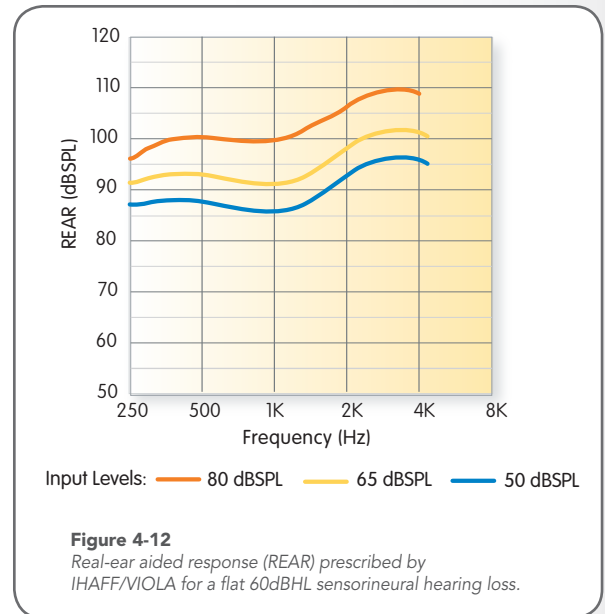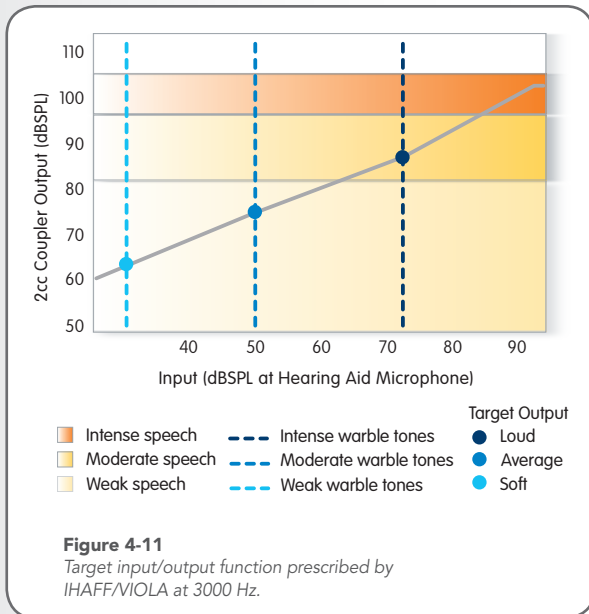
Fig6

Fig6, named for its original appearance in an article by Killion and Fikret-Pasa (1993), is a loudness normalization procedure based on average data. Killion and Fikret-Pasa identified three types of sensorineural hearing loss based on loudness perception. Type I hearing losses demonstrate a threshold sensitivity loss of about 40dB with normal perception of intense sounds (Figure 4-8A, page 34). For a Type II loss, the loss of threshold sensitivity of about 60dB is accompanied by a small decrease in sensitivity to intense sounds (Figure 4-8B, page 34). Finally, the loss of sensitivity for both soft and intense sounds is substantial for a Type III loss (Figure 4-8C, page 34). The distance between the loudness growth curves for persons with normal and impaired hearing provides an indication of the gain required.

IHAFF/VIOLA

In the mid 1990s, a group of engineers, clinicians and researchers formed the Independent Hearing Aid Fitting Forum (IHAFF) to address the issue of non-linear amplification. Believing that there was little direction for the dispensers in terms of fitting non-linear amplification, they developed a fitting protocol named for the group. The IHAFF protocol is comprised of three components: a loudness scaling procedure (Contour Test; Cox et al, 1997), a fitting algorithm to specify hearing aid characteristics (Visual Input/Output Locator Algorithm, VIOLA), and a measure of the fitting outcome (Abbreviated Profile of Hearing Aid Benefit, APHAB; Cox and Alexander, 1995). The goal of the VIOLA fitting algorithm is to normalize loudness perception and will be the focus of discussion for this handbook.

VIOLA generates a target 2-cc coupler output for weak, moderate and intense sounds based on the predicted loudness growth function for warble tones (Figure 4-11). The line connecting these target values represents the desired I/O function. According to Valente and Van Vliet (1997), a TK of 40-45 dBSPL should be used. It is recommended that target I/O functions be obtained for at least one low and one high frequency (e.g., 500 and 3000 Hz, respectively).

Figure 4-8
Loudness growth curves for three types of sensorineural hearing loss (Killion and Fikret-Pasa, 1993).
S = Soft, C = Comfortable, L = Loud, UL = Uncomfortably Loud

Figure 4-9
Real-ear insertion gain (REIG) prescribed by FIG6 for various degrees of hearing loss. Predicted elevation in loudness discomfort level (LDL) is also shown.

Figure 4-10
Real-ear aided response (REAR) prescribed by FIG6 for a flat 60dB sensorineural hearing loss.

Figure 4-11
Target input/output function prescribed by IHAFF/VIOLA at 3000 Hz.

Figure 4-12
Real-ear aided response (REAR) prescribed by IHAFF/VIOLA for a flat 60dBHL sensorineural hearing loss.

The shaded regions in Figure 4-11 represent target levels for weak, moderate and intense speech. These are higher than the target levels for warble tones because speech is a broadband signal (i.e., power summation). The targets can be further customized by measuring the individual loudness growth function via the Contour Test.

Figure 4-12 shows the REAR prescribed by IHAFF/VIOLA for a flat hearing loss of 60 dBHL. Amplification targets for input levels of 50, 65 and 80 dBSPL are displayed.
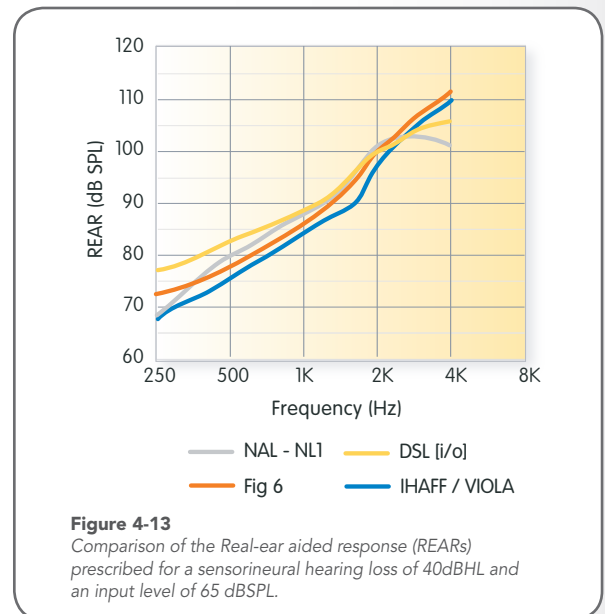
As is obvious from Figures 4-2, 4-4, 4-7, 4-10 and 4-12, the various prescriptive approaches differ in the amount of gain they recommend for weak, moderate and intense sounds, at low and high frequencies. Figure 4-13 compares the real-ear output prescribed by the various approaches for an input of 65 dBSPL. To some extent, discrepancies are to be expected due to the differences in the underlying philosophy. This discrepancy poses a dilemma from the point of view of choosing the "right" approach. However, it may be a moot point when one considers the prescriptive method a good starting point rather than a final destination.

Figure 4-13
Comparison of the Real-ear aided response (REARs) prescribed for a sensorineural hearing loss of 40dBHL and an input level of 65 dBSPL.

CHAPTER 5: EXPANSION

Issues Resulting from WDRC

As we have seen throughout this handbook, WDRC provides additional gain for weak inputs, as compared to linear amplification. The advantage of this approach is increased speech intelligibility, especially for soft speech, and normalized loudness perception. The bad news is that it also amplifies other soft sounds, including the drone of a refrigerator, the creaking of floor boards and the hum of an air conditioner. How can this be a disadvantage when a person with normal hearing is able to hear these everyday sounds?

Presbycusis, the most common cause of hearing loss in adults, typically has a gradual onset. As a result, by the time the hearing loss becomes noticeable to the individual, many of the weaker sounds in the environment may not have been heard for some time. WDRC enables amplification of weak sounds to within the audible range. This can be overwhelming to an auditory system that has been deprived for a long period of time. On a different note, individuals who have good residual hearing (i.e., thresholds that are within or close to the normal range) at some frequencies may be able to hear noise that is inherent to the hearing aid, such as circuit noise that is amplified by the hearing aid. Either of these scenarios may result in an unpleasant experience and, in turn, hearing aids that spend most of their time in a drawer. To avoid this rejection, the problem must be addressed.

The first approach to tackling the "problem" of increased audibility for weak sounds is counseling. Potential hearing aid users often do not realize that, in addition to hearing the soft voice of a grandchild, they will also hear other sounds in the environment that they may not have heard in a long time. It is important that they understand that these sounds are also heard by persons with normal hearing and that, over time, they will become accustomed to the abundance of environmental sounds. Some individuals are especially sensitive to these sounds and, as such, are eager for an immediate solution. Moreover, counseling will not alleviate the problem when it is caused by audible circuit noise. Gain reduction has traditionally been the second line of attack in this situation. Specifically, the gain for weak sounds is decreased in an attempt to reduce the audibility of environmental and/or circuit noise. The drawback of this solution is that it defeats the purpose of using WDRC in the first place – weak sounds are now no more audible than they would be with linear amplification or, worse, with no amplification at all. Further, the audibility of important speech sounds is adversely affected. Thus, alleviating one problem has given rise to a second, and arguably more serious, issue.

A more viable solution to the predicament, and one that is commonly available in modern digital hearing aids, is the use of expansion. Expansion is designed to make a hearing aid sound silent in quiet environments. One can think of it as a noise reduction strategy for quiet environments. The use of expansion can lead to greater listener satisfaction by reducing the intensity of weak environmental sounds and circuit noise, without sacrificing speech audibility and intelligibility.
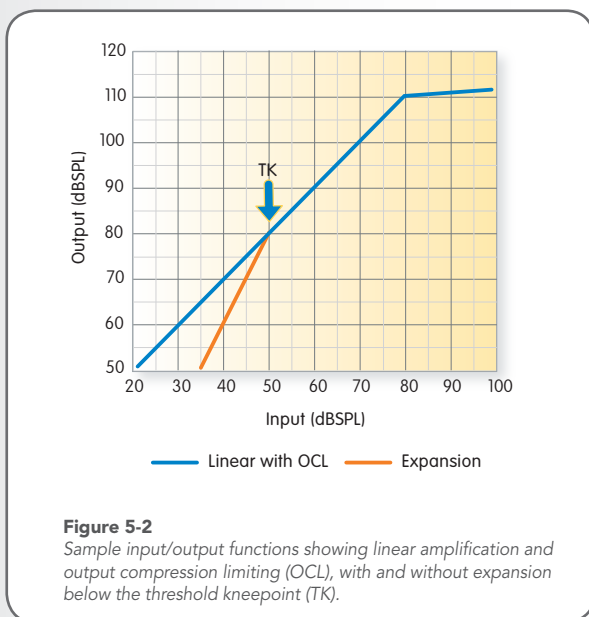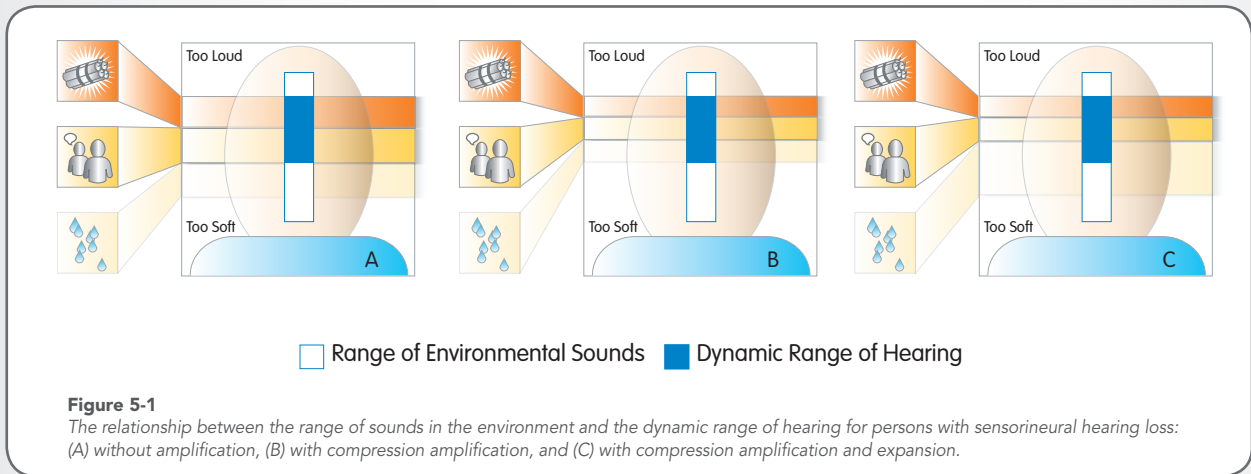
What is Expansion?

Consider the following example. Figure 5-1A compares the residual dynamic range of a person with sensorineural hearing loss to the range of sounds in the environment. As expected, weak sounds are inaudible, while intense sounds are perceived as loud. Figure 5-1B illustrates the effect of compression amplification – weak sounds are amplified to a greater extent than moderate and intense sounds. This allows the range of environmental sounds to fit within the residual dynamic range of the individual.

While this improves the intelligibility of speech, it also makes unwanted sounds audible, such as the quiet drone of a refrigerator. Figure 5-1C shows the effect of expansion in combination with compression. The band representing weak sounds is wider than those for moderate or intense sounds. Further, the weakest of the weak sounds are below the dynamic range, and only the more intense components of the weak sounds are audible to the individual. This is the principle behind expansion – the softer sounds are amplified less than louder sounds. In hearing aids, expansion is only applied to the weakest sounds in the environment. A comparison of Figures 5-1B and 5-1C shows that the bands representing moderate and intense are identical. In other words, compression is applied to moderate and intense sounds, while expansion is used only for weak sounds.

Figure 5-2 shows an I/O function for a linear hearing aid with OCL. A fixed amount of gain is applied to all input levels, as is characteristic of linear amplification. Specifically, for every 10dB change in input level, the output also changes by 10dB. When expansion is applied at low input levels (i.e., at or below the TK of 50 dBSPL), the output changes by 20dB for a 10dB change in input level (orange curve in Figure 5-2). Note that the I/O function above the TK is unaffected by whether or not expansion is applied.

Range of Environmental Sounds   Dynamic Range of Hearing

Figure 5-1
The relationship between the range of sounds in the environment and the dynamic range of hearing for persons with sensorineural hearing loss: (A) without amplification, (B) with compression amplification, and (C) with compression amplification and expansion.

Figure 5-2
Sample input/output functions showing linear amplification and output compression limiting (OCL), with and without expansion below the threshold kneepoint (TK).

Like compression, expansion delivers different amounts of gain depending on the input level. However, in expansion, the gain increases as the input level increases. This is in contrast to compression where the gain decreases as the input level increases. The change in gain as a function of input level is most easily illustrated by an I/G function. Figure 5-3 (page 38) shows the I/G function of a linear hearing aid, and the effect of expansion. In a linear hearing aid with no expansion, the gain remains constant below kneepoint. In a hearing aid with expansion, however, little or no gain is applied to input levels of 20 dBSPL or lower. As the input level increases, so does the gain, until it reaches a maximum of 30dB at the TK (i.e., 50 dBSPL). Note once again that the I/G function above the TK is unaffected by whether or not expansion is applied.
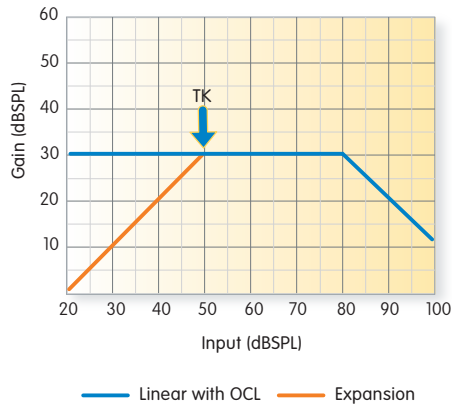
Figure 5-3
Sample input/gain functions showing linear amplification and output compression limiting (OCL), with and without expansion below the threshold kneepoint (TK).
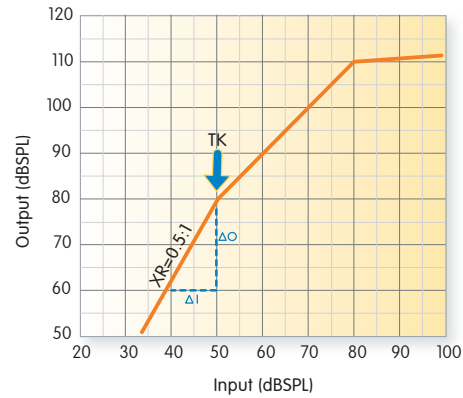

Figure 5-4
Identifying the threshold kneepoint (TK) for expansion and calculating the expansion ratio (XR) from an input/output function.

Characterizing Expansion

Expansion is characterized in much the same way as compression, via a threshold kneepoint, expansion ratio, and attack and release times. There are, however, a few key differences which will be highlighted here. [Note that, at the time of this writing, there are no standards for defining the characteristics of expansion.]

Expansion Threshold
Expansion threshold (XT) is the input level below which expansion operates. For the hearing aid shown in Figure 5-4, the XT is 50 dBSPL. Because it looks like a bent knee in an I/O function, the XT is also referred to as a TK.

The potential for confusion arises when expansion is coupled with compression, both of which are characterized by a TK. The TKs for both expansion and compression frequently occur at the same input level. Thus, as shown in Figure 5-5A, expansion operates below the TK, while compression takes over above it. A less commonly implemented design is one where the TK for compression (TK$_{comp}$) is higher than the TK for expansion (TK$_{exp}$) (Figure 5-5B). Note that amplification in the portion of the I/O function in between TK$_{exp}$ and TK$_{comp}$ is linear (i.e., CR=1:1).

A low TKexp (input levels of 50 dBSPL or lower) permits the use of more gain to ensure maximum speech audibility. In this scenario, the hearing aid may be in expansion for only a small portion of time and only in very, very quiet environments. As a consequence, the problem of a "noisy hearing aid" may not be alleviated at all. On the other hand, a high TKexp (input levels greater that 40 dBSPL) will result in a quiet hearing aid, but at the expense of speech audibility.

Expansion Ratio
Like its counterpart in compression, the **expansion ratio (XR)** is an indicator of the extent to which the input signal is expanded. Specifically, XR is calculated using the formula:

$$XR = \frac{\Delta Input}{\Delta Output}$$

For the hearing aid shown in Figure 5-4, increasing the input from 40 to 50 dBSPL (ΔInput = 10dB) increased the output from 60 to 80 dBSPL (ΔOutput = 20dB). Using the above formula, the XR of the hearing aid is:

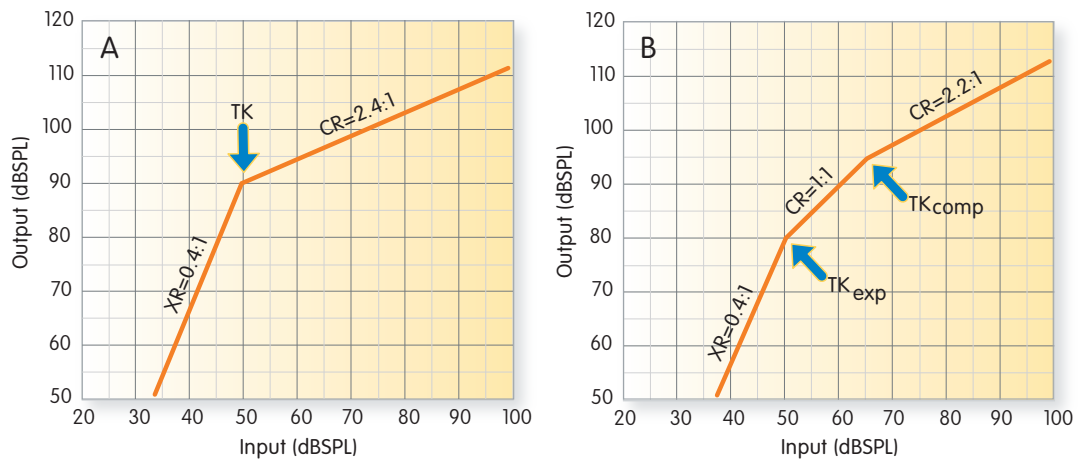$$XR = \frac{\Delta Input}{\Delta Output} = \frac{10}{20} = 0.5$$

Figure 5-5
Sample input/output functions for hearing aids with: (A) the same threshold kneepoint (TK) for compression and expansion, and (B) a threshold kneepoint for compression (TK_comp) higher than the threshold kneepoint for expansion (TK_exp).

CRs are always greater than 1 when compression is applied. Further, the larger the number, the greater is the degree of compression applied. On the other hand, XRs between 0 and 1 indicate the application of expansion. Figure 5-6 shows I/O functions for three hearing aids with the same TK but different XRs. Hearing Aid A, with an XR of 1, is linear below the TK. The I/O function for Hearing Aid C (XR=0.4) has a steeper slope than that for Hearing Aid B (XR=0.7). A steeper expansion slope indicates a greater degree expansion. Thus, in general, the smaller the XR (i.e., closer to 0), the steeper is the slope of the I/O function and the greater is the degree of expansion.
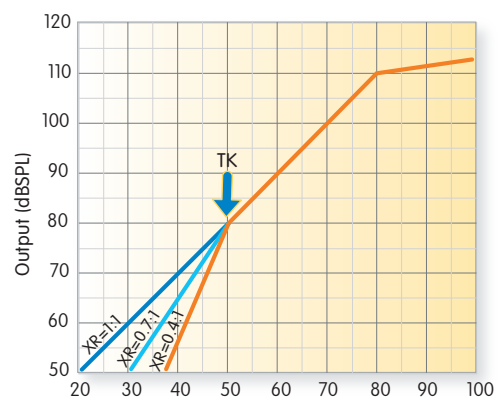
The higher the XR (i.e., closer to 1), the more likely it is that weak sounds will be amplified to within the residual dynamic range. In other words, the likelihood that a person will be able to hear the weakest of sounds is greatest at high XRs. On the other hand, low XRs are known to adversely affect the intelligibility of speech at low input levels (Plyler et al, 2005a). Either problem may be exacerbated by the exact location of the TK. Specifically, applying a lot of expansion (low XR) at high TKs may drastically reduce speech audibility, while applying only a little expansion (high XR) at low TKs may result in the hearing aid sounding noisy.

Figure 5-6
Sample input/output functions showing three hearing aids with the same threshold kneepoint (TKexp) and different expansion ratios (XR).

Attack and Release Times

Because expansion only occurs below the TK, attack and release are defined differently for expansion than for compression. Attack into expansion occurs when the signal level drops below the TK, while release from expansion occurs when the signal level exceeds the TK.

A fast AT limits the amplification of ambient sounds in the environment, and implies that the hearing aid may go into expansion during pauses in speech. Slow RTs are known to adversely affect speech intelligibility (Plyler et al, 2005b). This occurs because gain is considerably reduced when the hearing aid is in expansion, and is slow to recover when the input level increases; the small amount of gain may render speech inaudible. The effects of both of these problems are magnified if the TK is high and the XR is low. Thus, in theory, it appears that a combination of a slow AT and a fast RT offers the best balance, making weak environmental sounds inaudible with little effect on the audibility of speech.

Measuring Expansion

Expansion operates on low-level inputs. Ambient noise levels in most environments are often too high (i.e., above the TK) for expansion to engage. As a result, it can be difficult to hear and/or measure the effects of expansion.

While a sound treated room or anechoic chamber are not necessary, an important prerequisite for listening to and/or measuring expansion is that the environment be quiet. Objections to the hum of the refrigerator or circuit noise occur primarily when the hearing aid user is in a quiet environment, such as reading a book in the living room. Thus, a quiet office, away from the sounds of copy machines and people talking, will suffice for evaluating expansion.

Informal Listening

With expansion turned off, listen for quiet sounds (e.g., whirr of a computer, hum of the air conditioner, etc.). Turn on expansion and listen for those same sounds. The hearing aid should sound much quieter with expansion turned on. The difference between the two conditions will be more prominent at high TKs and low XRs.
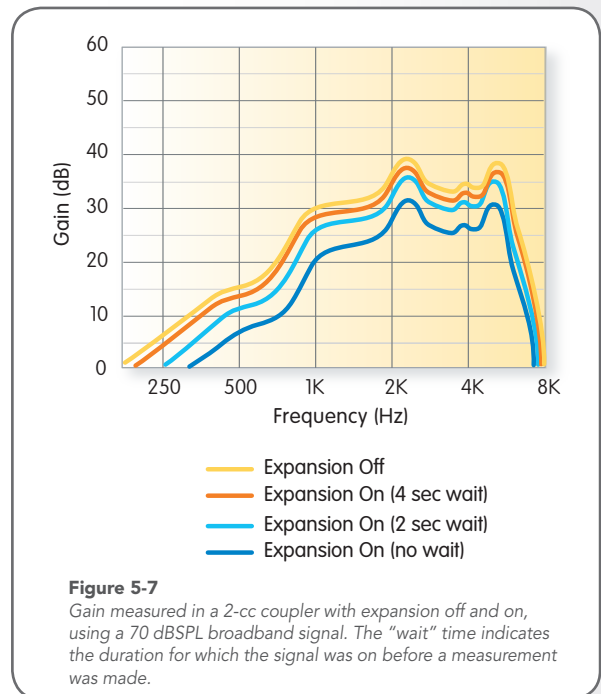

Figure 5-7
Gain measured in a 2-cc coupler with expansion off and on, using a 70 dBSPL broadband signal. The "wait" time indicates the duration for which the signal was on before a measurement was made.

Coupler Measurement

The effect of expansion can be measured using a standard 2-cc coupler, a test box and a broadband input signal such as white, pink, speech-shaped or composite noise. Figure 5-7 shows the gain of a hearing aid for a composite noise with expansion turned off and an overall input level of about 70 dBSPL (or above the TKexp of the hearing aid). To complete the measurement, turn expansion on and turn the stimulus in the test box off for longer than the AT to allow expansion to engage. Once this has occurred, turn the stimulus on again and make another measurement immediately ("Expansion On (no wait)" in Figure 5-7). You should find that the gain of the hearing aid is lower with expansion on than with expansion off. The longer the interval of time between turning on the stimulus and making a measurement, the

greater the gain of the hearing aid, as shown by "Expansion on (2 sec wait)" and "Expansion on (4 sec wait)" in Figure 5-7. The duration of time required for the gain to return to the expansion off condition is an indication of the expansion RT of the hearing aid.

While expansion is a useful tool for the audiologist, it should be used judiciously – experienced hearing aid users and those with severe hearing loss may welcome the added audibility provided by WDRC.

The prescriptive approaches described previously are based on average and/or predicted data. As such, they serve as a good starting point. Persons with impaired hearing come with a variety of experiences with hearing. It is a rare individual who does not require some further adjustment after the initial fitting of a hearing aid. This chapter describes techniques for troubleshooting the individual's complaints and fine-tuning the hearing aids. The ultimate goal should always be to maximize speech intelligibility, with minimal loudness discomfort.

Before attempting to fine-tune a hearing aid, consider this: we tend to gravitate toward conditions with which we are most familiar, and hearing aid users are no exception. For example, an experienced user of linear amplification generally prefers it over WDRC, even though the latter may provide better speech understanding and listening comfort. Another example of preference based on experience is that of a new hearing aid user preferring amplification that essentially matches the response of the unaided ear. Thus, at the initial fitting, the clinician should keep in mind how these experiences may affect the outcome of the fitting.
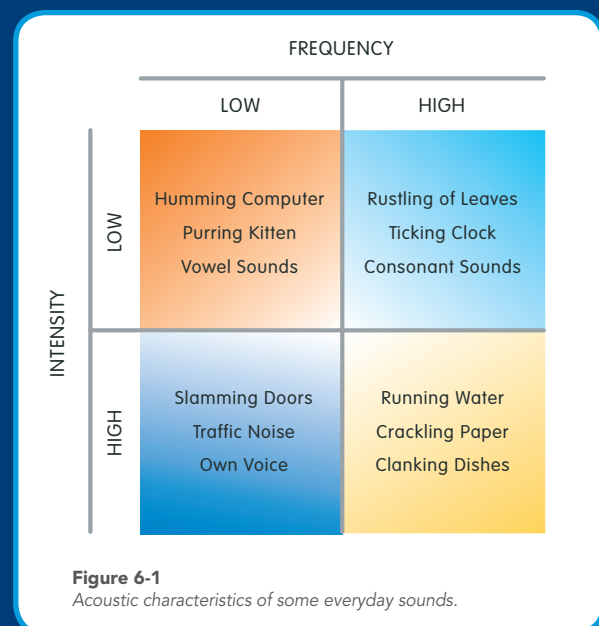
A Systematic Approach

There are two primary challenges to fine-tuning a hearing aid fitting – recognizing the source and possible solutions to the problems being described, and understanding the compression architecture of the specific hearing aid.

Recognizing the Problem and Solution

A systematic three-step approach is recommended for recognizing the problem and identifying potential solutions: (1) Simplify the individual's complaints, (2) determine the direction of change needed, and (3) identify the control(s) to effect the desired change. [Note that the examples cited in this section are only meant to represent the types of

problems that the clinician may encounter and one approach to dealing with them. A more detailed list can be found in the section on "Common Complaints and Solutions."]

Most users of hearing aids are unable to describe their complaints in hearing aid terms – gain, frequency, compression, etc. In a survey of approximately 300 audiologists, Jenstad et al (2003) reported over 40 frequently-used descriptors – "too loud," "booming," "can't hear well in noise," "in a barrel," and so on. Thus, a single problem may be described in multiple ways! The only way to get to the root of the problem is by asking appropriate questions. For example, if the hearing aids reportedly "sound funny," find out if this occurs with all sounds or only with the individual's own voice. The key to translating patient complaints is a good working knowledge of the acoustic characteristics of sounds. Figure 6-1 shows examples of sounds that are commonly associated with complaints. Thus, if the patient's own voice "sounds funny," the problem may be simplified to a need for adjusting the response of the hearing aid to intense, low-frequency sounds.

FREQUENCY		
	LOW	HIGH
INTENSITY LOW	Humming Computer Purring Kitten Vowel Sounds	Rustling of Leaves Ticking Clock Consonant Sounds
INTENSITY HIGH	Slamming Doors Traffic Noise Own Voice	Running Water Crackling Paper Clanking Dishes

Figure 6-1
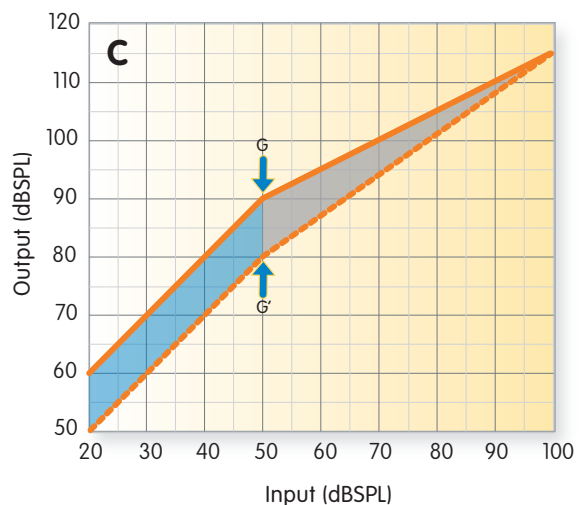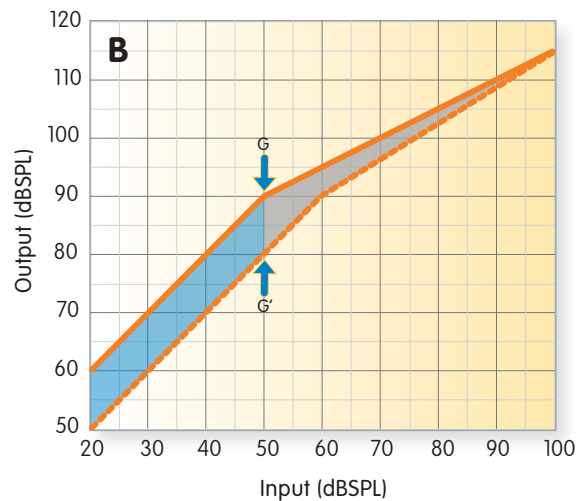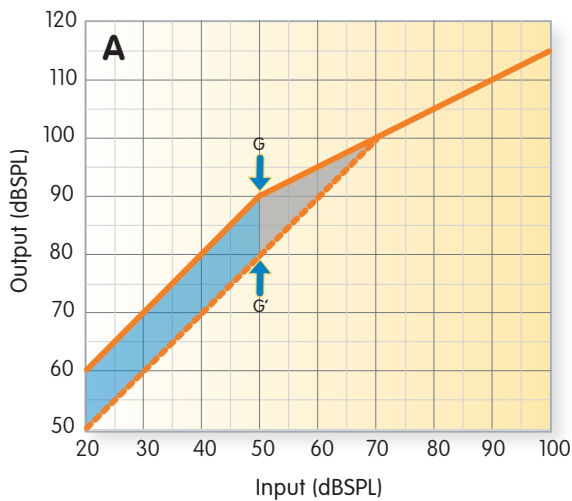*Acoustic characteristics of some everyday sounds.*

Figure 6-2
Sample input/output functions for three approaches to decreasing the gain for weak sounds by 10 dB. The original and adjusted gains are indicated by G and G' respectively. The blue shaded area shows the effect of the adjustment in the desired region; the gray shaded area demonstrates secondary effects of the change.

Determining the direction of change needed is the next step in addressing the problem. The good news is that, once the problem has been clarified, audiologists generally agree on the course of action. Frequently, this is the easiest part of the process because the direction of change is immediately apparent from the complaint. For example, the expressions "too loud," "uncomfortable," "boomy," "harsh," etc. indicate too much amplification, at least in some frequency regions and/or at some intensities. However, some complaints, such as that of a plugged sensation, are more difficult to interpret. For example, it may be that, when the ears are plugged (with hearing aids), sounds in the environment are harder to hear but one's own voice sounds louder than usual. Thus, if sound of the patient's own voice "sounds funny" due to a plugged sensation, one solution may be to decrease the gain for intense, low-frequency sounds.

With the root of the problem and a possible solution recognized, identifying the control(s) to effect the necessary change should be obvious. For example, to alleviate the problem of the patient's own voice sounding funny due to a plugged sensation, the gain for intense, low-frequency sounds may be decreased by adjusting a control in the low-frequency channel that manipulates gain for loud sounds. However, the appropriate control is sometimes either not available or cannot be adjusted directly. In such situations alternative solutions must be sought (for example, increasing the CR to decrease the gain for intense, low-frequency sounds), which requires a firm grasp of the mechanics of compression.

Understanding the Compression Architecture

One potential complication to fine-tuning hearing aids based on patient complaints is that alleviating one problem may inadvertently give rise to another.

For example, consider the complaint of soft speech being too loud, which requires a reduction in the gain for weak sounds. Figure 6-2 (page 43) shows three different ways in which gain for inputs of 50 dBSPL may be decreased by 10dB. In Figure 6-2A, the TK is increased from 50 to 70 dBSPL. This results in a 10-dB decrease in gain for inputs below 50 dBSPL (shown by the blue shaded region), indicating that the desired outcome of reducing gain for weak sounds is achieved. The gain for moderate sounds (50-70 dBSPL) is also inadvertently reduced (shown by the gray shaded region), but there is no impact on the gain for intense sounds (90 dBSPL). In Figure 6-2B, the gain for weak sounds is reduced by increasing the TK from 50 to 60 dBSPL. However, this action also decreases the gain for moderate and intense sounds. Finally, in Figure 6-2C, the gain for weak sounds is reduced simply by decreasing the amount of gain without affecting the TK. This impacts not only the gain for weak sounds, but also that for moderate and intense sounds. Thus, it appears that, regardless of the exact mechanism of gain change, secondary effects of the adjustment are virtually unavoidable. Whether or not this poses a problem depends on the accompanying complaints. If moderate and/or intense sounds are also reported as being loud, then this secondary effect is desirable. On the other hand, if the adjustment results in moderate and/or intense sounds becoming too soft, then a compromise must be made.
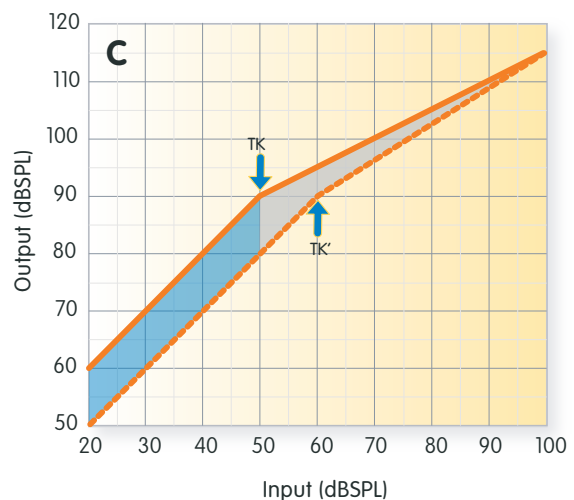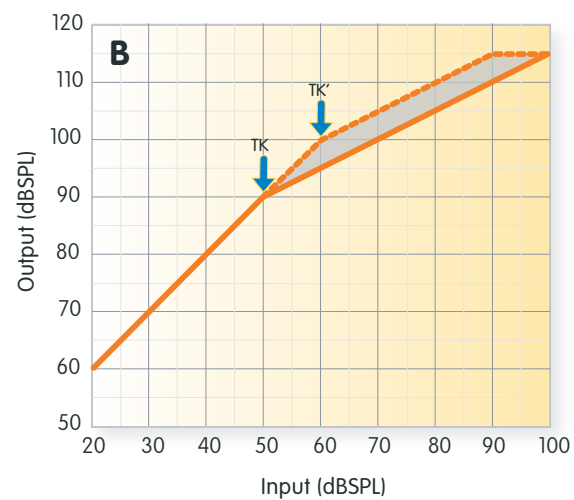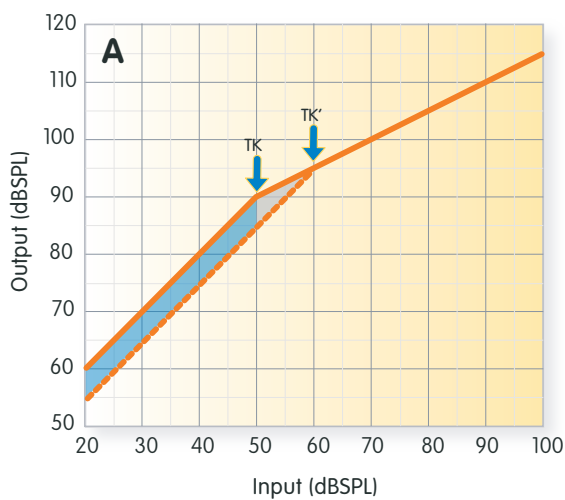

Figure 6-3
Sample input/output functions for three approaches to adjusting the threshold kneepoint. The original and adjusted kneepoints are indicated by TK and TK' respectively. The blue shaded area shows the effect of the adjustment in the desired region; the gray shaded area demonstrates secondary effects of the change.

A word of caution against assumptions about the function of various compression controls. For example, consider the commonly held notion that the gain for weak sounds can be decreased by increasing the TK. The I/O functions in Figure 6-3 demonstrate three ways of increasing the TK by 10dB. In Figure 6-3A, this action does indeed decrease the gain for weak sounds. In contrast, not only does increasing the TK not decrease the gain for weak sounds in Figure 6-3B, it actually increases the gain for moderate and intense sounds! In addition to not alleviating the primary complaint, this adjustment may give rise to a secondary complaint of moderate and intense sounds being too loud. Finally, as shown in Figure 6-3C, increasing the TK may decrease gain for all sounds.

These examples highlight the need for comprehending the compression architecture of the particular hearing aid being adjusted. Stated differently, the clinician must understand the impact that fine-tuning will have in the intended region as well as any secondary effects. This is especially useful for predicting and minimizing secondary complaints.

Common Complaints and Solutions

Just as there are multiple ways to describe a problem, there are often several solutions to a given problem (Jenstad et al, 2003). The following is a list of some common complaints and recommended solutions. Each problem is first categorized according to the general nature of the complaint, followed by a more specific description of the complaint, possible causes, recommended solutions, and other factors and options that should be considered. Note that this is not intended to be an exhaustive catalog of problems and solutions. Rather, it should be treated as a general guideline. Further, as indicated in the preceding section, the precise mechanism for addressing some of these changes will depend on the compression architecture of the particular hearing aid.

COMPLAINT	"My Voice Sounds Boomy"
ALTERNATE DESCRIPTIONS	"Sounds like I'm speaking with my head in a barrel (or tunnel)," "my voice echoes," "my voice sounds hollow," "my voice sounds like I have a cold," or "my ears feel plugged."
POSSIBLE CAUSE	Excessive low-frequency gain at high input levels.
RECOMMENDED SOLUTION(S)	Reduce gain for intense, low-frequency sounds.
OTHER CONSIDERATIONS	• Occlusion due to the presence of the hearing aid in the ear. To test, turn off the hearing aid and have the person vocalize /i/; the problem is occlusion if the "boomy" sensation persists. Address the problem acoustically by increasing the size of the vent. • Individual may have forgotten the sound of his/her own voice due to long-standing hearing loss.

OWN VOICE QUALITY

<div style="writing-mode: vertical-lr;">OWN VOICE QUALITY</div>

COMPLAINT	"My Voice Sounds Muffled"
ALTERNATE DESCRIPTIONS	"Hearing aid cuts out when I speak," "hearing aid is weak."
POSSIBLE CAUSE	Insufficient low-frequency gain for high input levels.
RECOMMENDED SOLUTION(S)	• Increase gain for intense, low-frequency sounds. • Increase maximum output at low frequencies.
OTHER CONSIDERATIONS	• Individual may be accustomed to linear amplification. • May be associated with complaints about other intense sounds (e.g., noisy restaurants, music, etc.) sounding muffled.

<div style="writing-mode: vertical-lr;">OWN VOICE QUALITY</div>

COMPLAINT	"My Voice Sounds Distorted"
ALTERNATE DESCRIPTIONS	"Hearing aid crackles when I speak," "my voice sounds unnatural."
POSSIBLE CAUSE	Hearing aid is in saturation.
RECOMMENDED SOLUTION(S)	• Change to compression limiting. • Increase maximum output. • Decrease gain for intense, low-frequency sounds.
OTHER CONSIDERATIONS	• Problems may be associated with complaints about other intense sounds (e.g., noisy restaurants, music, etc.) sounding distorted or unnatural. • Individual may have forgotten the sound of his/her own voice due to long-standing hearing loss. • Evaluate distortion of hearing aid per ANSI specifications (ANSI, 2003).

SPEECH INTELLIGIBILITY

COMPLAINT	"I Hear Better Without My Hearing Aids"
ALTERNATE DESCRIPTIONS	"Difficulty understanding speech in quiet," "speech is unclear," "hearing aid makes no difference," "difficulty understanding TV (or radio)."
POSSIBLE CAUSE	Insufficient gain.
RECOMMENDED SOLUTION(S)	• Increase overall gain. • Increase gain for high-frequency sounds. • Decrease overall gain (if person reports reducing the volume control to improve listening comfort).
OTHER CONSIDERATIONS	• Individual may have poor speech recognition ability unrelated to audibility and/or unrealistic expectations regarding impact of hearing aid on speech understanding. • Hearing aid may not be perceived as being loud enough due to previous experience with linear amplification.

SPEECH INTELLIGIBILITY

COMPLAINT	"I Can't Hear Well at Restaurants"
ALTERNATE DESCRIPTIONS	"Don't hear well in noise," "difficulty understanding speech in background noise."
POSSIBLE CAUSE	Insufficient gain.
RECOMMENDED SOLUTION(S)	• Increase overall gain. • Increase gain for high-frequency sounds. • Decrease gain for low-frequency sounds.
OTHER CONSIDERATIONS	• Individual may have poor speech recognition ability unrelated to audibility and/or unrealistic expectations regarding impact of hearing aid on speech understanding. • Hearing aid may not be perceived as being loud enough due to previous experience with linear amplification. • Recommend use of technology to improve the SNR (e.g., directional microphones).

SPEECH INTELLIGIBILITY

COMPLAINT	"I Hear People at a Distance Better Than Those at My Table"
ALTERNATE DESCRIPTIONS	"Hear better at a distance."
POSSIBLE CAUSE	Insufficient gain for high input levels.
RECOMMENDED SOLUTION(S)	• Increase gain for intense sounds. • Increase maximum output.
OTHER CONSIDERATIONS	• May be associated with complaints about other intense sounds (e.g., own voice, music, etc.) sounding muffled. • If sounds at a distance are reported as being too loud, consider decreasing gain for weak sounds. May be associated with complaints about other weak sounds being too loud.

LOUDNESS

COMPLAINT	"Hearing Aids are Too Loud"
ALTERNATE DESCRIPTIONS	"Background noise is too loud," "voices are too loud," "hearing aid is boomy," "sirens are too loud," "refrigerator hum is too loud."
POSSIBLE CAUSE	Excessive gain.
RECOMMENDED SOLUTION(S)	• Decrease overall gain if all sounds are too loud. • Decrease gain for intense sounds if only intense sounds are too loud. • Decrease gain for weak sounds if only weak sounds are too loud. • Use expansion if only weak sounds are too loud.
OTHER CONSIDERATIONS	New hearing aid users may have become accustomed to a quieter world due to long-standing hearing loss.

LOUDNESS

COMPLAINT	"Loud Sounds are Uncomfortable"
ALTERNATE DESCRIPTIONS	"Sounds are painful," "sound of clattering dishes is too loud," "sound of running water is uncomfortable."
POSSIBLE CAUSE	Excessive output for loud and/or transient inputs.
RECOMMENDED SOLUTION(S)	• Decrease the maximum output. • Decrease gain for intense sounds. • Decrease gain for intense, high-frequency sounds.
OTHER CONSIDERATIONS	New hearing aid users may have become accustomed to a quieter world due to long-standing hearing loss.

LOUDNESS

COMPLAINT	"Hearing Aids are Too Soft"
ALTERNATE DESCRIPTIONS	"Not loud enough," "sounds are not clear."
POSSIBLE CAUSE	Insufficient gain.
RECOMMENDED SOLUTION(S)	• Increase overall gain if all sounds are too soft. • Increase gain for soft sounds if weak sounds are too soft. • Increase gain for loud sounds if intense sounds are too soft. • Increase maximum output if intense sounds or all sounds are too soft.
OTHER CONSIDERATIONS	• May be associated with complaints regarding difficulty understanding speech (especially in quiet), or other intense sounds (e.g., own voice, music, etc.) sounding muffled. • Hearing aid may not be perceived as being loud enough due to previous experience with linear amplification.

SOUND QUALITY

COMPLAINT	"Hearing Aids are Noisy"
ALTERNATE DESCRIPTIONS	"Refrigerator hum is too loud," "noisy when I'm reading in my living room."
POSSIBLE CAUSE	Excessive gain for low input levels.
RECOMMENDED SOLUTION(S)	• Use expansion. • Decrease gain for soft sounds.
OTHER CONSIDERATIONS	Evaluate circuit noise of hearing aid per ANSI specifications (ANSI, 2003).

COMPLAINT	"Hearing Aids have Static"
ALTERNATE DESCRIPTIONS	"Sounds are distorted," "crackling sound in noisy environments."
POSSIBLE CAUSE	Hearing aid is in saturation.
RECOMMENDED SOLUTION(S)	• Change to compression limiting. • Increase maximum output. • Decrease gain for intense sounds.
OTHER CONSIDERATIONS	• Problem may be associated with complaints about intense sounds (e.g., noisy restaurants, music, own voice, etc.) sounding distorted or unnatural. • Evaluate distortion of hearing aid per ANSI specifications (ANSI, 2003).

SOUND QUALITY	COMPLAINT	"Hearing Aids Cut in and Out"
	ALTERNATE DESCRIPTIONS	"Pumping sound," "background noise fades in and out," "level of sound in hearing aids fluctuates."
	POSSIBLE CAUSE	Short time constants.
	RECOMMENDED SOLUTION(S)	• Increase attack and/or release time. • Decrease compression ration (i.e., make the hearing aids more linear).
	OTHER CONSIDERATIONS	Verify that the problem is not related to intermittency.

SOUND QUALITY	COMPLAINT	"Hearing Aids Shut Down with Loud Sounds"
	ALTERNATE DESCRIPTIONS	None.
	POSSIBLE CAUSE	OCL with slow release time.
	RECOMMENDED SOLUTION(S)	• Increase maximum output. • Use faster release time.
	OTHER CONSIDERATIONS	Verify that the problem is not related to intermittency or battery drainage.

SOUND QUALITY	COMPLAINT	"Sounds are Too Sharp"
	ALTERNATE DESCRIPTIONS	"Sounds harsh," "sounds tinny," "too much treble."
	POSSIBLE CAUSE	Excessive high-frequency gain (relative to low-frequency gain).
	RECOMMENDED SOLUTION(S)	• Decrease high-frequency gain. • Increase low-frequency gain.
	OTHER CONSIDERATIONS	Individual's auditory perception may be distorted due to long-standing high-frequency hearing loss.

SOUND QUALITY	COMPLAINT	"Sounds are Too Dull"
	ALTERNATE DESCRIPTIONS	"Sounds boomy," "sounds hollow," "too much bass."
	POSSIBLE CAUSE	Excessive low-frequency gain (relative to high-frequency gain).
	RECOMMENDED SOLUTION(S)	• Decrease low-frequency gain. • Increase high-frequency gain.
	OTHER CONSIDERATIONS	• May be associated with occlusion due to the presence of the hearing aid in the ear. • Individual's auditory perception may be distorted due to long-standing hearing loss.

FEEDBACK	COMPLAINT	"Hearing Aids Whistle"
	ALTERNATE DESCRIPTIONS	"Squeal."
	POSSIBLE CAUSE	Excessive gain.
	RECOMMENDED SOLUTION(S)	• Decrease gain for weak, high-frequency sounds. • Decrease gain for weak sounds. • Decrease overall gain.
	OTHER CONSIDERATIONS	• Address excessive acoustic leakage due to ill-fitting hearing aid or a vent that is larger than necessary. • Use electronic feedback solutions such as notch filters and feedback cancellers.

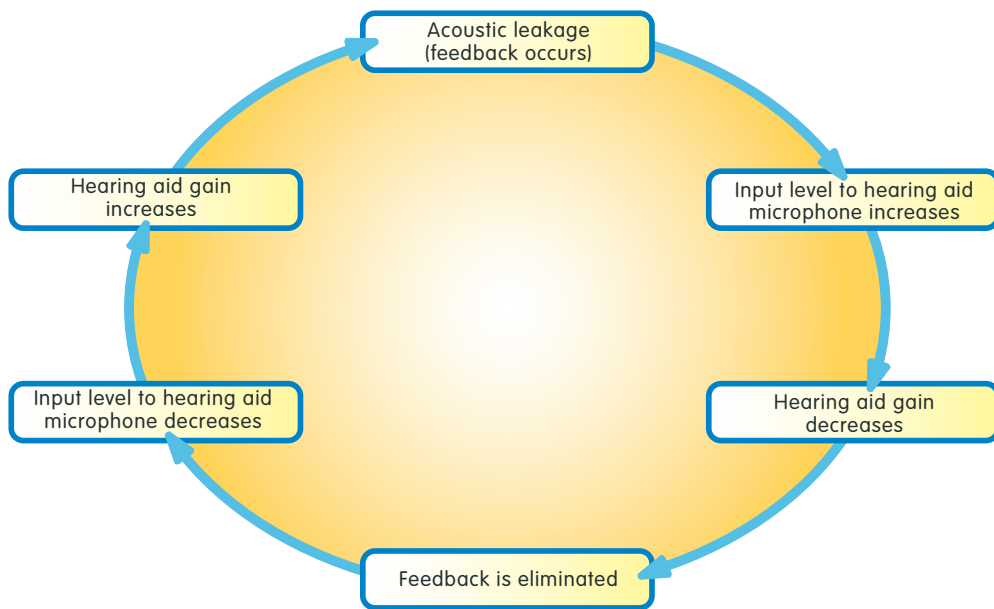| COMPLAINT	"Chirping Noise"
ALTERNATE DESCRIPTIONS	"Whooping sound," "beeping sound."
POSSIBLE CAUSE	Interaction between acoustic leakage and compression (Figure 6-4).
RECOMMENDED SOLUTION(S)	• Decrease gain for weak, high-frequency sounds. • Decrease gain for weak sounds. • Decrease overall gain.
OTHER CONSIDERATIONS	• Address excessive acoustic leakage due to ill-fitting hearing aid or a vent that is larger than necessary. • Use electronic feedback solutions such as notch filters and feedback cancellers.

FEEDBACK

Figure 6-4
Mechanism behind "chirping" feedback.

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