The Compression Handbook

Fourth Edition

An overview of the characteristics and applications of compression amplification
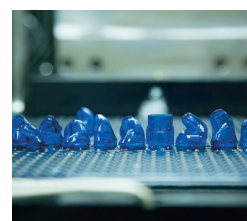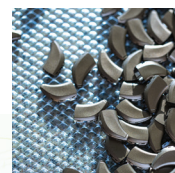
Starkey Hearing **Technologies**
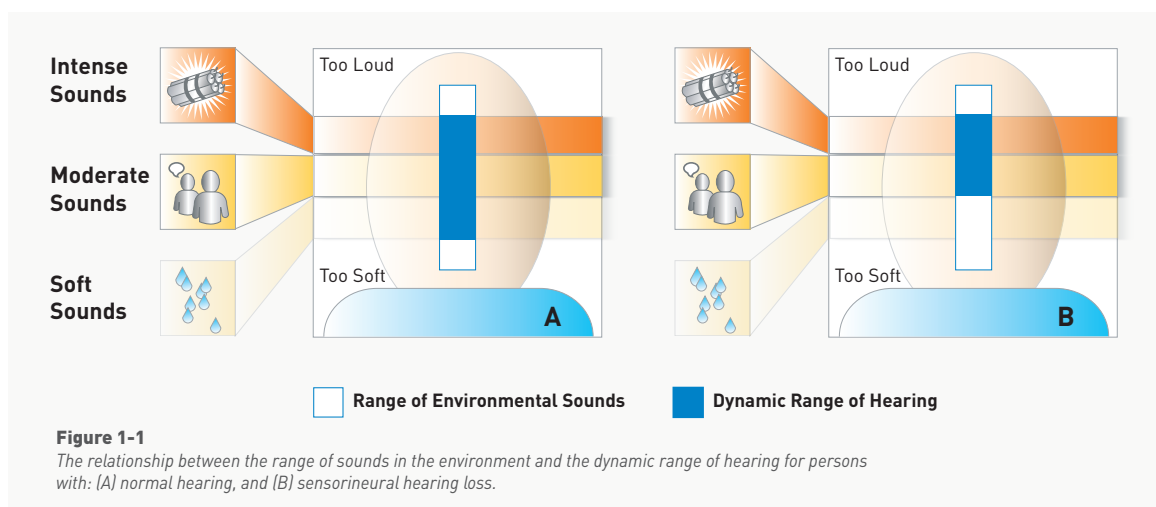
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Chapter 1: Understanding Hearing Loss

According to the World Health Organization (WHO), there are 360 million persons in the world with disabling hearing loss (WHO, 2012). 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 soft to extremely intense, that may occur in an individual's environment. The softest sounds that are audible lie at the threshold of hearing sensitivity. At the opposite end is the loudness discomfort level, 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, 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 will notice three things immediately. First, average conversational speech is now barely audible to the individual. Second, soft 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		Too Loud
Moderate Sounds			
Soft Sounds	Too Soft A		Too Soft 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 help. 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 soft 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.
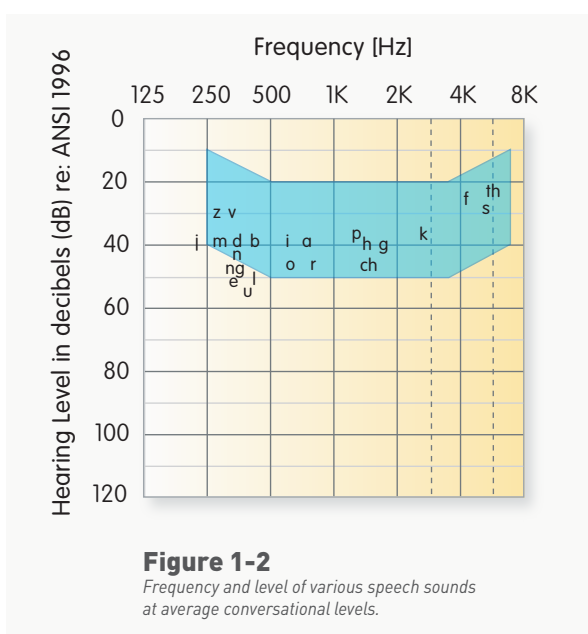
In light of these considerations, the hearing healthcare professional is faced with squeezing an elephant into a suitcase when fitting hearing aids – soft sounds must be made audible without loud sounds becoming uncomfortable, and speech should remain at a comfortable level. Compression amplification is a means for fitting the world of sound (the elephant) into the narrow 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, as well as 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.

ANSI
American National Standards Institute has created ANSI 3.22. The ANSI S3.221 is the hearing aid standard that has been the defining document for hearing instrument performance parameters in the United States since 1977. It defines the terminology around hearing aids and how manufacturers test the hearing aids.

Input
Input refers to the acoustic signal entering the hearing aid. Specifically, the American National Standards Institute (ANSI, 2014) 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
Basic components of a hearing aid: microphone, amplifier, receiver and battery.
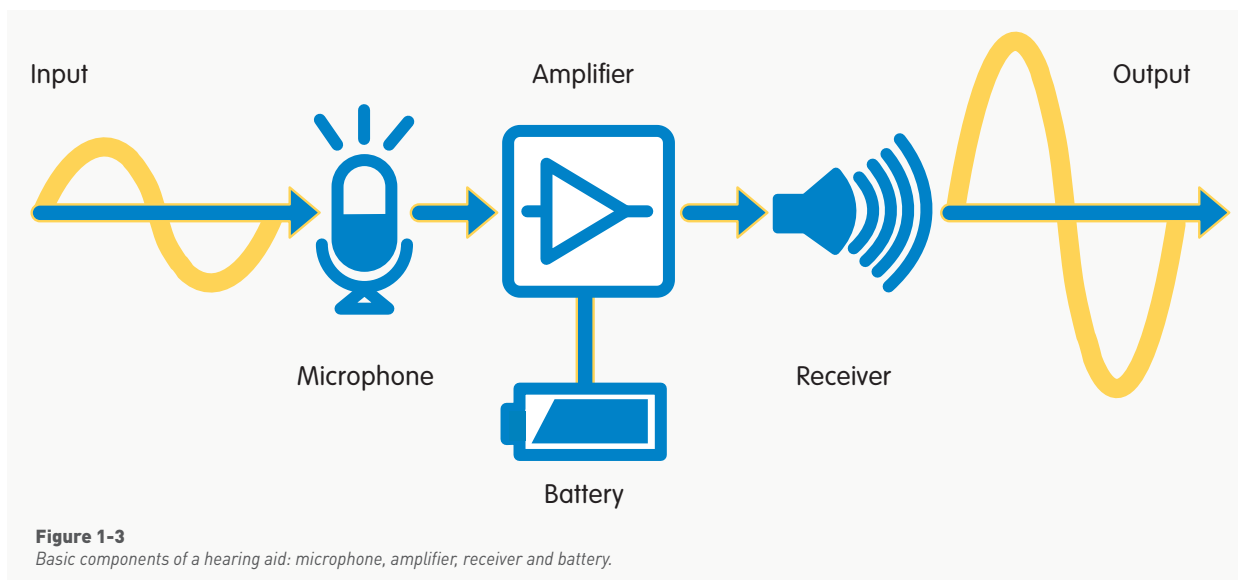
Output

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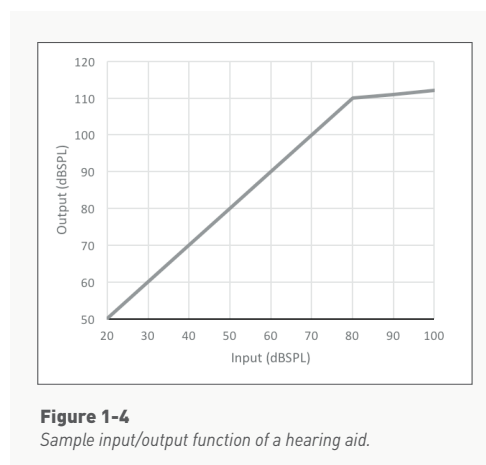
Input/Output Function

An input/output (I/O) function is a graphical representation of the output of a hearing aid at various input levels. According to ANSI (2014) the I/O graph has the output SPL on the ordinate (Y axis) with the input on the abscissa (X axis). The scales for both axis should be linear and of equal spacing. I/O curves are run at individual frequencies. 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 (2014) 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:

$$Gain = Output - Input$$

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

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
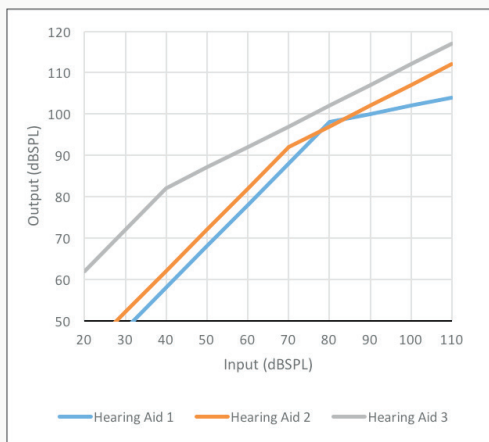*Sample input/output functions for three different hearing aids.*

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

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:

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

Figure 1-7

Figure 1-7
Sample input/gain functions for three different hearing aids.

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 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.
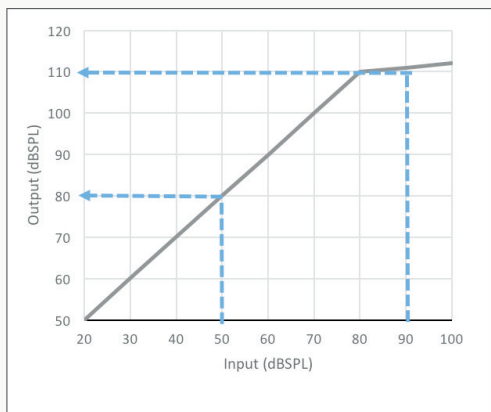
Figure 1-8A
Sample input/output function for a hearing aid.
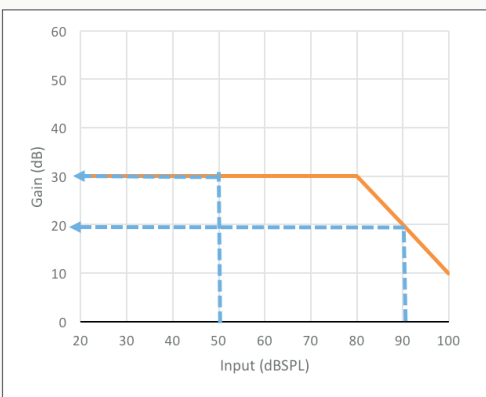
Figure 1-8B
Sample input/gain (B) function for a hearing aid.

Frequency Response Curve

A **frequency response curve** is a graphical representation of the hearing aid output as a function of frequency. 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.

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

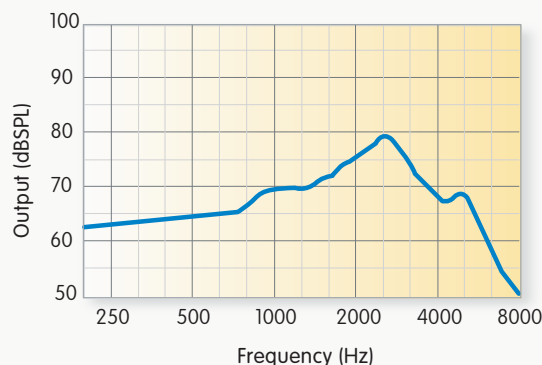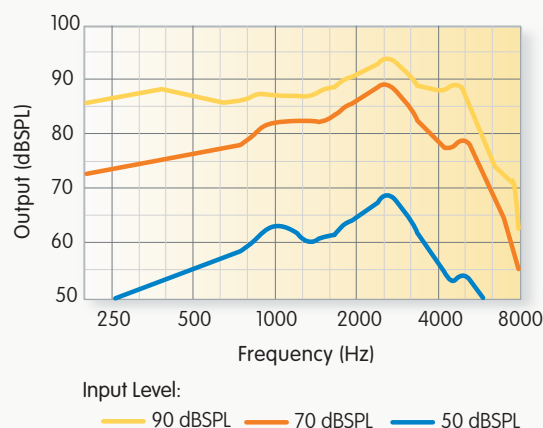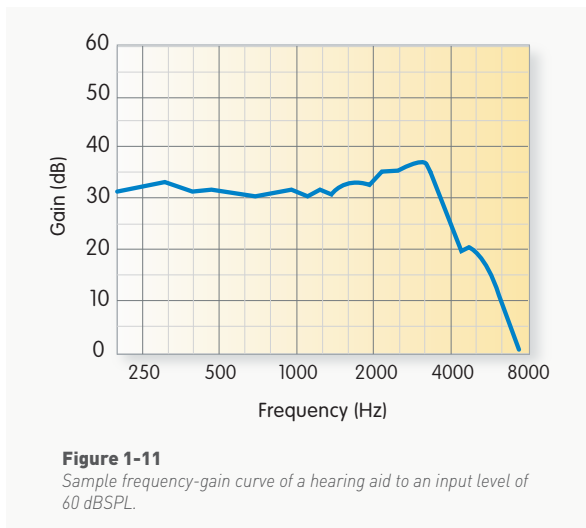In general, the output of a hearing aid increases as the input level increases. However, once the output reaches a certain level, the hearing aid is incapable of producing a louder signal. **Maximum Peak Output** (MPO) is the highest possible signal that a hearing aid is capable of delivering, regardless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

As long as the output of the hearing aid remains below the MPO, 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 MPO 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 occurs when a single frequency is presented to the input of a hearing aid and the output contains the original frequency plus additional undesired frequencies that are harmonically related to the original frequency. 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 occurs when two frequencies are presented simultaneously to a hearing aid and the output contains one or more frequencies that are related to the sum and difference of the two input frequencies. 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.

A Input Ouput

Maximum Output of Device

B Input Ouput

Figure 1-12
Schematic of the response of a hearing aid: (A) without peak clipping, and (B) with peak clipping.
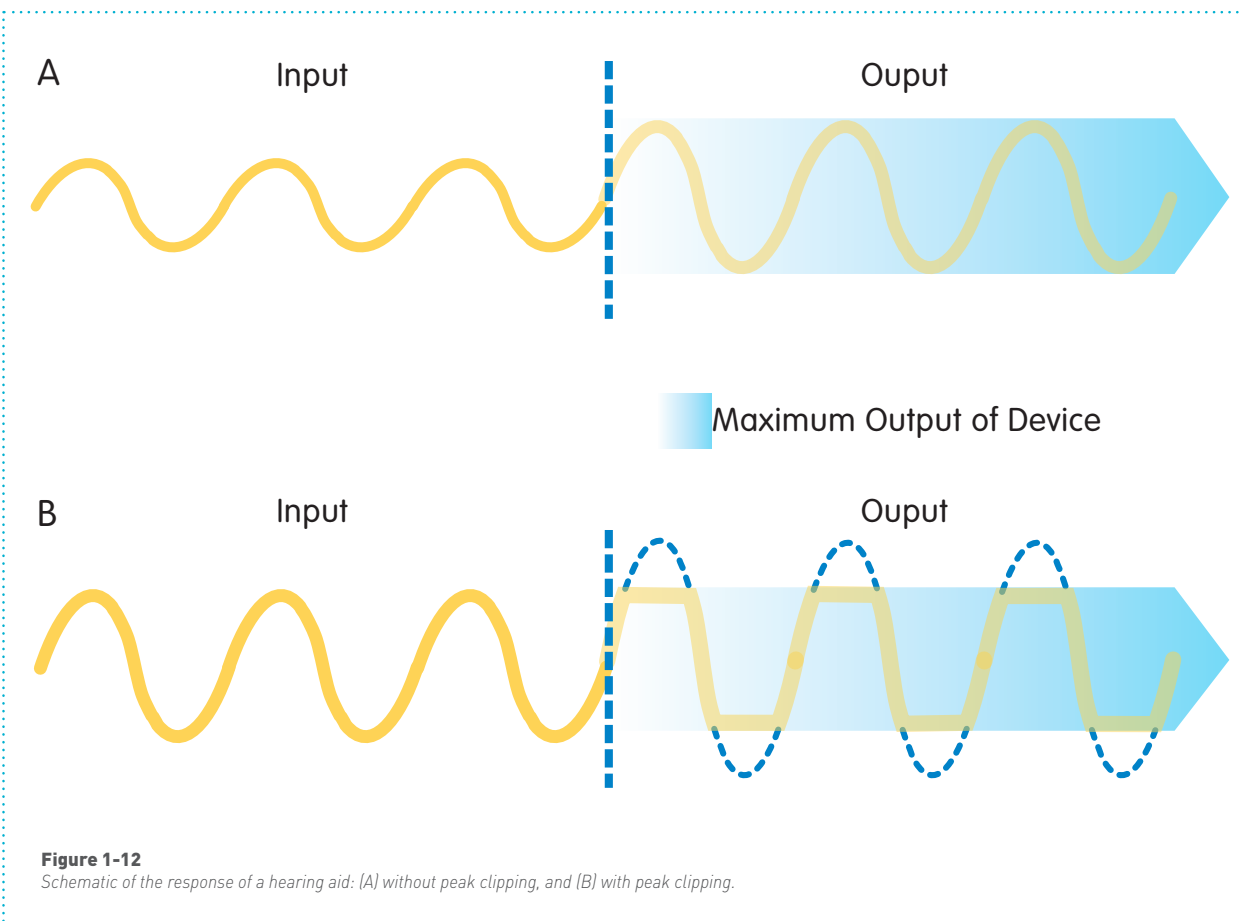
Chapter 2: Basics of Compression

As discussed in Chapter 1, the intact auditory system has an exquisite ability to hear a wide range of sounds in the environment, from soft 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

known as the **residual dynamic range**. That is, the individual is unable to hear soft sounds, average conversational speech is barely audible, and intense sounds are heard as loudly as the normal-hearing ear (Figure 2-1B).

Range of Environmental Sounds   Dynamic Range of Hearing

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.
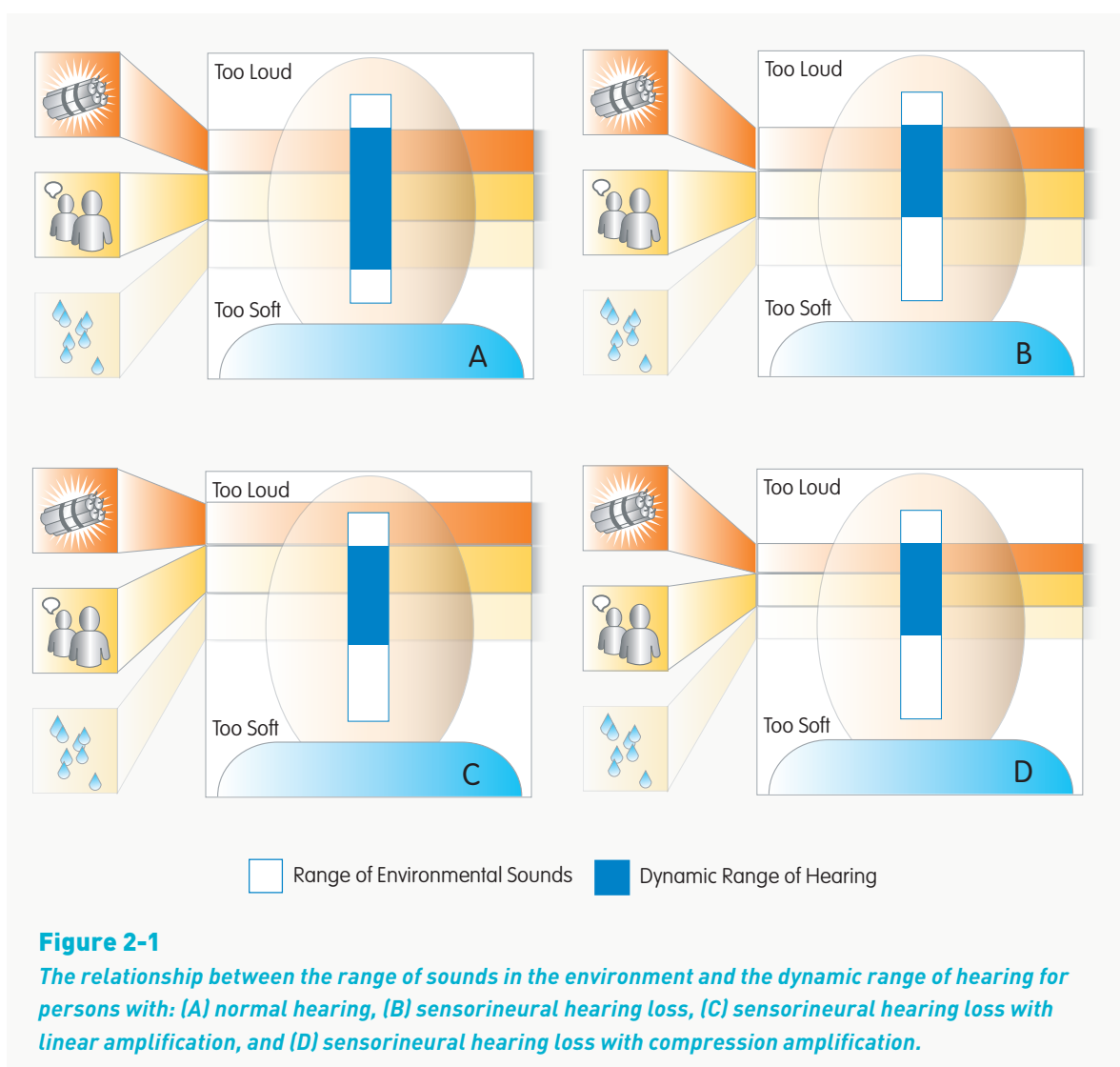
Because the most common complaint associated with hearing impairment is an inability to hear, it is tempting to alleviate the problem simply by making sounds louder. Figure 2-1C shows the sensorineural hearing loss to which the same amount of gain is applied to all sounds – i.e., amplification is **linear**. This makes soft 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 soft, 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. Soft 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.
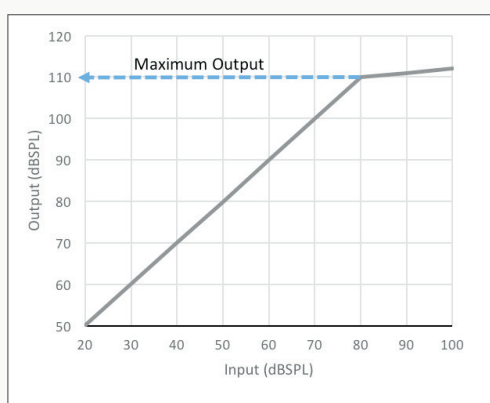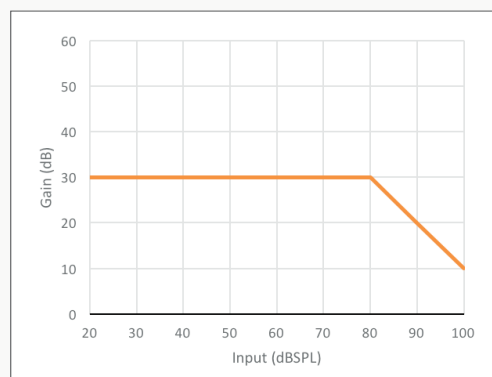

Figure 2-3
Input/gain function of a linear hearing aid with 30dB gain and a maximum output of 110 dBSPL.

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 MPO 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 the same 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 soft, moderate and intense sounds.

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

Characteristics of a Compressor

The function and application of compression circuits are defined by their features, such as compression threshold, compression ratio, attack time, and release time.

Compression Threshold

Compression threshold (CT) is defined as 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 40 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 knee point** (TK). These two terms are used interchangeably.

Figure 2-4
Sample input/output function for a hearing aid showing the compression threshold (CT) and the threshold knee point (TK).
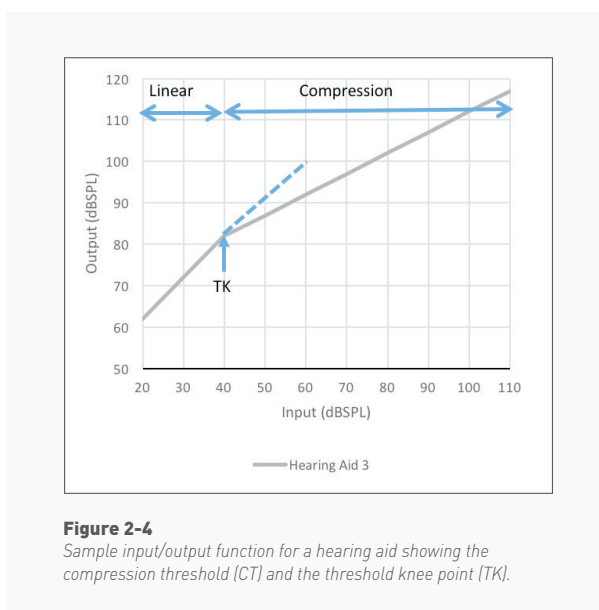
Depending on the purpose, a compression system may have high or low TKs. A high TK, 85 dBSPL or greater, is used to limit the output of a hearing aid so that it does not exceed the individual's loudness discomfort levels and to maximize listening comfort (for example, Hearing Aids 1 and 2, respectively, in Figure 2-5). On the other hand, a low TK, typically set below 50 dBSPL, may be used to improve audibility of the softer components of speech and/
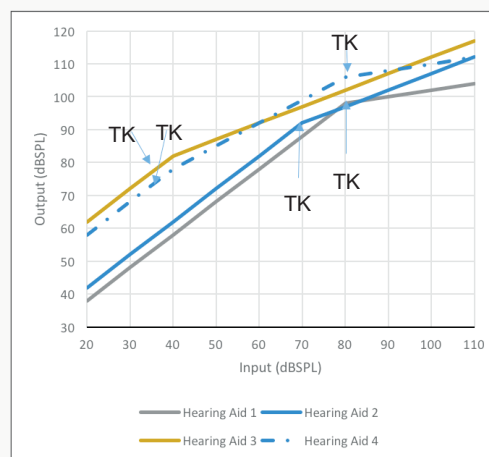
Figure 2-5
Sample input/output functions showing threshold knee points (TK) for four different hearing aids.

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 incremental change in input to the resulting incremental change in output (ANSI, 2014). 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. Although necessary for some applications, high CRs are known to adversely affect the clarity and pleasantness of the amplified sound (Neuman, Bakke, Mackersie, Hellman, & Levitt, 1998).
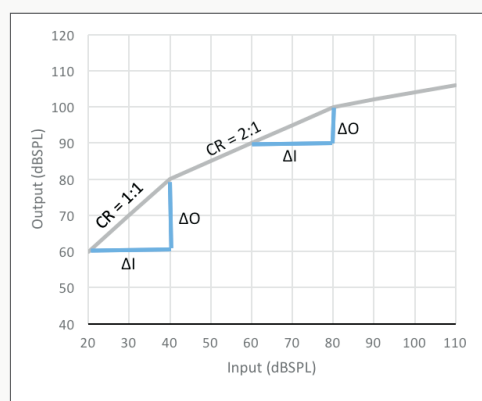

Figure 2-6
Calculating compression ratios (CRs) from an input/output function. ΔI = Change in input, ΔO = Change in output.
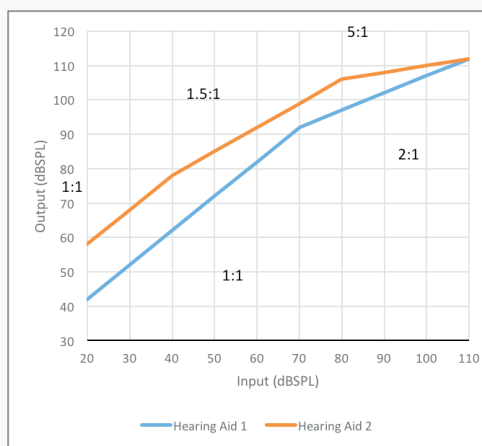
Figure 2-7
Sample input/output functions showing compression ratios (CR) for four different hearing aids.

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 (2014) 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.

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 knee point.
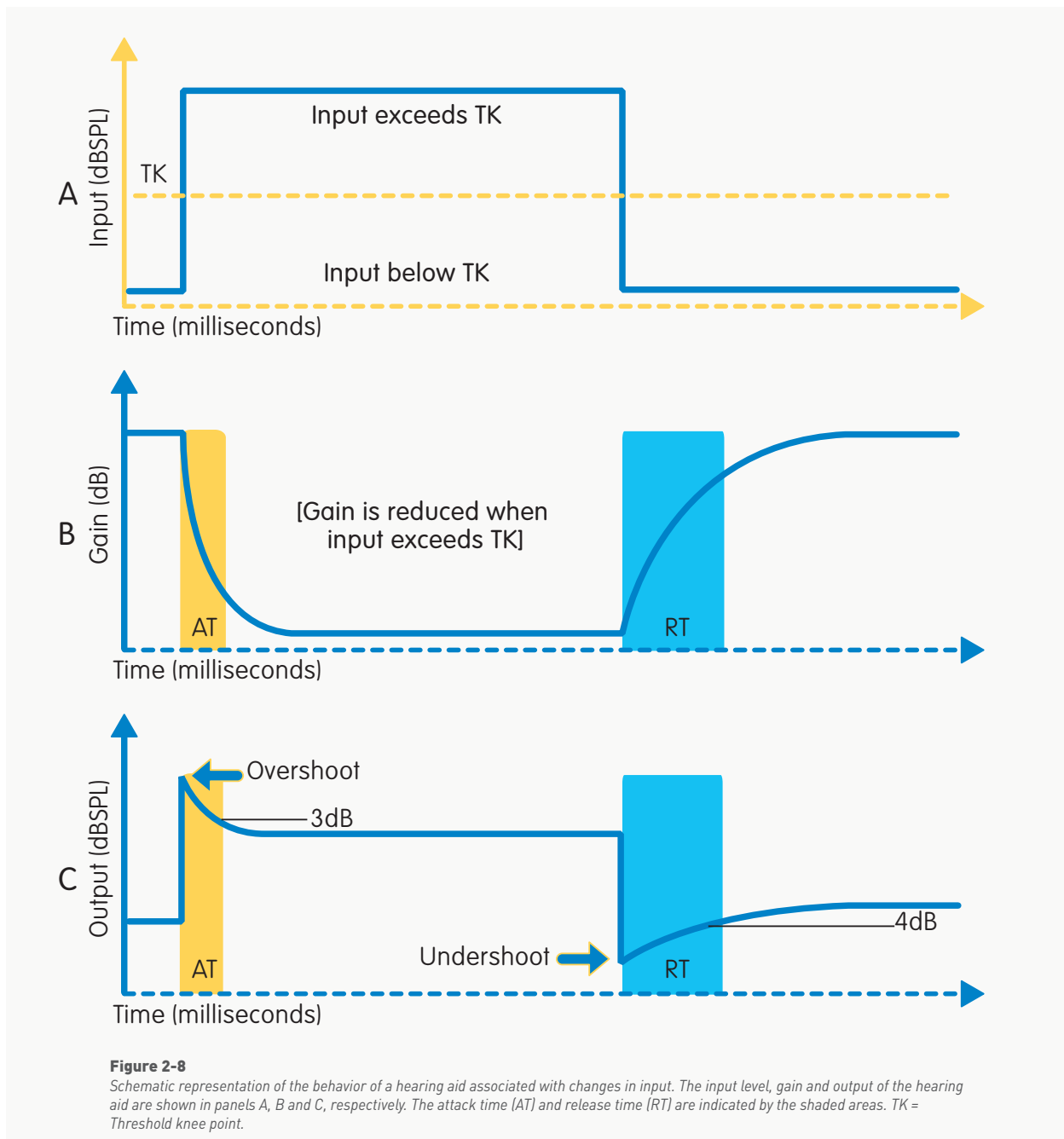
In Figure 2-8, when the input increases to a level above the TK (Figure 2-8A), the gain of the hearing aid does not change immediately (Figure 2-8B). The result is an overshoot in the output (Figure 2-8C). As the gain approaches its target, so too does the output of the hearing aid – i.e., reaches within 3dB of its final value.

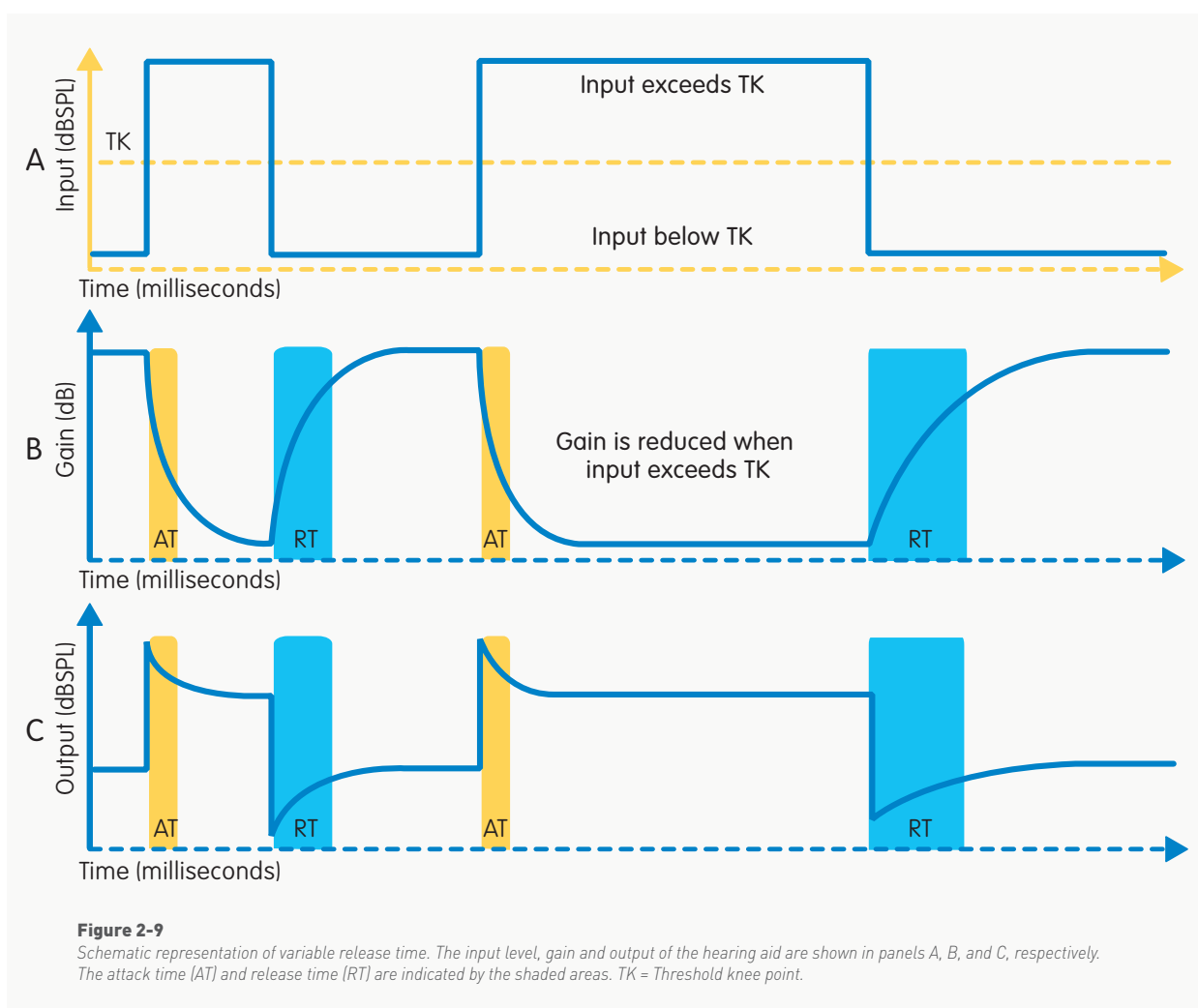**Release** (or recovery) **time** (RT) is the time delay that occurs between the offset of an input signal sufficiently loud to activate compression (i.e., input signal falls below the TK) and the resulting increase of gain to its target value. Specifically, ANSI (2014) defines RT as the interval between 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. 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 knee point (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 knee point.

Although it is convenient to discuss the 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 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. That 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 knee point. However, neither the TK nor the CR changes as a result.
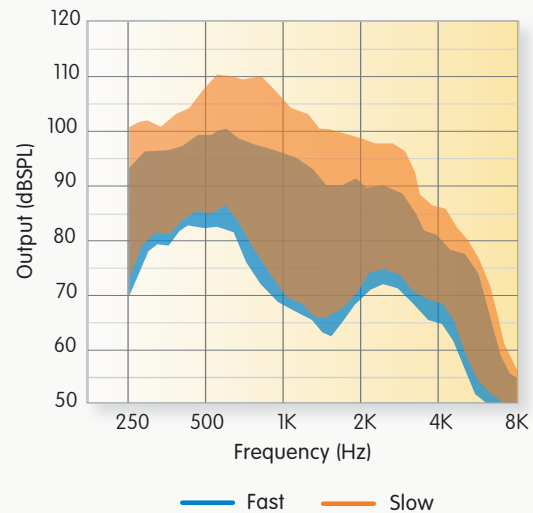

Figure 2-10
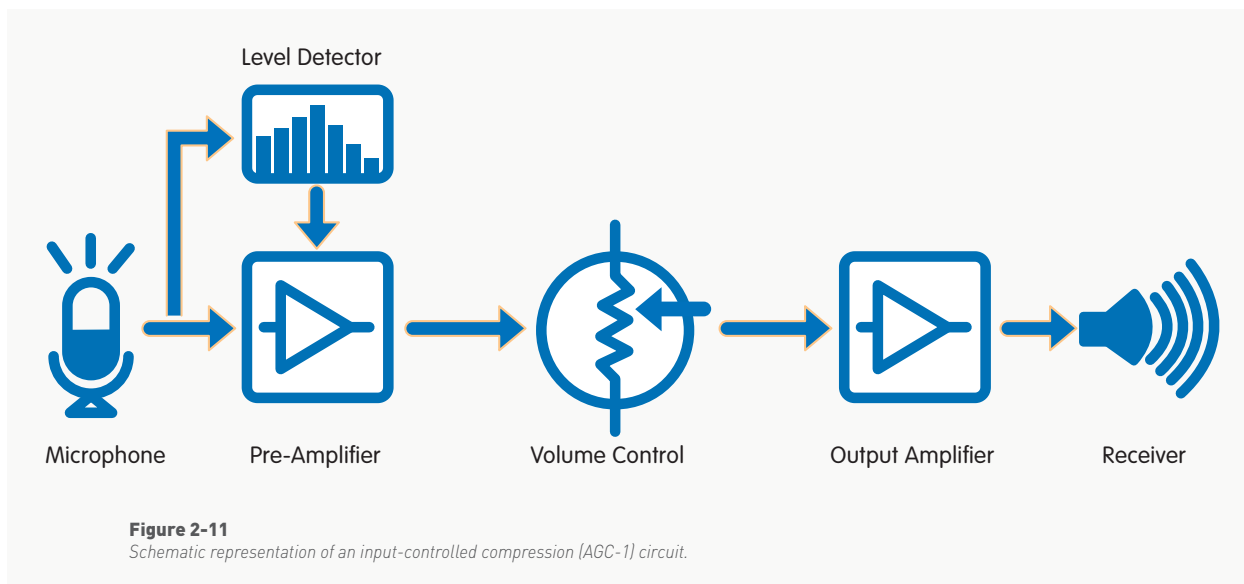*Effects of fast and slow attack and release times on the response of a hearing aid to a speech signal.*

Figure 2-11
Schematic representation of an input-controlled compression (AGC-1) circuit.
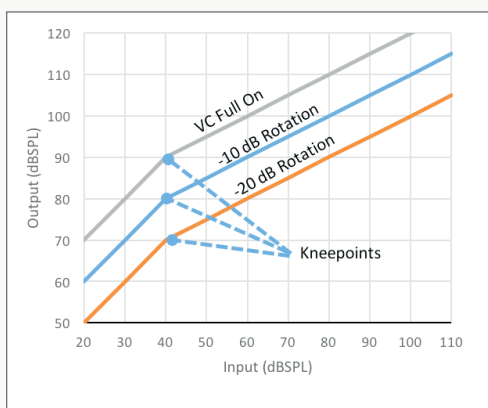
Figure 2-12
Sample input/output functions for an input-controlled compression (AGC-1) circuit showing the effects of volume control rotation.

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 soft, moderate and intense sounds at frequencies within that band. The compression parameters are unaffected. In contrast, **compression channels** allow separate adjustments for soft and intense input levels; the effect on moderate level inputs depends on the compression architecture.
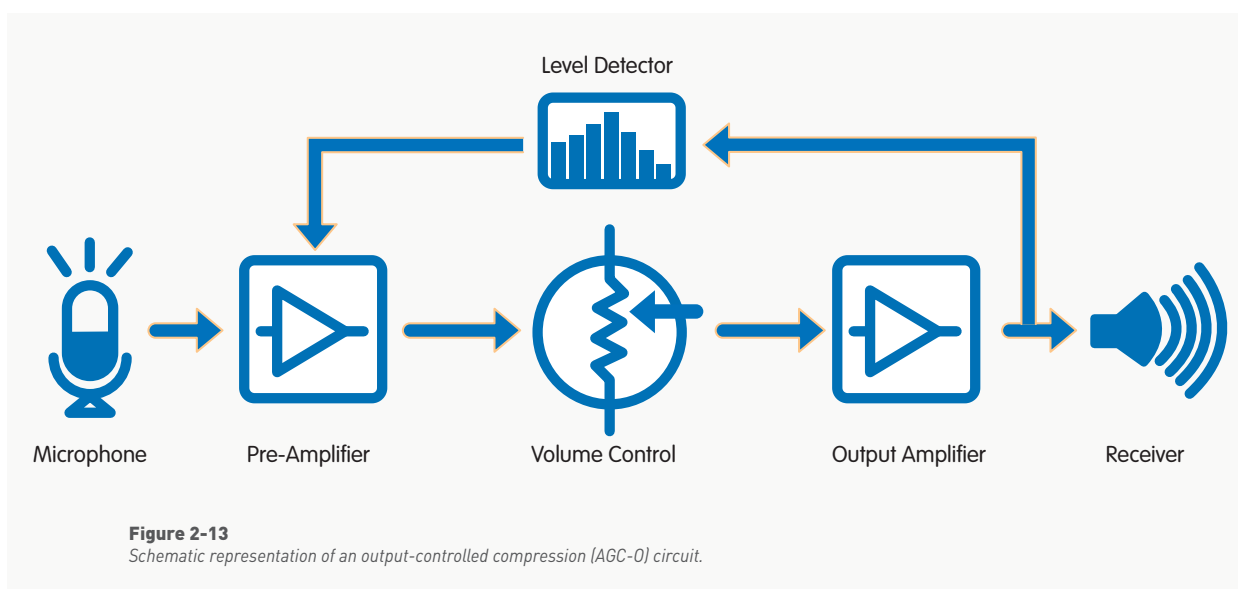

Figure 2-13
Schematic representation of an output-controlled compression (AGC-O) circuit.

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.
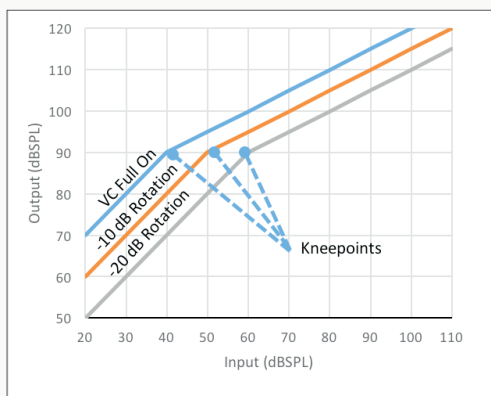

Figure 2-14
Sample input/output functions for an output-controlled compression (AGC-O) circuit showing the effect of volume control rotation.
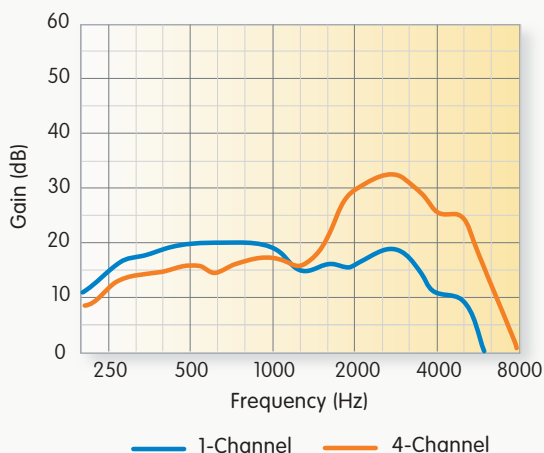
A hearing aid may have the flexibility to allow adjustments in multiple frequency bands and/or compression channels. Multiple channels are separated by **crossover frequencies**, which may or may not be adjustable. The ability to change gain individually in bands and/or channels is useful for two reasons. 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.

Increasing the number of hearing aid channels will increase the ability to restore frequency-specific audibility, particularly for steeply sloping audiograms. Using predictive metrics, researchers have completed systematic investigations of this relationship. It appears that frequency-specific audibility can be restored for flat or mildly sloping hearing loss with a three channel hearing aid; for patients with a steeply sloping hearing loss, the ideal number of channels required for restoration of audibility may fall between 9 and 18 channels (Galster & Galster, 2011).

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.

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 (soft 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.
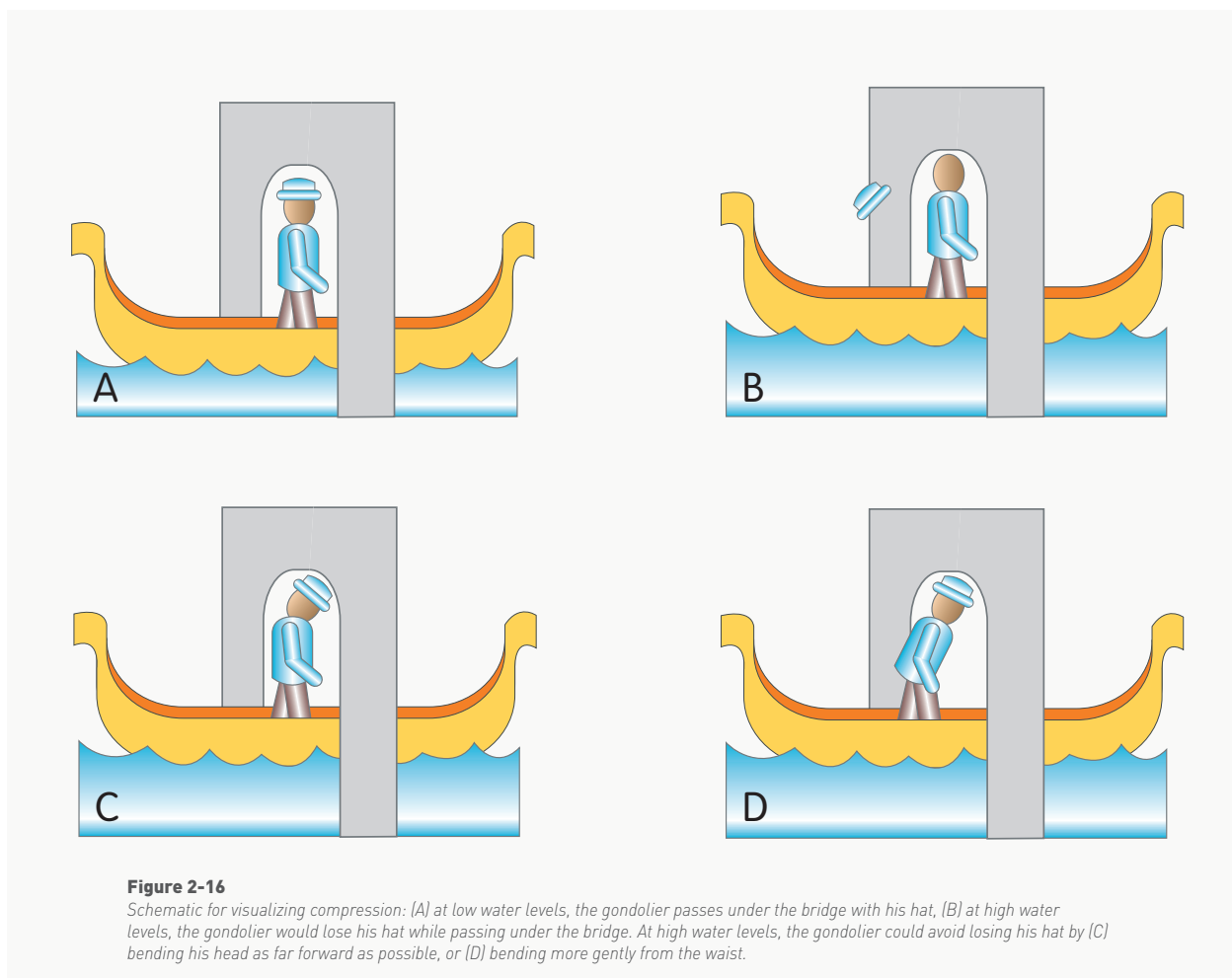
Chapter 3: Applications of Compression

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 knee point, 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.
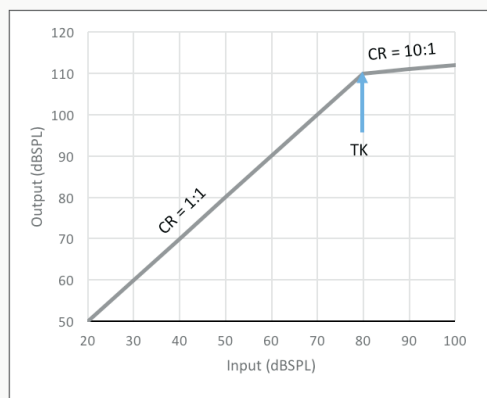
Figure 3-1
Sample input/output function for a hearing aid with output compression limiting.

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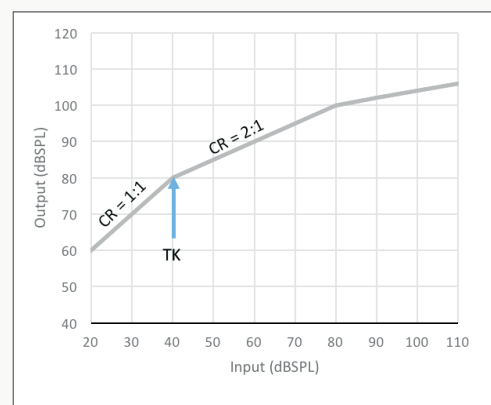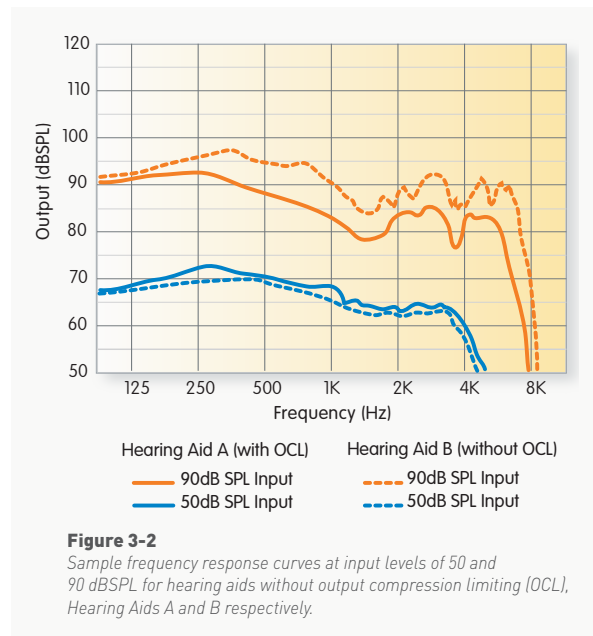
- AGC-I is used because the goal is to achieve a certain degree of audibility and/or loudness for incoming signals. If average, rather than individual, data are used to derive the amount of amplification necessary, the option may exist to increase or decrease the output of the hearing aid by adjusting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

- The TK is as low as possible in order to make soft 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 range 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 1) is visibly different from that of a linear hearing aid with OCL (Hearing Aid 2). In addition to the TK and CR being lower, Hearing Aid 1 also provides more gain than Hearing Aid 2 at and below the TK. This additional gain results in improved audibility of soft sounds.
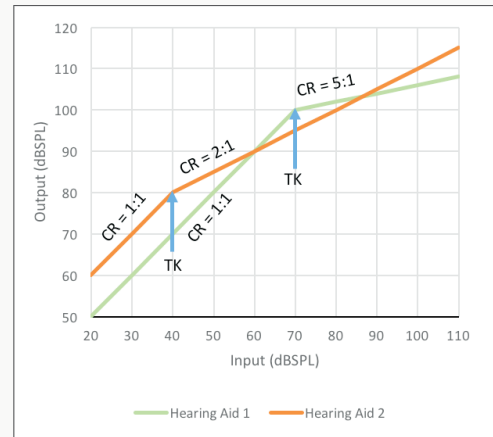
Figure 3-4
Sample input/output functions for a hearing aid with wide dynamic range compression (Hearing Aid A) and a linear hearing aid with output compression limiting (Hearing Aid B).
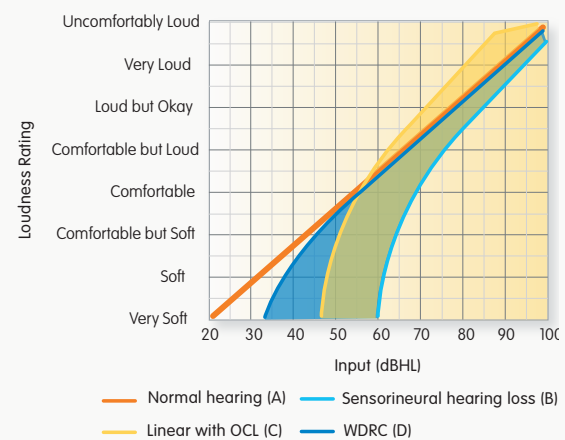
Figure 3-5
Sample loudness growth curves for persons with: (A) Normal hearing, (B) Sensorineural hearing loss, (C) Sensorineural hearing loss with linear amplification and output compression limiting (OCL), and (D) Sensorineural hearing loss with wide dynamic range compression (WDRC).

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 soft 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, soft 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 soft 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 1), WDRC (Hearing Aid 2) and linear amplification with OCL (Hearing Aid 3). Note that below the knee point, 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.
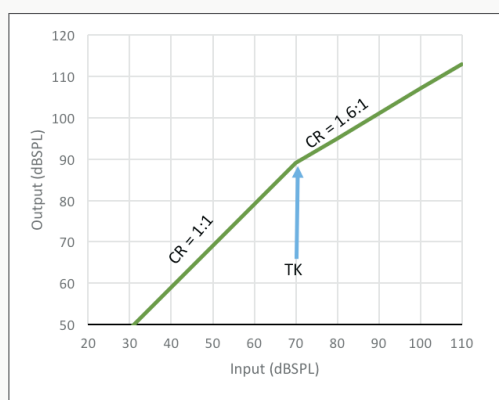

Figure 3-6
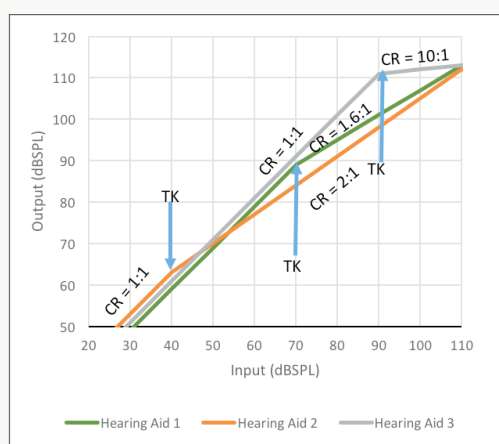*Sample input/output function for a hearing aid with mid-level compression.*

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.

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 softer 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 softer components than for the more intense components of speech.

- Multichannel compression is used so that soft 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 3-8. 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, soft 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. Speech has a dynamic range of 30dB. As a result, there may be components that are even softer with linear amplification and OCL than shown in the figure.

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 3-9 summarizes the compression characteristics associated with each application.

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 3-8
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.

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 3-9
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 perceived 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 soft, 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 soft 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, Dirks, & Mickey, 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.
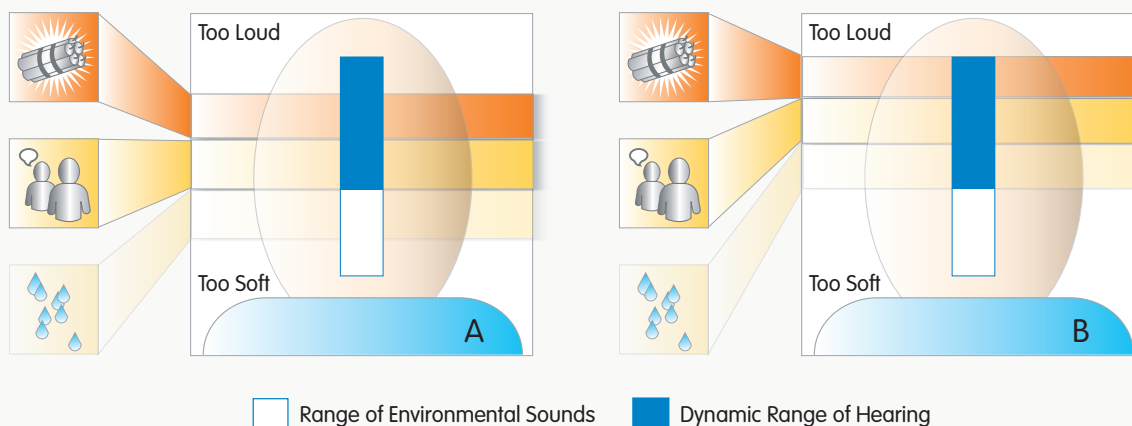
Figure 4-1
The relationship between the range of sounds in the environment and the dynamic range of hearing for persons with:
(A) conductive hearing loss, and (B) conductive hearing loss with linear amplification.

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.

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 soft sounds than for intense sounds; the amount of amplification required for moderate sounds lies in between these extremes. The goal is for soft 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, Noble, & LePage, 1992), avoidance of auditory deprivation (Silman, Gelfand, & Silverman, 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, Prosek, Walden, & Montgomery, 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 7 years before seeking help (BHI, 2016). 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 old 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 over amplification 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 formulas 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 two most commonly-used non-linear prescriptive approaches will be discussed – NAL-NL2 and DSL [i/o] v5.

NAL-NL2

NAL-NL2 was developed by the National Acoustics Laboratory in Australia. It is a 2nd generation nonlinear fitting formula. The goal of NAL-NL2 is to make speech intelligible and overall loudness comfortable. NAL-NL2 uses a speech intelligibility model combined with a loudness model. This formula would be classified as a loudness equalization formula.

When the second generation was released, adjustments were made based upon empirical data from the original NAL-NL1 fitting. In comparison to NAL-NL1, NAL-NL2 prescribes a different gain-frequency response shape, and slightly higher compression ratios are prescribed for those with mild or moderate hearing loss.

DSL i/o v5

The Desired Sensation Level fitting formula was developed out of the University of Western Ontario. It was originally developed primarily for children, but the more recent versions can be used for adults as well. The underlying rationale for the DSL procedure has been to normalize loudness at each frequency.

The most recent version takes into consideration the age of the patient, the type of hearing loss and if the fitting is monaural or binaural. The non-linear aspect of the formula takes channel structure into consideration, including the number of channels, the variability of CT (allowing for different thresholds for different frequencies) and CR across all channels. DSI i/o v5 also limits the peaks of speech to protect from over-amplification and control for comfort.

Proprietary

Proprietary is another type of fitting formula. NAL-NL2 and DLS i/o v5 could be considered "generic." They can work with any hearing aid on any patient. However, hearing aid manufacturers will have a "Proprietary" formula that is unique to them and designed for their specific products. eSTAT is the Starkey proprietary formula. It has been designed for our channel structure and based upon our philosophy of optimizing audibility whenever possible.

Chapter 5: Expansion

Issues Resulting from WDRC

As we have seen throughout this handbook, WDRC provides additional gain for soft 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 softer sounds in the environment may not have been heard for some time. WDRC enables amplification of soft 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 soft 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 soft 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 – soft 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 soft environmental sounds and circuit noise, without sacrificing speech audibility and intelligibility.
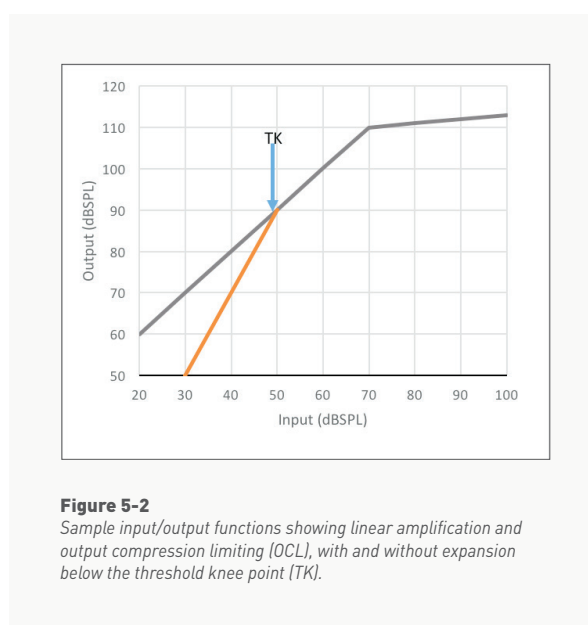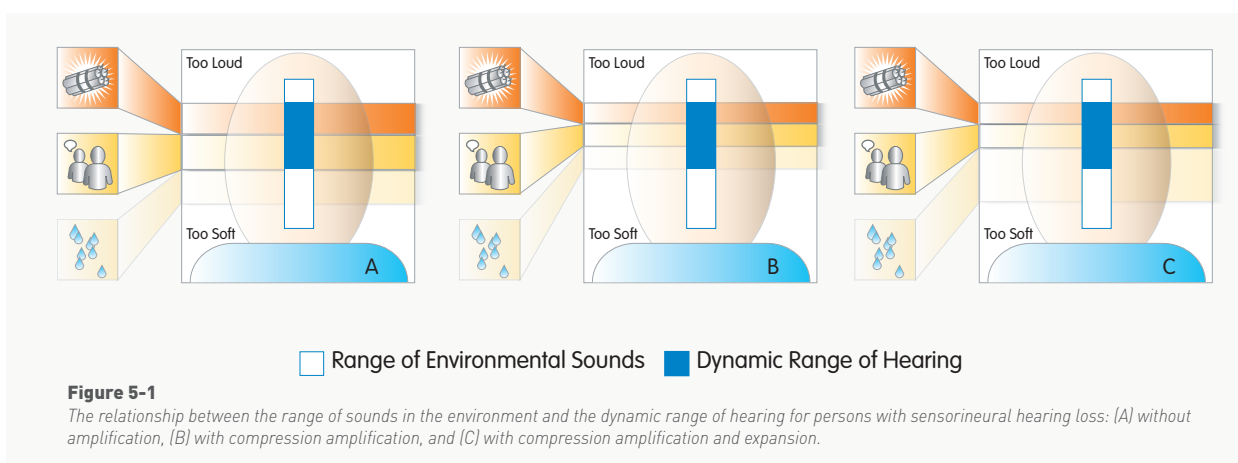
What is Expansion?

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, soft sounds are inaudible, while intense sounds are perceived as loud. Figure 5-1B illustrates the effect of compression amplification – soft 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 soft sounds is wider than those for moderate or intense sounds. Further, the softest of the soft sounds are below the dynamic range, and only the more intense components of the soft 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 softest 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 soft 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.

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 knee point (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 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 knee point. 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.
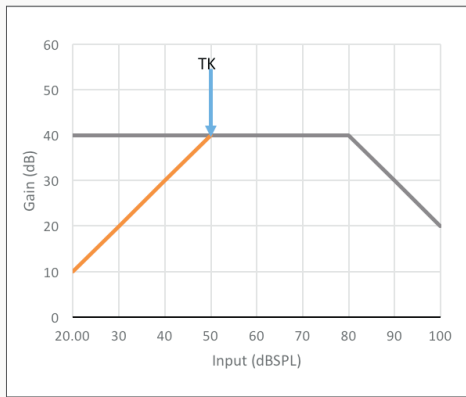
Figure 5-3
Sample input/gain functions showing linear amplification and output compression limiting (OCL), with and without expansion below the threshold knee point (TK).
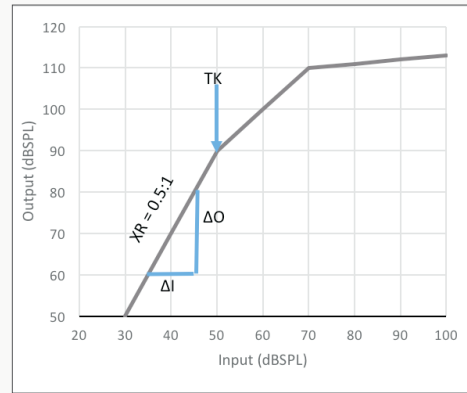

Figure 5-4
Identifying the threshold knee point (TK) for expansion and calculating the expansion ratio (XR) from an input/output function.

Characterizing Expansion

Expansion is characterized in much the same way as compression, via a threshold knee point, expansion ratio, and attack and release times. There are, however, a few key differences which are highlighted below.

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 (page 34), 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 35 to 45 dBSPL (ΔInput = 10dB) increased the output from 60 to 80 dBSPL (ΔOutput = 20dB). Using the above formula, the XR of the hearing aid is:

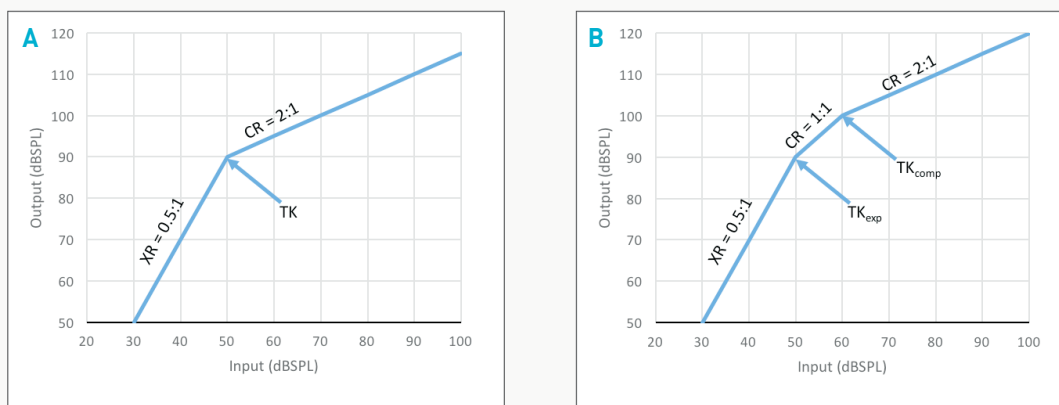$$XR = \frac{\Delta Input}{\Delta Output} = \frac{10}{20} = 0.5$$

Figure 5-5
Sample input/output functions for hearing aids with: (A) the same threshold knee point (TK) for compression and expansion, and (B) a threshold knee point for compression (TK_{comp}) higher than the threshold knee point 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 1, with an XR of 1, is linear below the TK. The I/O function for Hearing Aid 3 (XR=0.5) has a steeper slope than that for Hearing Aid 2 (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 soft sounds will be amplified to within the residual dynamic range. In other words, the likelihood that a person will be able to hear the softest 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, Hill, & Trine, 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.
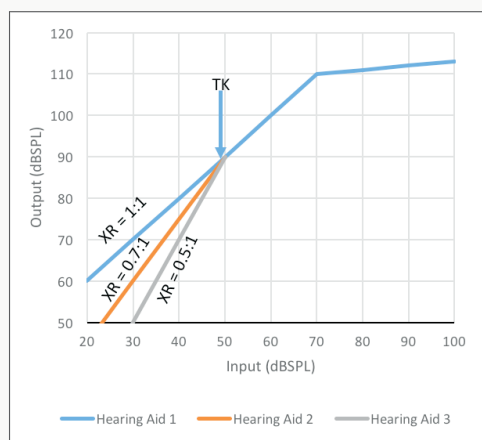

Figure 5-6
Sample input/output functions showing three hearing aids with the same threshold knee point (TKexp) and different expansion ratios (XR).

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, Hill, & Trine, 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 soft environmental sounds inaudible with little effect on the audibility of speech.
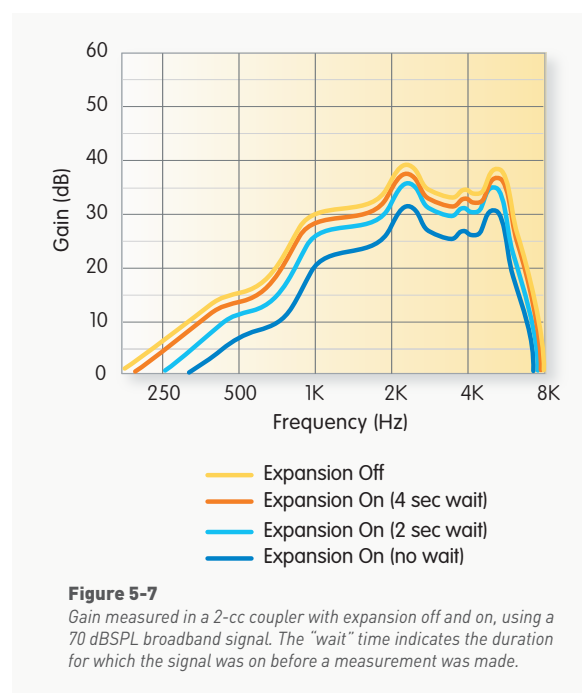
Measuring Expansion

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.

Chapter 6: Troubleshooting & Fine-Tuning

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 concerns, (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 and colleagues (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.

Figure 6-1
Acoustic characteristics of some everyday sounds.
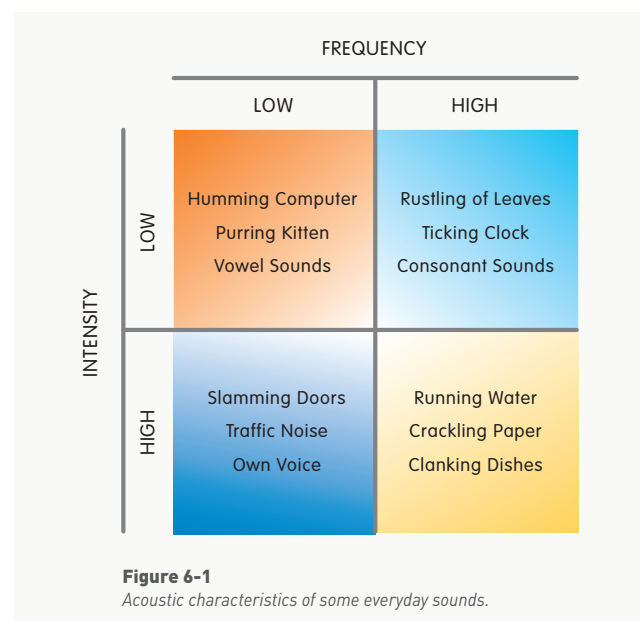
Determining the direction of change needed is the next step in addressing the problem. The good news is that, once the problem has been clarified, professionals generally agree on the course of action. Frequently, this is the easiest part of the process because the direction of change is immediately 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

Historically

In older technology and early digital products, while there were often independent CT and CR controls, there may have been only one CT and CR that was applied over the wide range of input levels and/or frequencies. Typically, if we could not adjust gain for an input level without adjusting other input levels, we would manipulate the threshold and ratio to create the desired effect. Often, the result was a compromise that got us as close as possible to the desired outcome. In the example mentioned above, we might lower the CT or increase the CR to minimize the gain for intense, low frequency inputs. This may provide the desired result for intense sounds, but may also impact the perception of moderate sounds. This is the compromise.

Current Technology

Modern digital technology allows us to offer multiple knee points at different input levels. Ultimately, it means, rather than thinking about adjusting the ratio or knee point, clinicians can focus on adjusting the gain for a specific input level while minimizing the impact on other input levels. Again, using the example from above, we could lower in the gain for intense, low frequency inputs without impacting how soft or moderate low frequency sounds are perceived.

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.

Patient Adjustment Guide

PRODUCT: Synergy Products: Muse, SoundLens Synergy, Halo 2

	Patient Report	Inspire Adjustment	Other Considerations

Patients Own Voice	**VOICE SOUNDS**
	> In a barrel/tunnel
	> Echoes
	> Hollow
	> Like they have a cold/ears plugged
		> Decrease Overall Gain using Occlusion Control
		> Decrease Low Frequency Gain
		> Decrease Moderate Gain at 1000Hz and/or 1500Hz
		> Acoustic Options dialog box will update based on connected hearing aid
			> Occlusion may be due to the physical presence of the hearing aid and not as a result of amplification
			> To test, turn off the hearing aid (open battery door) and have the patient speak
			  1. Report persists – issue is occlusion; address with acoustic modifications
			  2. Report resolved – issue is amplification; address with frequency adjustments
			> Enlarge Vent Diameter
			> Shorten and/or Taper Canal
			> Remake hearing aid with either deeper or shorter canal length; depending on current hearing aid
			> Inspire-selected Acoustic Options must match options connected to hearing aid

	VOICE SOUNDS
	> Muffled
		> Increase Moderate Gain at 1000Hz and/or 1500Hz
		> Increase Loud Gain
		> Increase Overall Output
		> Increase High Frequency Gain
		> Decrease Low Frequency Gain
		> Enlarge Vent Diameter
		> Shorten and/or Taper Canal
		> Acoustic Options dialog box will update based on the connected hearing aid
			> Compression Ratios are increaed as the curves move closer together; decreased as the curves move farther apart
			> Occlusion may be due to the physical presence of the hearing aid and not as a result of amplification
			> To test, turn off the hearing aid (open battery door) and have the patient speak
			  1. Report persists – issue is occlusion; address with acoustic modifications
			  2. Report resolved – issue is amplification; address with frequency adjustments
			> Enlarge Vent Diameter
			> Shorten and/or Taper Canal
			> Inspire-selected Acoustic Options must match options connected to hearing aid

	VOICE SOUNDS > Distorted > Crackles > Unnatural/like a megaphone	> Decrease Moderate Gain at 1000Hz and/or 1500Hz > Decrease Loud Gain > Decrease Overall Output	> Compression Ratios are increased as the curves move closer together; decreased as the curves move farther apart > If decreasing Overall Output worsens sound quality, consider increasing Overall Output
Hearing in Noise	**PATIENT HAS DIFFICULTY** > Understanding speech in background noise	> Verify Adaptive Directionality is enabled > Increase Overall Gain at 1000Hz and/or 1500Hz, then higher Frequency Gain > Enable Directionality Plus via Environment Manager screen > Decrease Soft Low Frequency Gain > Increase Speech and Noise via Environment Manager screen	> Verify Adaptive Directionality is activated > If device does not have directional microphones, consider recommending a directional device > Directionality Plus adjusts the level of activation of Speech and Noise to engage at a lower input level > Consider SurfLink Mobile or SurfLink Remote set to iQ Boost for Wireless Devices
	PATIENT HEARS > Voices at a distance better than near	> Increase Loud Gain at 1000Hz and/or 1500Hz, then try Higher Frequency Loud Gain > Decrease Soft Gain > Increase Overall Output	> Compression Ratios are increased as the curves move closer together; decreased as the curves move farther apart > Consider SurfLink Mobile Remote Microphone set to Focus or SurfLink Remote Microphone for Wireless Devices
	PATIENT REPORTS > Low tolerance for noise > Background noise too loud	> Decrease Overall Output > Enable Directionality Plus via Noise Control screen > Set Speech and Noise for More activity via Environment Manager screen	> Verify Adaptive Directionality is activated > If device does not have directional microphones, consider recommending directional device > Directionality Plus adjusts the level of activation of Speech and Noise to engage at a lower input level > Consider SurfLink Mobile or SurfLink Remote set to iQ Boost for Wireless Devices
Music	**MUSIC SOUNDS** > Unable to hear background music > Background music is too dominant > Unexexpected fluctuations in background music > Music too tinny in the Music Memory > Music has too much bass in the Music Memory	> Set Music adaptation to a higher scrength via Environment Manager screen > Set Music adaptation to a lower strength via Environment Manager screen > Set Music adaptation to a lower strength via Environment Manager screen > Decrease treble via QuickFit screen > Increase bass via QuickFit screen > Decrease bass via QuickFit screen > Increase treble via QuickFit screen	Consider creating a dedicated Music Memory for improved music sound quality Consider the Fine Tuning screen for musicians who require very discrete frequency specific adjustments of loud and soft music

Patient Adjustment Guide

PRODUCT: Synergy Products: Muse, SoundLens Synergy, Halo 2

	Patient Report	Inspire Adjustment	Other Considerations
Loudness	**OVERALL TOO LOUD** > Voices too loud > All sounds too Loud > Harsh/too loud	> Engage Experience Manager, selecting a lower level starting point > Change Experience Level to provide less gain (3 to 2 or 2 to 1) > Decrease Overall Gain > Decrease Gain using Occlusion Control > Decrease High Frequency Loud Gain	> Patient may not be accustomed to amplification or may be accustomed to lower gain devices > May need to start with lower gain settings than the prescriptive target recommends > May need to consider a different fitting formula > Compression Ratios are increased as the curves move closer together; decreased as the curves move farther apart
	LOUDNESS COMFORT > Sounds are painful > Clattering dishes too loud > Running water > Other environmental sounds too loud	> Decrease High Frequency Loud Gain > Decrease Overall Output > Decrease Overall Loud Gain > Set Machine Noise for More Activity via Environment Manager screen	> Ensure Best Fit is using e-STAT Fitting Formula > Compression Ratios are increased as the curves move closer together; decreased as the curves move farther apart > Utilize Speech Mapping to identify frequencies causing discomfort
	OVERALL TOO SOFT > Voices too soft > All sounds too soft > Hearing aids too soft	> Increase Overall Gain > Increase Overall Output > Engage MPO Boost > Increase Overall Soft Gain > Increase Overall Moderate Gain > Increase Low Frequency Overall Gain > Set Quiet for Less Activity via Environment Manager screen	> Patient may not perceive the aid as being loud enough depending on previous hearing aid experience > Compression Ratios are increased as the curves move closer together; decreased as the curves move farher apart > Utilize Speech Mapping to verify audibilty
Feedback	**HEARING AIDS** > Whistle > Chirp	> Initialize Whistlefree Feedback Cancellation with hearing aid seated in ear > View Maximum Stable Gain to check for areas of possible feedback > Use Auto Gain Adjust > Decrease Overall Soft Gain > Decrease Overall Moderate Gain > Decrease Overall Loud Gain > Decrease Overall Gain	> Whistlefree Feedback Cancellation needs to be re-initialized any time the acoustic characteristics of the hearing aid are changed (ex: shell modification, new earmold or earbud) > Utilize Speech Mapping to identify feedback peak and decrease gain at peak
Sound Quality	**NOISY** > Hearing aids are noisy > Refrigerator hum too loud > Heairng aids are noisy in quiet environments	> Set Quiet for More Activity via Environment Manager screen > Decrease Soft Gain at 750Hz and below > Decrease Overall Soft Gain	Quiet adjusts expansion to ensure the hearing aids are quiet in a quiet environment
	PUMPING > Hearing aids cut in and out > Hearing aids cut in and out when patient speaks > Loud sounds fade in and out	> Increase Overall Loud Gain > Set Machine Noise for Less Activity via Environment Manager screen	> Compression Ratios are increased as the curves move closer together; decreased as the curves move farther apart > Adjust Time Constants if available, (slower) for appropriate classification on Environment Manager screen
	SHUTTING DOWN > Hearing aids shut down with loud sounds > Hearing aids cut out when patient speaks > Loud sounds fade in and out	> Increase Overall Output > Increase Overall Soft Gain > Increase Overall Loud Gain > Engage MPO Boost	Compression Ratios are increased as the curves move closer together; decreased as the curves move farther apart
	SOUNDS ARE > Sharp > Tinny	> Increase gain between 2000 - 4000 Hz; then increase gain at 750 Hz > Increase Low Frequency Gain > Decrease Overall Output > Engage Experience Manager, selecting a lower level starting point > Change Experience Level to provide less gain (3 to 2 or 2 to 1)	> Patient's auditory perception may be distorted due to long standing high-frequency hearing loss; counseling is key > Consider Best Fit using a different fitting formula > Utilize Speech Mapping or Verify Comfort to identify areas of sharpness
	SOUNDS ARE > Hollow > Muffled	> Decrease Loud Gain at 500Hz and 750Hz > Increase Moderate Gain at 1000Hz and/or 1500Hz > Increase Moderate High Frequency Gain > Acoustic Options dialog box will update based on connected hearing aid	> Increase Vent Size and update Acoustic Options to match hearing aid > Inspire-selected Acoustic Options must match options connected to hearing aid
Intelligibility	**REPORTS** > I hear better without my hearing aids > Speech is unclear/unnatural > Speech in quiet is not clear > TV/Radio is not clear	> Increase Overall Gain at 1000Hz and/or 1500Hz > Set Speech and Noise for Less Activity via Environment Manager screen > Enable Television Memory via Memories Menu or Memories screen	> May need to counsel on fact that poor speech clarity may be due to poor speech discrimination > Consider SurfLink Mobile set to Start/Stop Audio Stream and/or SurfLink Media 2 for Wireless Devices

Recommended Reading

Dillon, H. (2000). *Hearing Aids.* Boomerang Press: Turramurra, Australia.

Mueller, H. G., Hawkins, D. B., & Northern, J. L. (1992).
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