Compression for Clinicians

A Compass for Hearing Aid Fittings

THIRD EDITION

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Preface

This book is intended for those studying to become hearing health care professionals, be they audiologists or hearing instrument practitioners; it's intended as well as for practicing clinicians who simply want some refreshment of their knowledge base concerning hearing loss and hearing aids. Readers looking for cutting-edge research will be disappointed. The book mainly summarizes knowledge that is already "out there." More than that though, it is my take on things, my own way of expressing and explaining developments that have occurred in the world of hearing aid compression, fitting methods, and real ear measurement.

Readers will likely notice in this edition a distinct lack of specific products, models and equipment names. The idea here was to keep the contents of this edition conceptual, and as timeless as possible. The few instances where specific names *are* mentioned, will be found only where historical reference is required.

It was amazing how much these things had changed between the time of the first edition (1998) and the second edition (2006). Now it is 2017. . . It was high time then, and it is high time now that this book is updated.

My own learning process in the world of hearing aids began after leaving academia, while working at Unitron from 1995 until 2001. What I had covered in the the first hearing aids course I ever taught at Auburn University from 1993 to 1995 had only snippets of compression (and those were mostly wrong)! I can truly say Unitron was my alma mater when it comes to hearing aids. The 1990s was a rather exciting time in the world of hearing aids, "Wide dynamic range compression" (WDRC) was developing and emerging as a new compression type. Multi-channel features were being added to programmability. This was all taking place in the world of analog hearing aids, where a hearing aid was either one type of compression or another type. Clinicians *had* to know their compression types. In a way then, the 1990s can be considered as the "golden age" of compression.

The second edition of this book (2006) was intended to be a bridge spanning the transition from analog to digital hearing aids. In it, the reader encountered many historical references. As digital hearing aids became the norm, the complexity of their features and associated fitting software has continued to increase dramatically. The golden age of compression (1990s) however, has long since passed, and the focus shifted elsewhere. The compression types and characteristics seem to be buried beneath the glossy surface of the fitting software. I sometimes tease about the psychosocial questions posited by the software, such as, "Does your client have trouble hearing the preacher from a 45° angle at a distance of 50 feet every second Sunday? If so, push this button."

This situation does not mean we no longer need to know our compression. All of the compression types utilized in yesterday's analog hearing aids—and much more—continue to be utilized in today's digital hearing aids. This then only highlights the fact that we must not lose our grip on the concepts surrounding compression. To truly appreciate and understand compression in today's digital hearing aids however, one must still consult the old definitions of compression as they were used in yesterday's analog hearing aids. To that end, this third edition continues to retain an historical perspective on compression.

What's new in this third edition? To begin with, my own knowledge base has continued to evolve (maybe not improved, but evolved nonetheless). Some things have remained the same; in the preface of the second edition, I urged clinicians to verify software fitting predictions with real ear measurement (REM). I still hold to that position. I knew that if I were ever to write a third edition, a new chapter on REM would be included. It has always been my strong contention that REM is inextricably intertwined with the development and evolution of fitting methods. To that end, the chapter on REM is situated precisely between the chapter that covers linear-based fitting methods and the chapter that covers compression-based fitting methods.

Two chapters from the second edition are gone. In this third edition, the topic of cochlear dead regions as Chapter 2 is now included as part of Chapter 3. As today's digital hearing aids almost all use multiple channels and programmability, Chapter 6 (Multi-Channel Programmable Hearing Aids) in the second edition has been folded into a section of the central chapter on compression (7) in this third edition.

Readers will see that there are a couple of themes that run like twin rivers throughout this third edition. One of these is the recognition of two distinct clinical populations of sensorineural hearing loss (SNHL): mild to moderate ("sensory"), and more severe ("neural"). These two clinical populations are well served by a corresponding pair of compression types—namely, WDRC and output limiting compression.

A second theme held throughout this book is the two-part task for all hearing aids—namely, (1) providing gain and (2) increasing the signal-to-noise ratio (SNR). Compression (Chapters 7 and 8) is a gain-related issue. Directional microphones (Chapter 9) and digital noise reduction (Chapters 8 and 9) both address the SNR issue.

The first and last chapters are new additions to this third edition. Chapter 1 covers the topic of Common Clinical Encounters, which has nothing to do with compression per se, but I hope it can make for some interesting reading. Many of these "encounters" do not seem to be deliberately laid out and explained elsewhere, and so the first chapter aims to do just that. The final chapter covers the topic of adaptive dynamic range optimization (ADRO). In the second edition, this topic was covered in the chapter on compression. Since that time, however, I have come to learn more about it. I feel strongly that linear gain *can* be a good thing; accordingly, I thought it might be a good idea to include this topic as a "postscript," as an "antidote" to the world of "compression as usual." Besides, many hearing aid manufacturers have been using linear gain as part of their compression schemes as well. I hope the readers unfamiliar with ADRO enjoy looking at things from this "other side of the fence."

targets of today's compression-based fitting methods are *output* targets, displayed in units of dB SPL. Target gain today is no longer in the center of the picture.

GAIN IN dB VERSUS OUTPUT IN dB SPL

As outlined in the previous chapter, the most important formula for understanding hearing aids and their function is: *Input + Gain = Output*. Input is the sound arriving at the microphone of the hearing aid, gain is the added amplification to the input, and output is the sum total arriving at the TM. *Input and output are always measured in units of dB SPL, while gain is always measured in units of dB*. Can we explain why this is the case? Dredging our turbulent memories, we all recall hazy shades of past agonies trying to absorb the decibel, one of these being the fact that "You cannot add decibels like 1 + 2 = 3!" Wait a minute though; we just did. Input + Gain = Output.

To find our way home, we must look at or define "absolute" versus "relative" decibel values. First, what do we mean when we say "0 dB SPL"? Contrary to what one might think, this does not represent the absence of sound. Ever test otoacoustic emissions? Check out the decibel values there. The noise floor in the ear canal is often −10 to −20 dB SPL! It thus behooves us to get a sane grip on the decibel (from hell).

Recall from Chapter 1 that 0 dB SPL simply represents the softest sound pressure for a normal-hearing person to hear a (1) 1000-Hz tone, (2) at a 1-meter distance from a speaker, (3) with both ears. All greater (and lesser) sound pressure levels are related to this "ground or defining level." If, for example, we ever want to say some sound is twice as intense as another, we must have a defining ground. Think of it this way: if we want to say an apartment building is twice as high as the house next to it, we need to know where the ground is, because that is the starting point for both buildings. All such decibel values that are related to 0 dB SPL are "absolute" values. Inputs and outputs are absolute values, as their intensities are defined in relation to their common ground of 0 dB SPL.

Absolute decibel values are based on logarithms (base 10). This is simply because for the normal-hearing person, the range of intensity from just barely audible to the threshold of feeling or pain is *huge*. The largest sound pressure one can generally tolerate (120 dB SPL) has a million times the pressure of 0 dB SPL. We do not want to deal with millions when dealing with audiometry; we would much rather deal with an audiometric range of 0 to 120. In terms of sound pressure, then, 20 dB SPL has 10 times the pressure of 0 dB SPL, 40 dB SPL has 100 times the pressure of 0 dB SPL, and so on until we get to 120 dB SPL, which has 1,000,000 times the pressure of 0 dB SPL.

Because the decibel is based on logarithms as just described, two absolute dB SPL values cannot simply be added together like 1 + 2 = 3. Consider, for example, a 1000-Hz tone at an intensity of 20 dB SPL. If we double its sound pressure by adding 20 + 20, the sum total is now 26 dB SPL. If we increase its pressure by a factor of 10, then we are now at 40 dB SPL.

The "fun" increases further still when we consider adding two tones together that are of equal intensity but different in frequency. A 1000-Hz tone at 20 dB SPL plus a 1500-Hz tone at 20 dB SPL equals a sum total of 23 dB SPL. Two *identical* machines each producing 85 dB SPL of noise, when combined together, would total 91 dB SPL. Of course, this is not usually the case in the real world when combining intensities. Two *different* machines, each producing 85 dB SPL of noise, when combined together total only 88 dB SPL!

Then again, in the real world, we are not always adding together two equal decibel values. Due to the fact that the decibel is based on logarithms, a 60-dB SPL sound has *lots* more pressure than a 50-dB SPL sound. Adding these two together basically produces a sum total that is slightly (but not much more than) 60 dB SPL. Here, 60 + 50 basically totals 60. By analogy, an elephant plus a mouse is basically an elephant.

Gain is a completely different decibel matter. As opposed to input and output, gain is a "relative" decibel value. One can add a 50-dB gain to a 10-dB SPL input or add it to a 50-dB SPL input. The gain here is therefore relative. That is why gain is stipulated in terms of simple "dB." Along with gain, then, here comes the good news. One *can* add a relative decibel value to an absolute decibel value like simple arithmetic (e.g., like 1 + 2 = 3). That's why in the world of hearing aid fittings, input + gain = output.

Back to our examples of machines, a combined total of two identical machines each making 85 dB SPL of noise results in a *gain* of 6 dB (85 dB SPL + 6 dB gain = 91 dB SPL). A combined total of two different machines each making 85 dB SPL of noise provides a *gain* of 3 dB (85 dB SPL + 3 dB gain = 88 dB SPL).

Those who measure dB SPL in worksite environments to assess the risk of noise-induced hearing loss must deal with the more complicated situation of adding absolute dB SPL values together. In our world of hearing aids, where gain is added to inputs to create outputs, we can be glad of the simpler way to add decibels. Of course with hearing aids, the gain is almost always more than 6 or 3 dB. For example, 10 dB SPL input plus 50 dB of gain equals an output of 60 dB SPL. So, also, 50 dB SPL input plus 50 dB of gain totals 100 dB SPL of output. One can readily see here that the gain of 50 dB is a relative value; it can be added to any input having any dB SPL.

There are other situations where we refer to dB and not dB SPL. For example, the ear canal resonance shown in Figure 5–3. Readers may note that the resonance in this figure is plotted in terms of gain, and in simple units of "dB." This is because it shows only the *added* decibels (or gain) resulting from resonance of the outer ear canal. The added gain due to the resonance is a relative value. Recall from our earlier discussion on that topic in the previous section, that this resonance—the unique shape of the gain added across the frequencies—could be added to any input intensity level, and not just to the 55 dB SPL input commonly used in yesterday's REM. On the other hand, the REURs showing the same resonance in Figures 5–5 and 5–6 are correctly depicted as the sum-total *output*; that is, the input (55 dB SPL) *plus* the gain provided by the outer ear canal resonance. That's why in those figures they are plotted in units of dB SPL.

For another example, the signal-to-noise (SNR) ratio described in Chapter 4 is also a *relative* value. Noise might be 80 dB SPL and the speech signal of interest might be 85 dB SPL. Then again, noise might be 60 dB SPL and the speech signal of interest might be 65 dB SPL. In both instances, the SNR is 5 dB, not 5 dB SPL.

For yet another example, consider the directional index (DI), which will be discussed in Chapter 9. The DI quantifies the sensitivity of a directional microphone to sounds coming from the front, compared to sounds arriving from all other directions. A microphone that is equally sensitive to sounds from all directions would have a DI of 0 dB. Another microphone might be 5 dB more sensitive to frontal sounds than to sounds from other directions, and so its DI would be 5 dB. Here again, the difference is a *relative* value; hence it is expressed in units of dB rather than in units of dB SPL.

So, also, dynamic range is expressed in units of dB and not in units of dB SPL. Let's say someone has a hearing threshold of 0 dB HL for 1000 Hz and a loudness tolerance level for the same frequency at 100 dB HL. The dynamic range here is the *difference* or "decibel distance" between the threshold or "floor" of hearing sensitivity and the "ceiling" of loudness tolerance; in this instance, the dynamic range is 100 dB (not 100 dB SPL). Compare this to another person whose threshold for 1000 Hz is 50 dB HL and loudness tolerance for the same frequency is 100 dB HL. Here, the difference or decibel distance between the "floor" and "ceiling" is smaller, giving a dynamic range of only 50 dB.

Later on, when we discuss the procedures of today's REM, we will describe what is known as real ear-to-coupler difference (RECD). ANSI testing, referred to earlier in this chapter, employs the use of a closed 2-cc coupler when measuring the output from a hearing aid. This means the air volume is 2 cc. The closed ear canal however, is smaller, about 1 to 1.5 cc in volume. The RECD then is simply the difference in the frequency response of a hearing aid while measured in a 2-cc coupler versus being measured in a real ear canal. Again, the relative difference here again is always expressed in simple "dB."

EFFECTS OF COMPRESSION ON GAIN (dB) VERSUS OUTPUT (dB SPL)

Today's REM displays the targets of today's compression-based fitting methods (to be discussed in the next chapter). It also can show the results of hearing aid compression per se (to be discussed in Chapter 7). As we have already addressed in the previous chapter, compression provides *different* amounts of gain for different input intensity levels. In so doing, the hearing aid's frequency response may change accordingly. Frequency