

Adaptive Dynamic Range Optimization (ADRO): A Digital Amplification Strategy for Hearing Aids and Cochlear Implants

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Adaptive dynamic range optimization (ADRO) is an amplification strategy that uses digital signal processing techniques to improve the audibility, comfort, and intelligibility of sounds for people who use cochlear implants and/or hearing aids. The strategy uses statistical analysis to select the most information-rich section of the input dynamic range in multiple-frequency channels. Fuzzy logic rules control the gain in each frequency channel so that the selected section of the dynamic range is presented at an audible and comfortable level. The ADRO processing thus adaptively optimizes the dynamic range of the signal in multiple-frequency channels. Clinical studies show that ADRO can be fitted easily to all degrees of hearing loss for hearing aids and cochlear implants in a direct and intuitive manner, taking the preferences of the listener into account. The result is high acceptance by new and experienced hearing aid users and strong preferences for ADRO compared with alternative amplification strategies. The ADRO processing is particularly well suited to bimodal and hybrid stimulation which combine electric and acoustic stimulation in opposite ears or in the same ear, respectively.

Introduction

Electronic hearing aids have been in existence for over 50 years, and new developments are common. New books continue to be written on the subject (Dillon, 2001; Valente, 2002a). Despite

this long history of research and clinical experience, there is still no consensus on the best amplification scheme or hearing aid fitting rationale for clinical use, although compression hearing aids continue to dominate (Dillon, 1996). The fundamental ideas underlying compression were

The research has been supported by the National Health and Medical Research Council of Australia, the Cooperative Research Centre for Cochlear Implant and Hearing Aid Innovation, the University of Melbourne, and the Bionic Ear Institute. The research was approved by the Human Research and Ethics Committee of the Royal Victorian Eye and Ear Hospital (Projects 99/379H, 02/458H, and 03/529H).

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suggested well before the advent of digital processing for hearing aids, and the principles of compression amplification were developed originally for analog implementation. The goal of the research and development described here was to use the capacity of modern digital signal processors to overcome some fundamental problems that are inherent in other amplification schemes.

One of the main constraints on hearing aid development is the sheer difficulty of implementing a practical solution to the amplification problem in a small hearing aid within a very limited power-consumption budget. Technologic developments are gradually easing these restrictions, and impressive feats of engineering have led to hearing aids with greater computing capability than the first personal computers. The first big step in this technologic revolution was the introduction of digital hearing aids. As shown in Figure 1, this happened in a series of stages that

has made the technology more flexible, from analog circuits, to application-specific integrated circuits (ASIC), and finally, to open-platform computers. In 2004, more than 80% of hearing aids sold in the United States used digital technology (Strom, 2005), and in 2005, one of the world's largest suppliers of components for hearing aids announced that they would no longer supply analog amplifiers.

The most obvious change that digital technology has brought to the hearing aid industry is the addition of digital signal-processing features such as adaptive directional microphones, feedback cancellation, and noise reduction (Chung, 2004a, 2004b). These features can offer advantages in many difficult listening situations, but the scientific evidence that they produce a significant increase in the satisfaction of hearing aid users is still fairly weak (Wong *et al.*, 2003). It is also clear that modern hearing aids and fit-

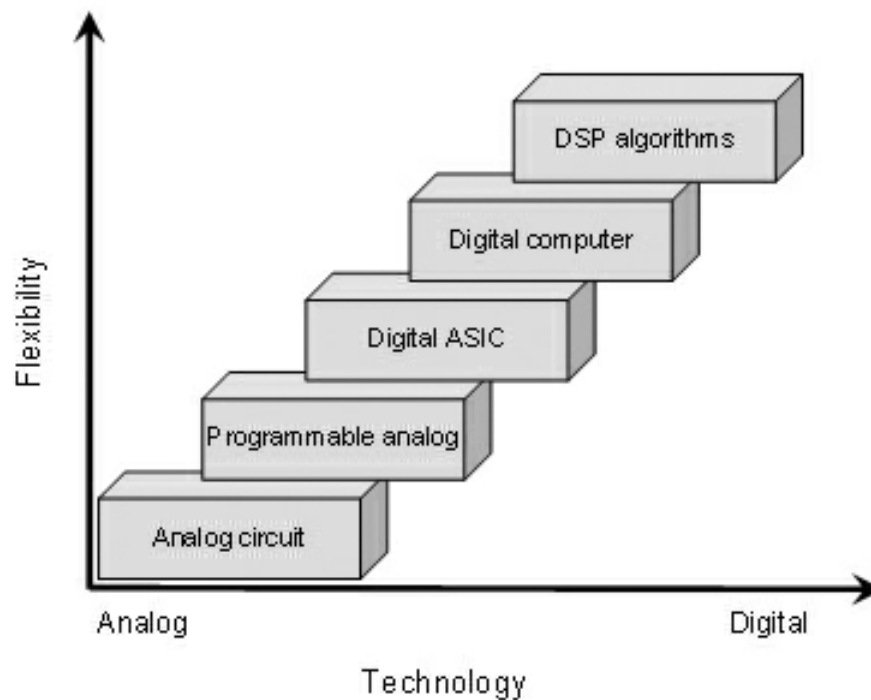


Figure 1. Hearing aid technology developments over time have only recently achieved the capacity and flexibility required to implement innovative digital signal processing algorithms (ASIC, application-specific integrated circuits; DSP, digital signal processor).

ting rationales do not adequately address some of the most fundamental requirements of hearing aid users. For example, Kochkin (2002) reports that 83% of hearing aid users would like to hear more soft sounds, and 81% find loud sounds uncomfortable.

The first goal of this paper is to review the fundamental requirements for a hearing aid amplification scheme and describe a fresh approach to meeting the needs of hearing aid users. This new approach would have been impractical to implement with analog technology and is thus an inherently digital approach that is especially designed to use the new capabilities of the emerging digital technology. The amplification strategy is called adaptive dynamic range optimization (ADRO) (Dynamic Hearing Pty Ltd) (Blamey *et al.*, 1999).

In parallel with the development of digital technology, the multichannel cochlear implant has revolutionized the treatment of severe-to-profound deafness in adults and children (Clark *et al.*, 1987). Through direct electrical stimulation of neurons in the cochlea, a cochlear implant produces a controlled sensation of hearing in the brain of the listener. The translation from a sound picked up by the cochlear implant system's microphone to the electrical stimulus requires a "sound processor" and a "sound-coding strategy". The sound-coding strategy performs the same function as the amplification scheme in a hearing aid: to make sounds audible, comfortable, and recognizable to the listener. It is not surprising, therefore, that improvements in amplification schemes and digital signal processing for hearing aids can often be used to improve the performance of cochlear implant sound processors. The converse is also true. The second goal here is to show how ideas originally applied to cochlear implants have been used to improve performance of hearing aids.

The ADRO amplification scheme is both a cochlear implant sound-coding strategy and a hearing aid amplification scheme, with similar implementations and fitting methods for both devices. It was originally designed to be used in a bimodal processor for a cochlear implant in one ear and a hearing aid in the other. It has now been validated in scientific studies for hearing aids (Martin *et al.*, 2001a, 2001b; Blamey *et al.*, 2004a, 2005) and for cochlear implants (James *et al.*, 2002; Dawson *et al.*, 2004). The results of these studies are summarized below. Cochlear implant

systems (Cochlear Ltd, Lane Cove, Australia) and hearing aids (Interton GmbH, Bergish Gladbach, Germany) that use ADRO are commercially available and are now being used in combination in bimodal studies.

The Differing Dynamic Ranges of Sound Signals and Residual Hearing

The sounds in our environment span a very wide range of intensities and frequencies, and the human ear also has a very wide range of sensitivity to both intensity and frequency of sound. Throughout this paper, *dynamic range* will refer to the range of intensity levels for an acoustic or an electric signal, either at the input or the output of a sound-processing device.

It is important to note that the dynamic range and the frequency range of human hearing are matched to that part of the sound environment that is most useful to humans. The matching is disturbed when hearing is impaired. The listener with impaired hearing misses out on important acoustic information in the environment, including those parts of speech, music, warning signals, and localization cues that fall outside their dynamic range of hearing. It is the main function of a hearing aid or cochlear implant to restore the listener's access to the information that would otherwise be missed. In other words, the input dynamic range and frequency range should be restored to the normal dynamic range and frequency range of hearing, if possible.

If the input dynamic range cannot be completely restored, it should at least be optimized. In this context, optimization means selecting the part of the environmental dynamic range that conveys the most information and presenting the information at a level that makes it most accessible to the listener. Unlike other sound processing schemes, ADRO uses continuous statistical analysis of the sound in each frequency channel to ensure that the maximum amount of information is selected in every channel. As Studebaker and Sherbecoe (2002) have shown, the most important information is contained within the upper part of the signal's dynamic range. They also indicate that the important dynamic range is different for different frequencies.

The second part of the optimization process is to present the information to the optimal part of

the hearing range. The requirement here is that the sound should be audible and comfortable to the listener. In other words, the sound should be not too soft and not too loud.

It is well known that the dynamic range of hearing (between threshold and maximum comfortable level) varies from one frequency to another. The equal-loudness curves in Figure 2 were produced by matching the loudness of pure tones at different frequencies to a standard tone at 1 kHz (Robinson and Dadson, 1956; ISO 226, 2003). Figure 2 uses a loudness scale called the *Phon scale*. The Phon scale assigns the loudness in Phons to be equal to the intensity of the standard 1 kHz tone in dB sound pressure level (SPL). As Figure 2 shows, equal-loudness curves vary across frequency, and the dynamic range is greatest for frequencies between 500 Hz and 6 kHz. Outside this frequency range, the normal ear has a smaller dynamic range of hearing. In Figure 2, the 90 Phon curve is the point at which sounds

tend to become very loud, corresponding to a 90 dB SPL pure tone at 1,000 Hz.

The curves shown in Figure 2 are averages for a sample of the normal-hearing population. When dealing with individual listeners, there will be variation from one person to another in judgments of what is “too loud” and what is “too soft”. These individual variations, or preferences, are catered for in normal-hearing listeners by using volume and tone controls on devices such as mobile phones, radios, televisions, and compact disc players. Similarly, there are variations between hard-of-hearing individuals, and an effective optimization scheme for hearing aids needs to adjust the output dynamic range to the listener’s individual judgments of what is too loud and too soft at each frequency.

For hard-of-hearing people, the dynamic range at each frequency will be reduced compared with the ranges shown in Figure 2. Figure 3 shows equal loudness contours for nine subjects

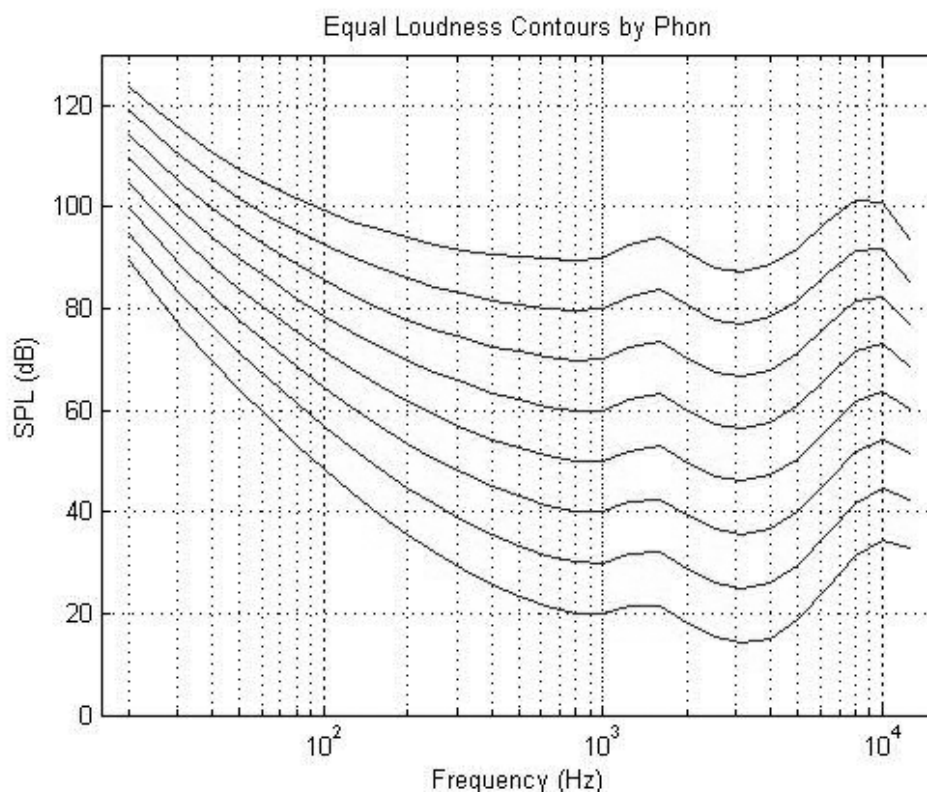


Figure 2. Equal loudness contours for listeners with normal hearing according to ISO 226 (2003). SPL, sound pressure level.

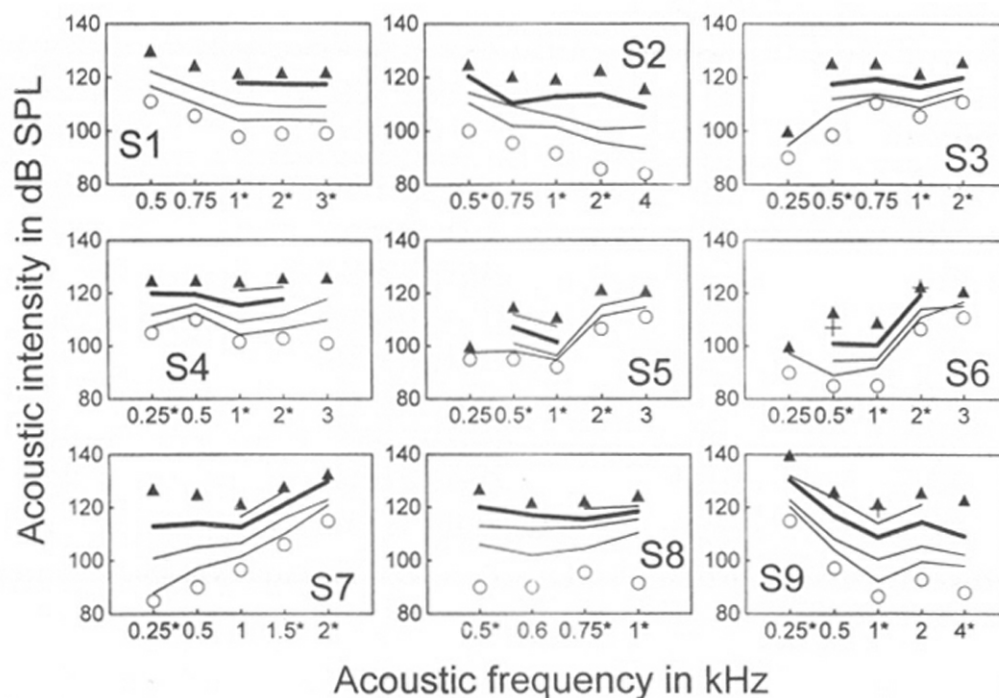


Figure 3. Acoustic equal loudness contours for nine listeners with profound hearing loss. The contours correspond to very soft, soft, medium (bold), and loud sounds. Minimum and maximum levels presented are shown by circles and triangles, respectively. SPL, sound pressure level. Reprinted with permission.

who have a profound hearing loss (Blamey *et al.*, 2000). Each panel represents the data for one subject. These results for individuals are analogous to the equal loudness contours for the normal-hearing population in Figure 2 (with some differences). It is not possible to assign the loudness in Phons to each contour because the loudness of a 1,000 Hz tone is unknown for each subject. (Note that there is an underlying assumption in the Phon scale that all people with normal hearing perceive the same loudness for the standard 1,000 Hz tone. This assumption does not apply to people with impaired hearing.) Clearly, large differences exist between individuals, and the dynamic range of residual hearing is much narrower than for normal-hearing listeners.

As shown in Figure 3, thresholds are usually elevated more than maximum comfortable levels or loudness discomfort levels when hearing is impaired. This is the well-known “recruitment” phenomenon described by Fletcher and Munson (1937). They suggested that there was abnormal

loudness growth for people with impaired hearing. Subsequent physiologic research has shown that outer hair cells operate nonlinearly at relatively low intensity levels to expand the dynamic range of hearing. When outer hair cells are damaged or lost altogether, the dynamic range of hearing is reduced to the range of sensitivity of the inner hair cells. If inner hair cells are damaged or lost, the dynamic range of hearing will be reduced further. Fletcher and Munson (1937) were also the first to suggest compression of the acoustic signal as a means of compensating for steeper than normal loudness growth.

The recent psychoacoustic research of Buus and Florentine (2002) questions the “abnormal loudness growth” explanation and suggests that loudness growth is normal within the reduced range of hearing. This research brings into question the theoretical basis for compression as an amplification strategy that compensates for abnormal loudness growth. Compression can be an effective method for reducing dynamic range, but

it also has consequences for the relative loudness of sounds that may not be in accord with the latest psychoacoustic research. ADRO does not compress the dynamic range of the signal. Instead, it maintains the natural intensity variations over time within the selected portion of the dynamic range in each frequency band as long as the sounds are both audible and comfortable.

There is also a considerable body of psychophysical literature on the loudness of electrical signals applied to the cochlea (Tong *et al.*, 1983; Shannon, 1983; McKay *et al.*, 2003; Blamey *et al.*, 2000, 2004b). The variability in electrical thresholds and comfortable levels for individuals using electrical stimulation is greater even than the variability for acoustic stimulation in people with impaired hearing. There is also no universally accepted theoretical description of loudness growth for electrical stimulation. Thus for pragmatic reasons, the output dynamic range used by ADRO and other cochlear implant sound-processing

schemes is usually defined subjectively as the electrical stimulation range between threshold and maximum comfortable level on an individual basis.

Figure 4 shows equal loudness contours for nine subjects who have a cochlear implant in one ear (Blamey *et al.*, 2000). Each panel represents the data for one subject. These electric stimulation loudness contours are analogous to the acoustic equal loudness contours shown in Figure 3. In fact, the nine subjects are the same people for the two sets of data that were collected using an identical procedure. Each subject normally wore a cochlear implant in one ear and a hearing aid in the other. The vertical axis in Figure 4 is in stimulus level units, and the horizontal axis represents different electrode positions in the cochlea. Stimulus level units are used in programming the Nucleus 22-channel implant that were devised to give 255 approximately equal loudness steps by varying both pulse width and

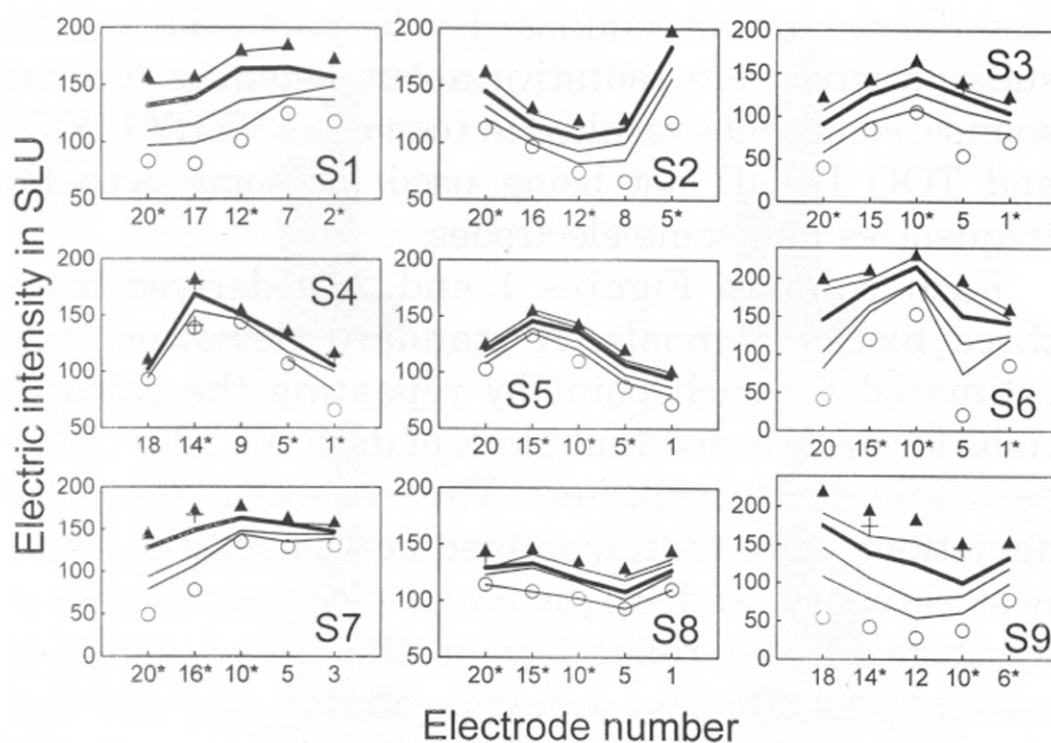


Figure 4. Electric equal loudness contours for nine listeners with cochlear implants. The contours correspond to very soft, soft, medium (bold), and loud sounds. Minimum and maximum levels presented are shown by circles and triangles, respectively. SLU, stimulus level units. Reprinted with permission.

amplitude. The steps are equally spaced on a logarithmic scale like a dB scale. An increase of 12 stimulus level units corresponds to an approximately 1-dB increase in electric charge per pulse (see Blamey *et al.*, 2000, Appendix A for further details). For the implanted ear, as well as the nonimplanted ear, there are large differences between individuals, and the dynamic range of hearing is narrow.

The Need for a Bimodal Amplification Scheme

In the 1990s, it became clear that people with severe hearing loss could benefit from cochlear implants and they could gain added benefit if they continued to use a hearing aid in the non-implanted ear (Dooley *et al.*, 1993). Although early bimodal listeners clearly benefited from

both devices, it was a natural question to ask how the bimodal benefit could be maximized. An obvious requirement is that both devices should produce an audible and comfortable signal over a range of input signals so that both ears can contribute to the binaural hearing sensation. This requirement was investigated using loudness data from bimodal listeners (Blamey *et al.*, 2000).

The equal loudness contours shown in Figures 3 and 4 were used to calculate the input dynamic range corresponding to the audible and comfortable dynamic range at the output of the hearing aid and cochlear implant using standard hearing aid and cochlear implant fittings appropriate for the nine subjects. The results of the calculations are shown in Figure 5. In every case, the implant provided access to a wider input dynamic range and frequency range than the hearing aid, which explains why these subjects were good cochlear implant candidates.

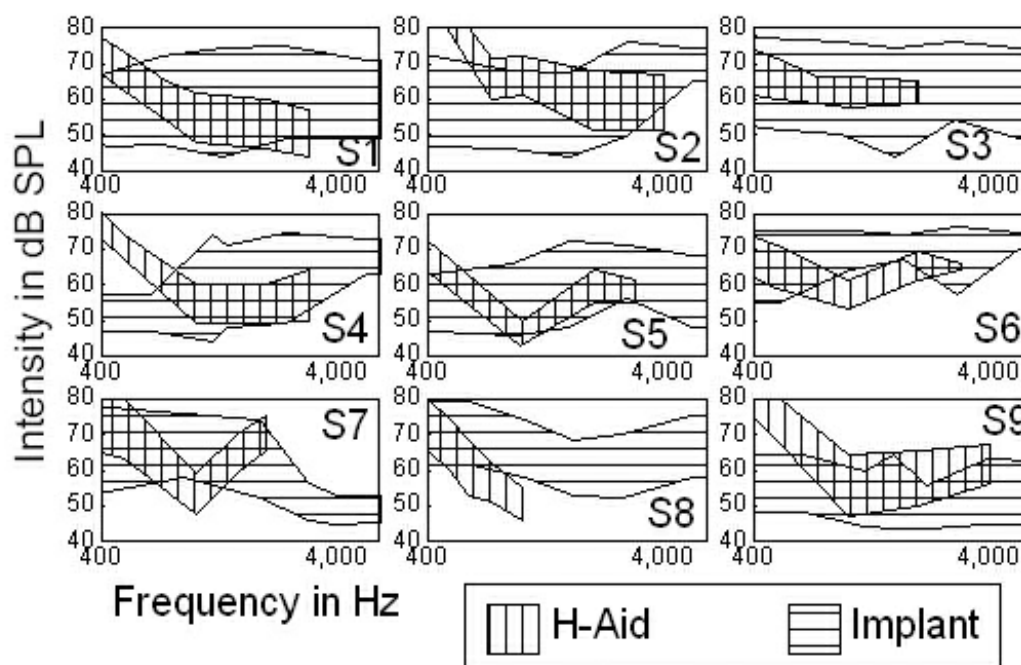


Figure 5. Calculated free-field input dynamic ranges for nine listeners with cochlear implants and hearing aids in opposite ears. The areas indicate the intensity and frequency of input signals that fall within the audible to comfortable range of each listener. The vertically shaded region is the input dynamic range of the hearing aid. The horizontally shaded region is the input dynamic range of the implant. SPL, sound pressure level. Reprinted with permission.

The overlap of the hearing aid and implant input dynamic range is the “sweet spot” of hearing where sounds are audible and comfortable in both ears at once. Sounds in the sweet spot can be processed binaurally, giving rise to improved speech perception and other benefits (Armstrong *et al.*, 1997; Blamey *et al.*, 1997; Dooley *et al.*, 1993; Ching *et al.*, 2001, 2004; Tyler *et al.*, 2002). Ideally, the sweet spot should be as large as possible to maximize the binaural benefits of fitting a cochlear implant and hearing aid in opposite ears. ADRO was specifically designed to optimize the input dynamic range for implants and hearing aids alike, and thus maximize the sweet spot for bimodal listeners.

In summary, there is often a mismatch between the dynamic range of the information-carrying signals in the environment and the dynamic range of hearing for people with impaired hearing. The ADRO processing scheme is designed to select the most information-rich part of the environmental dynamic range for every frequency channel and present the information to the optimal part of the individual listener’s dynamic range of hearing for every frequency channel. This approach applies equally well to hearing aids and cochlear implants and is thus also ideal for bimodal use.

The Differences and Similarities of Hearing Aids and Cochlear Implants

Conventional amplification schemes for hearing aids make assumptions about the dynamic ranges of both the input signals and the output signals. On the input side, research by Dunn and White (1940) provided information about typical speech levels, dynamic ranges, and long-term average speech spectra that were the basis for most linear and nonlinear fitting prescriptions for hearing aids (Skinner, 1988; Valente, 2002b; Byrne and Dillon, 1986; Cornelisse *et al.*, 1995). Subsequent research has indicated that long-term average speech spectra are significantly different for male and female speakers in the low and high frequencies (Cox and Moore, 1988).

Environmental conditions also vary widely from the quiet conditions on which hearing aid prescriptions are based. Thus, volume controls, multiple programs, and additional processing are often required to maintain optimal sound pro-

cessing in environments that are not within the range for which the amplification scheme was designed.

Wide dynamic range compression provides wider input dynamic range than other compression schemes and, therefore, encompasses a greater range of input levels and environments. ADRO uses an alternative adaptive approach that does not make assumptions about the input spectrum or level and can broaden the effective input dynamic range even further than wide dynamic range compression.

Hearing threshold measures are usually used to predict the required input/output functions for the hearing aid, to place the output signal into the listener’s dynamic range of hearing. For many years, optimizing the hearing aid fitting was synonymous with adjusting the gain and frequency response of the hearing aid to achieve a close fit to the prescribed gain function, either in a hearing aid test box, or with real ear measures. Fittings prescribed in this manner are based on averages of what is suitable or preferred by sample populations of hearing aid users. They do not take into account individual differences or preferences, and so hearing aids usually require a volume control and may also require tuning of the frequency response to suit the individual. The amount of tuning may be small for a listener who is close to the hypothetical average listener or greater for a listener who is far from the average.

Cochlear implants and hearing aids perform similar functions. They continuously monitor sounds in the listener’s environment, process the sounds, and present them to the listener in a form that is more accessible to the impaired auditory system. At the input, the range and types of sounds to be processed are the same. At the output, the goals of the processing are also the same: to produce hearing sensations that are audible, comfortable, and convey the maximum amount of information to the listener. In between the input and the output, the processing requirements are different: a cochlear implant excites residual neurons in the cochlea directly using electrical stimulation, and a hearing aid excites residual neurons in the cochlea indirectly via the residual hair cells using acoustic stimulation.

Cochlear implants deliver multiple electrical stimuli to electrodes at different positions in the cochlea, representing multiple-frequency bands in the input sound. Each electrode is designed to stimulate neurons that are close to the electrode.

The tonotopic ordering of neurons in the cochlea ensures that electrodes that are inserted further into the cochlea (apical electrodes) produce lower pitched or duller hearing sensations, and electrodes closer to the basal end of the cochlea produce higher pitched or sharper hearing sensations (Tong *et al.*, 1983; Blamey *et al.*, 1996). All present-day cochlear implants separate the input signal into multiple-frequency bands and deliver the band-limited signals to separate electrodes. They differ in the numbers of channels and electrodes as well as in the manner in which the band-limited signals are delivered to the electrodes.

Alternative methods are sometimes called coding schemes or sound-processing strategies. Examples include simultaneous analog stimulation (SAS), continuous interleaved sampling (CIS), and a variety of pulsatile coding strategies that selectively excite electrodes in a pattern that represents the sound spectrum (SPEAK, ACE, etc). ADRO has been implemented as a pulsatile sound-coding strategy for the 22-channel cochlear implant produced by Cochlear Ltd.

SAS is the cochlear implant processing strategy that is closest to a compression hearing aid. It delivers a highly compressed electrical version of each frequency band signal to each electrode pair. SAS relies on the physical separation and geometry of the electrode array in the cochlea to keep the frequency channels separated. When the electrical signals are delivered to the cochlea, they add together in regions where they overlap. One of the reasons why SAS systems usually have a relatively low number of channels is that the channels need to be relatively far apart in the cochlea. For cochlear implants with relatively high numbers of channels, pulsatile stimulation is used. The pulses are separated in time (interleaved) to avoid the summation of electric currents that occurs when stimuli are simultaneous.

In a hearing aid, the processing is often carried out in frequency bands, as for cochlear implants. However, the frequency bands in a hearing aid are added together after processing and before delivery to the output transducer of the hearing aid. The ear processes the combined broadband signal, and a second separation into frequency components takes place along the basilar membrane.

There is, of course, no technical requirement that the processing bands in the hearing aid should match the biophysical frequency bands that occur in the acoustically stimulated ear. This

is a major difference between hearing aids and cochlear implants where the number of processing bands is equal to the number of electrical channels for delivery of the signal. A more subtle difference is the degree of overlap or independence of the frequency bands. In a hearing aid, the degree of overlap depends on the design of the filter bank used to separate the signal into frequency bands. This is entirely under the control of the hearing aid designer. In a cochlear implant, the overlap of the channels depends on the spread of electric current in the cochlea and the number and position of the surviving neurons. These are not parameters that are controlled by the signal processor in the cochlear implant.

The Goals of Amplification for Hearing Aids and Cochlear Implants

Despite the inherent differences between hearing aids and cochlear implants, the processing requirements are very similar. The most fundamental requirement is that sounds should be audible and comfortable—not too soft and not too loud. Additional goals that have been applied in the past include maximum intelligibility for speech in quiet, maximum intelligibility for speech in noise, high fidelity for music, and natural sound quality. It should be noted that the concepts of “too soft,” “too loud,” and “natural,” are all subjective, and listeners may vary quite widely in their preferences for how a hearing aid should be tuned to meet their preferences, even if they have similar audiograms.

It should also be noted that even normal hearing does not satisfy all the requirements. Sometimes, sounds are too soft to hear. Some sounds will be too loud. Speech in noise is hard to understand for low signal-to-noise ratios. Therefore, even if we were able to restore normal hearing with a cochlear implant or a hearing aid, there would still be some situations in which the listener might not be satisfied with the performance of the device.

There is also the possibility that a hearing aid could be designed to provide hearing that is “better than normal.” In fact, such devices do exist, although they are not classified as hearing aids. For example, directional microphones are used to improve speech intelligibility in noisy situations, active noise cancellation headsets are used to re-

duce background noise in airplanes and cars, and hearing protectors are used to make loud sounds more comfortable in industrial environments. ADRO is being applied in headsets and other devices for listeners with normal hearing to provide improved audibility and intelligibility in noise, to compensate for poor telephone transmission lines, and to protect hearing from loud sounds and acoustic trauma.

Just as people with normal hearing may have special requirements and preferences, listeners with hearing aids and cochlear implants may also have special requirements and preferences. The goals of both implants and hearing aids, therefore, are to meet the hearing needs and preferences of the individuals who will use the devices.

The ADRO Amplification Scheme

The fundamental requirements of audibility and comfort apply to cochlear implants as well as to hearing aids. It is therefore an easy transition to use ADRO processing for cochlear implants and for hearing aids. It can be fitted in a very similar way to both devices, and the implementation of

the amplification scheme is also very similar in both devices. Figure 6 shows a typical implementation of an ADRO hearing aid with 64-channels. For a cochlear implant implementation, the inverse fast Fourier transform stage would be omitted, and the output levels of the individual channels would be “mapped” onto the electrical stimulus levels of the electrodes. If the cochlear implant had fewer than 64 electrodes, the number of frequency channels from the fast Fourier transform could be reduced by combining adjacent frequency components either before or after the ADRO processing was applied.

ADRO uses “fuzzy logic rules” to optimize the output signal of the hearing aid in each narrow frequency channel. A fuzzy logic rule is one that is not always true or false, but can be true for part of the time (see the description of the comfort and audibility rules below). The rules ensure the comfort and audibility of sounds by keeping the output level between a comfort target and an audibility target. If a sound falls below the audibility target, it is made louder. If it rises above the comfort target, it is made softer. While the sound is within the audible and comfortable range, the hearing aid gain does not change. Instead, it operates in a linear fashion.

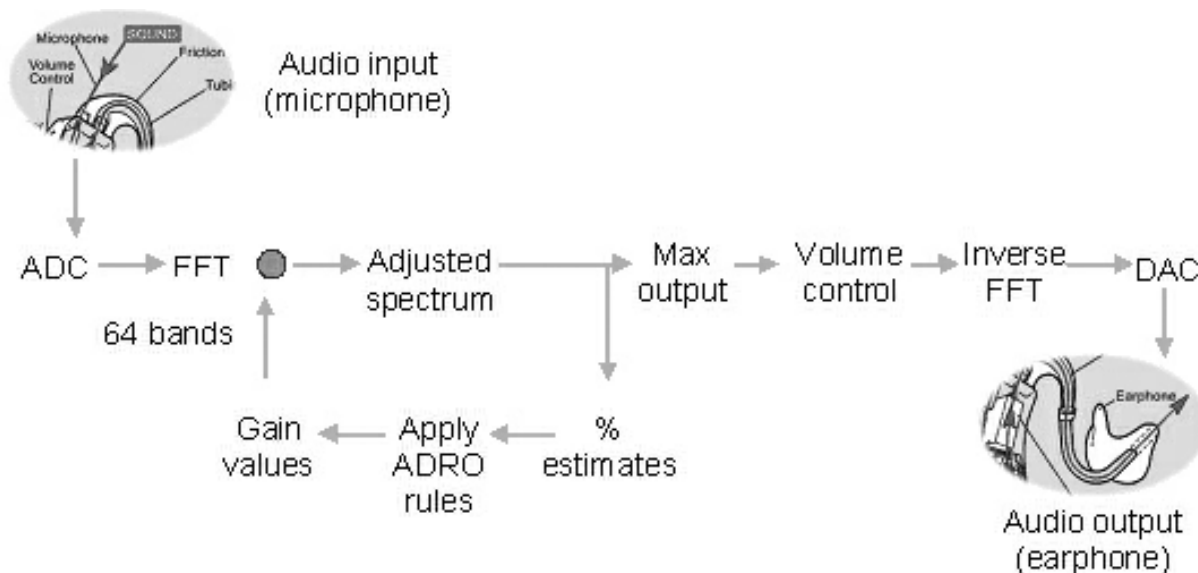


Figure 6. Sound processing stages for a typical 64-channel implementation of ADRO in a hearing aid. ADC, analog-to-digital converter; DAC, digital-to-analog converter; FFT, fast Fourier transform.

This approach is quite different from the alternative compression strategies that continuously vary gain according to fixed input/output functions in a smaller number of broader, overlapping frequency bands. Figure 7 shows the difference between a fixed input/output function for one channel of a compression hearing aid (in the right-hand panel) and the adaptive linear processing for one ADRO channel (in the left-hand panel).

The linear input/output function for ADRO is selected by using statistical rules that keep the sound comfortable and audible. Four ADRO processing rules are applied independently in each frequency channel. They use statistical analysis of the sound intensity in each channel to control the loudness of the sound.

- The “comfort rule” reduces the gain in the frequency channel if the output level for the channel exceeds the “comfort target” more than 10% of the time. This rule ensures that sustained sounds are not too loud.

- The “audibility rule” increases the gain in the frequency channel if the output level for the

channel falls below the “audibility target” more than 30% of the time. The audibility rule ensures that sustained sounds are not too soft.

- The “hearing protection rule” limits the output level in each channel so that it never exceeds a maximum value. This rule ensures that sudden loud sounds are not uncomfortably loud.

- The “background noise rule” limits the maximum gain in each channel. The background noise rule ensures that low-level background noise is not over-amplified to a level that becomes annoying to the listener.

The overall effect of the four rules is to select the most information-rich part of the sound dynamic range and place it between the audibility and comfort targets. This is illustrated visually in Figures 8A, B, C, and D. In this visual analogy, the upper part of the picture corresponds to high-intensity sounds, and the lower part to low-intensity background noise. The central part of the picture is the most information-rich section. Figure 8A illustrates the fact that normal-hearing listeners have a broad dynamic range of hearing encompassing both high- and low-intensity

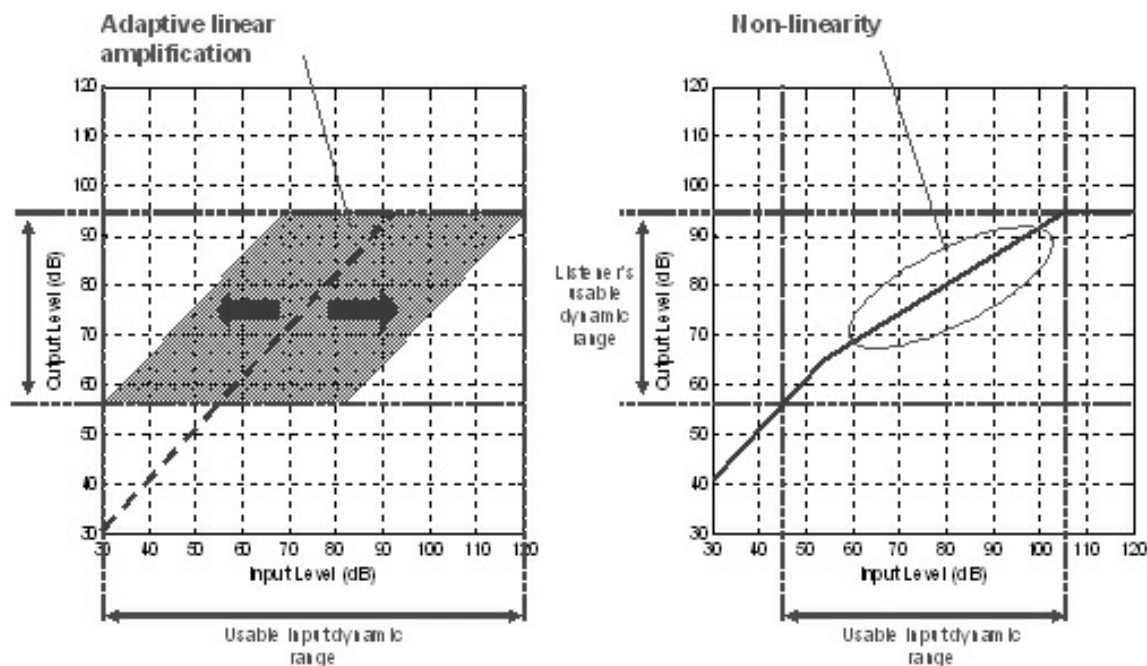


Figure 7. Compression amplification uses a fixed non-linear input/output function (right panel) in each frequency-channel. ADRO operates on a linear input/output function within the shaded region (left panel) in each frequency channel.

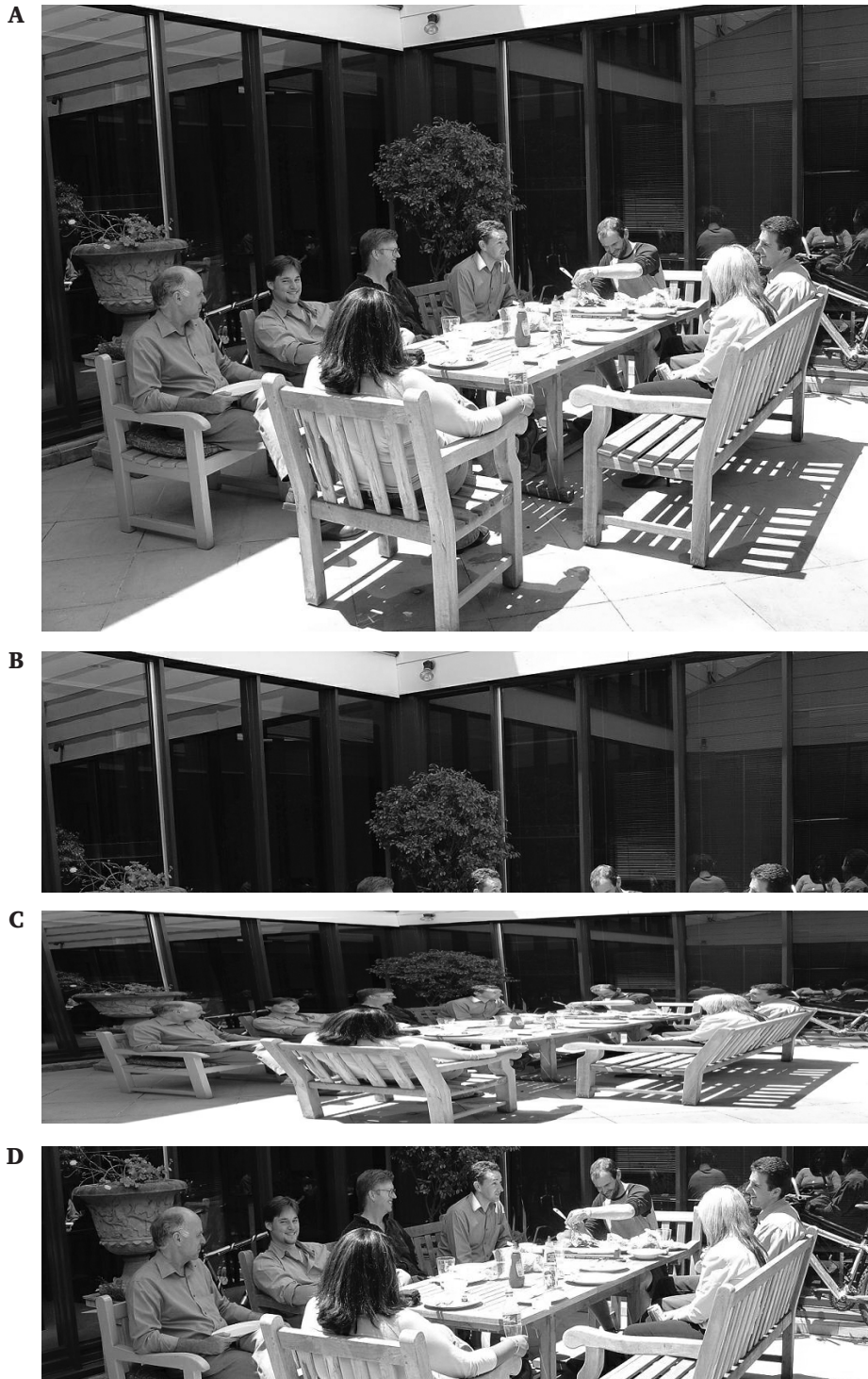


Figure 8. (A) A listener with normal hearing has access to sounds from low to high intensity to “hear the full picture”. (B) Without amplification, a listener with impaired hearing can hear only the most intense sounds, and the lower part of the picture is inaudible. (C) A compression hearing aid amplifies low intensity sounds more than high-intensity sounds so they all fall within the listener’s reduced range of hearing. (D) An ADRO hearing aid selects the most informative part of the sound and presents it within the listener’s reduced range of hearing without distortion.

sounds. Hearing loss limits the range of hearing by making less intense sounds inaudible (Figure 8B). Compression restores the audibility of soft sounds but at the expense of distortion and reduction of the loudness contrasts within the sound, as illustrated by Figure 8C. Figure 8D shows the effect of ADRO, which selects the most information-rich section of the picture and aligns it with the reduced dynamic range of hearing.

Figures 7 and 8 illustrate fundamental differences between compression and ADRO amplification schemes for sounds that include both low- and high-intensity components. A further difference between compression and ADRO amplification is found with hearing aid test box measurements where the most commonly used signal is a pure tone. The input/output function shown in the right panel of Figure 7 is the shape that would be traced out in a hearing aid test box if a pure tone was increased and decreased in intensity at a reasonably slow rate.

Following the same procedure for an ADRO hearing aid in a hearing aid test box will result in a different shape curve, as shown in Figure 9.

Starting with a low input level at the lower left corner, the input/output function will trace out a linear path until the output level reaches the comfort target. At this point, the comfort rule will reduce the gain of the frequency channel containing the pure tone, and the output level will no longer increase as the input level increases, keeping the output “comfortable.” If the input level of the pure tone is then reduced, the output level will fall, following a linear path, until the output level reaches the audibility target. At this point, the audibility rule will increase the gain of the channel to keep the output “audible.” When the channel gain reaches the maximum value specified by the background noise rule, the output level will again fall linearly as the input level is reduced. Thus the input/output function for ADRO is a loop, with linear sides and nonlinear sections where the comfort and audibility rules come into play. By contrast, the input/output function for compression is nonlinear over the entire input range above the kneepoint, as shown in Figure 7.

This section has described the concepts underlying the ADRO amplification strategy and

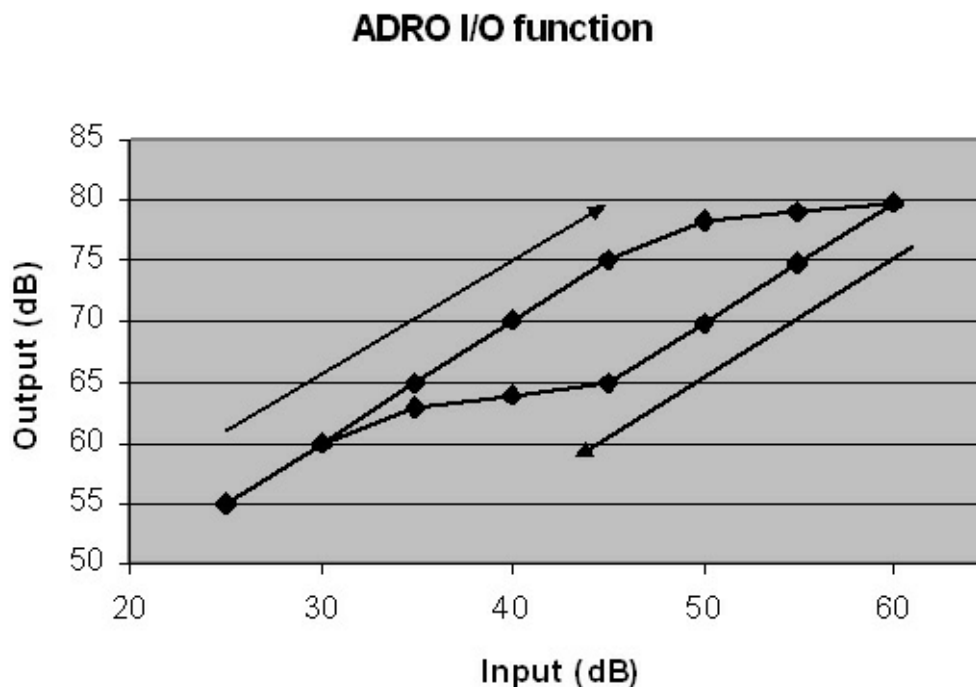


Figure 9. The input/output function for ADRO measured with a pure-tone in a hearing aid test box is a closed loop.

compared its effects with compression amplification. Clinical studies have shown that ADRO processing can improve both audibility and comfort without compromising sound quality and intelligibility. For a compression amplification scheme to improve both audibility of low-intensity sounds and comfort of high-intensity sounds, an increase in both gains and compression ratios is required. It is well known, however, that high compression ratios tend to distort sounds, reducing intelligibility and sound quality (Hornsby and Ricketts 2001; Neuman *et al.*, 1994; 1998). This is the main advantage of ADRO over alternative amplification schemes, but there are also other differences that are described below.

Narrow-Band Frequency Analysis

The frequency analysis used by ADRO uses a greater number of narrower bands than alternative amplification schemes for hearing aids. For example, the initial clinical evaluations of ADRO compared 64-channel ADRO with single-channel linear processing (Martin *et al.*, 2001a), three-channel compression in a behind-the-ear hearing aid, and nine-channel compression in an in-the-ear hearing aid (Blamey *et al.*, 2004a). The 64 channels each had a bandwidth of 125 Hz, spanning the range from 125 to 8000 Hz. A 32-channel version with a 250-Hz bandwidth for each channel has also been developed and used in further unpublished evaluations.

The initial reasons for choosing a large number of narrow channels were purely pragmatic, but this choice has turned out to have some interesting and worthwhile benefits. The initial choice was made because the first open-platform digital signal processor (DSP) available in commercial hearing aids was the Toccata hybrid developed by the Dspfactory, LTD of Waterloo, Canada (Brennan and Schneider, 1998). This device has an architecture that is highly optimized for the calculation of discrete Fourier transforms and is ideal for the implementation of the basic ADRO processing. The only potential disadvantage of the choice to use many channels is the relatively long group delay of 13 milliseconds imposed by the digital sampling in blocks of 128 samples at a sampling rate of 16 kHz. This is discussed in the subsequent section on time delay.

The benefits of using many narrow channels are flexibility and optimum performance in background noise. The flexibility to shape the maximum gain, maximum output levels, comfort targets, and audibility targets at many frequencies makes it relatively easy to fit ADRO to any hearing loss. With a smaller number of frequency channels, the frequency boundaries and the degree of overlap of the channels can limit the shapes of hearing losses that can be fitted effectively. Steep ski-slope audiograms are difficult to fit with a hearing aid that has only a few frequency channels. With many narrow channels, variable frequency boundaries between the channels are not necessary, and it is much easier to fit all shapes of hearing loss accurately, including ski-slope audiograms, because the gain and other parameters can be specified at more frequency points.

It is sometimes possible to improve the signal-to-noise ratio. If, for example, the noise has a narrow frequency distribution, then the gain in the frequency bands containing the noise can be reduced relative to the other frequency bands and the overall signal-to-noise ratio will be increased (Eatwell, 2002). The part of the speech signal that lies within the same frequency channels as the noise will also be reduced, but speech in other frequency channels will be preserved. If there are only a few frequency channels, it is likely that more of the speech will be reduced than is necessary to reduce the noise, and the improvement in signal-to-noise ratio will be smaller than if there were a larger number of narrower channels.

In cases where the signal and noise have the same average spectral shape, then filtering out the noise with a fixed filter will not change the long-term signal-to-noise ratio because the speech will also be filtered out. On the other hand, there will be short time windows in which the spectral shapes of the speech and the noise are different. A narrow-band analysis with fine-time resolution followed by adaptive filtering is required to take advantage of the spectral and temporal differences between speech and noise. The result is that evaluations of ADRO have shown improved performance in background noise compared with alternative amplification schemes with broader frequency channels. This result is consistent with the work of Yund and Buckles (1995), who showed improvements for speech intelligibility in noise for 8-channel and 16-channel compression hearing aids compared with aids with lower numbers of channels.

Implementation of Percentile Estimators

This section contains a detailed technical description of the statistical analysis. It can be skipped without affecting the reader's understanding of ADRO.

After splitting the input signal into frequency channels and multiplying each channel by its corresponding channel gain, the ADRO processor accumulates statistical measures of the output dynamic range for each channel. This is a feature of the ADRO amplification scheme that distinguishes it from alternative sound-processing schemes. The statistical calculations are performed continually by two “percentile estimators” for each frequency channel. There is no fixed time window for the statistical analysis. A “percentile” is a statistic calculated from a distribution of values. For example, the 90th percentile is the value below which 90% of the values in the distribution lie and above which 10% of the values lie. Figure 10 shows a hypothetical distribution and the positions of the 90th percentile and the 30th percentile.

For speech in quiet, the double-peaked shape of the intensity distribution in Figure 10 is common, with the lower intensity peak representing low background noise and the upper intensity peak representing speech. In a hearing aid implementation of ADRO, the 90th percentile of the intensity distribution is used to implement the comfort rule and the 30th percentile is used to implement the audibility rule in each frequency channel.

The percentile estimation process is an iterative one. The sound intensity in each frequency channel is measured at regular time intervals. At any instant in time, there is a number that represents the current value of the estimate. Each successive measurement of the sound intensity in a channel is compared with the current estimate. If the new measurement is greater than the current estimate, then the estimate is increased by a small amount (the upward step). If the new measurement is below the current estimate, then the estimate is reduced by a small amount (the downward step).

For example, if the size of the upward step is nine times the size of the downward step, then

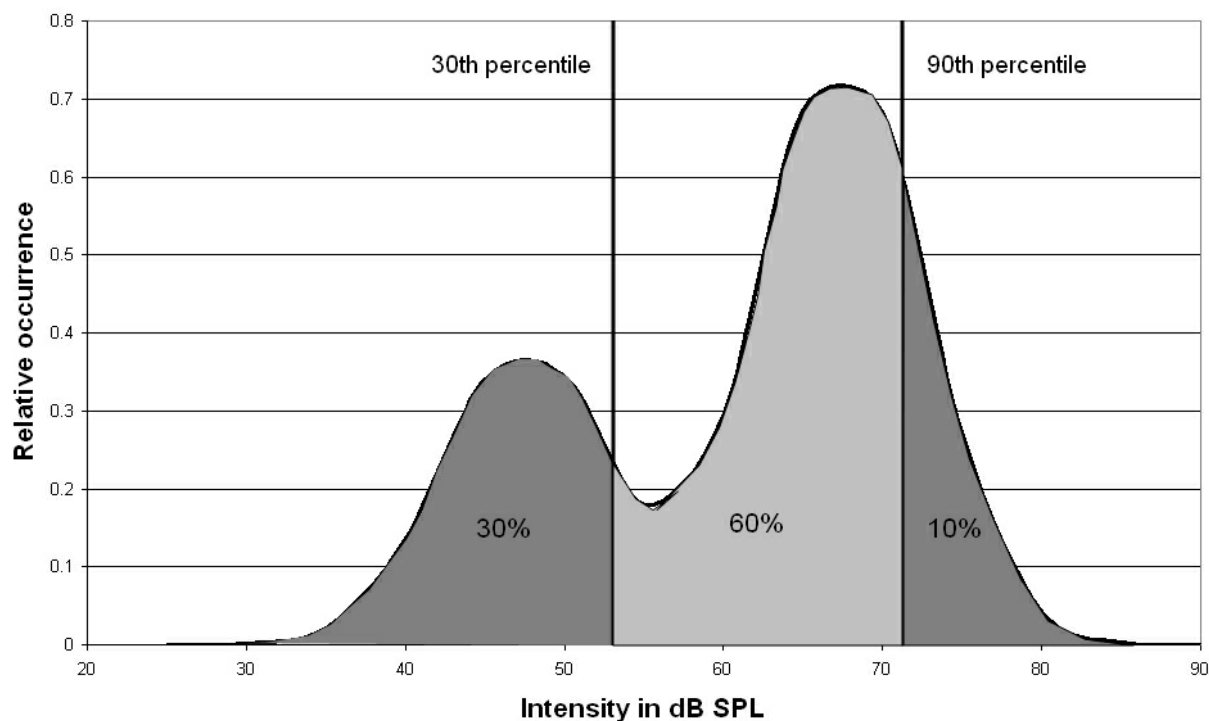


Figure 10. Hypothetical distribution of intensity measurements in one frequency channel, showing 30th and 90th percentiles. SPL, sound pressure level.

the current value of the percentile estimate will converge to the 90th percentile after many iterations. If the current estimate is below the 90th percentile, more than 10% of the intensity measurements will result in upward steps, and the estimate will increase. If the current estimate is above the 90th percentile, then more than 90% of the intensity measurements will result in downward steps, and the estimate will decrease. When the percentile estimate has converged to the 90th percentile, then 90% of the intensity comparisons will result in downward steps and 10% will result in upward steps. Therefore there will be nine times as many downward steps as upward steps, but the upward steps are nine times the size of the downward ones, so the overall effect is no change to the estimate. Similarly, the 30th percentile may be estimated by making the ratio of the upward step to the downward step three as to seven.

The percentile estimates may be made to converge quickly if the upward and downward step sizes are chosen to be relatively large. Conversely, the step sizes may be chosen to be smaller, and the percentile estimates will change more slowly. In both cochlear implant and hearing aid ADRO processors, the percentile estimates change at about 20 dB/s.

Time Delays and Time Constants in Adaptive Processing

Digital signal processing takes time, and therefore, there is a delay between the input and the output of a hearing aid or cochlear implant speech processor. Typically, the analog-to-digital and digital-to-analog converters may each incur about a 0.5-millisecond delay. These delays are unavoidable and are determined by the hardware rather than by the digital signal-processing algorithm itself. When a discrete Fourier transform or a fast Fourier transform is used, a further time delay is incurred because the transform operates on a block of digital samples that must be collected before the processing can begin. A delay is also incurred during the inverse transform and overlap add processing that is used to synthesize the output waveform as the final digital signal-processing step.

Most of the ADRO processors use fast Fourier transform processing and so they incur a delay of

several milliseconds. At a sampling frequency of 16 kHz, and using 64 frequency channels, the delay is 13 milliseconds. The delay is 7 milliseconds for 32 frequency channels, and 4 milliseconds for 16 channels. Equal delays would be incurred by alternative amplification schemes running on the same digital signal processor and using the same fast Fourier transform configuration and sampling frequency. Versions of ADRO processing with lower time delays are under development, using time domain filtering instead of fast Fourier transform.

The 13-millisecond delay for 64-channel ADRO processing is well below the maximum tolerable delay of 20 milliseconds recommended by Stone and Moore (1999). Research by Agnew and Thornton (2000) showed that listeners with normal hearing find delays between 3 and 5 milliseconds “noticeable”. The 16-channel ADRO processing falls within this range.

Other time constants are more important than time delays for ADRO and other adaptive processing. Time constants such as the attack and decay times for compression have strong effects on sound quality for the listener (Neuman *et al.*, 1998; Hansen, 2002). Short time constants result in rapid changes in gain and therefore introduce nonlinear distortion into the signals. Short time constants also affect sound quality by reducing the contrasts between successive sounds with different levels and reducing the dynamics of music, for example. On the other hand, fast reduction of gain is required to protect the listener from loud sounds with abrupt onsets. A compromise is therefore necessary between reduced sound quality (fast time constants) and discomfort for sudden loud sounds (slow time constants).

The ADRO processing avoids this compromise by using different rules to protect the listener from sudden loud sounds and to adapt gain in response to milder changes in sound intensity. The hearing protection rule operates instantaneously to limit sudden loud sounds. These sounds are never allowed to exceed a limit that is set in each frequency channel. The audibility rule and the comfort rule operate more slowly to keep the sound within the listener’s range of hearing in each frequency channel. The rate of change of gain is 3 to 6 dB/s for the comfort and audibility rules of ADRO. Experience with ADRO shows that most listeners prefer the slower adaptation rate of 3 dB/s and a few prefer the faster rate of 6 dB/s.

Loudness Summation and Loudness Control

A consequence of using many narrow frequency channels to control the loudness of a sound signal is the loudness summation that occurs when many channels are excited simultaneously. In these circumstances, the total loudness can be considerably greater than when only one channel is excited. To compensate for loudness summation, the maximum output level for each narrow channel needs to be below the maximum level that would be applied to a broader frequency channel. As a rule of thumb, when a broader channel is split into two narrower channels, the maximum output level for each of the two narrow channels should be about 3 dB lower than the maximum output level for the broader channel. Similarly, there is loudness summation at lower intensities, and the comfort and audibility targets for narrow channels are generally lower than for broader channels.

The loudness summation effect is used to good advantage by ADRO in the hearing protection and comfort rules. Narrow-band sounds such as high-frequency whistles and tones often have a piercing and unpleasant quality (unless they are part of a musical sound with multiple tonal components that make up a more harmonious sound). Because these sounds usually lie within one processing channel and at most two channels, they will be limited to the maximum output level of the corresponding channel(s). As pointed out in the preceding paragraph, narrow channels have lower output limits than broader channels to allow for loudness summation of broadband signals. Therefore narrowband sounds will be limited to lower levels and will be more comfortable with narrowband processing. The reduction in the maximum output level of narrowband signals is achieved without compromising the overall loudness of broadband signals that are more pleasant to listen to.

Fitting ADRO for Hearing Aids and Cochlear Implants

Hearing aids and cochlear implants alike require fitting adjustment for individual listeners. Conventionally, hearing aids are fitted by prescribing the gain of the hearing aid based on the

listener's audiogram. Conventional fitting of a cochlear implant is based on measurements of threshold and maximum comfortable stimulation levels for each electrode. The fitting of ADRO for hearing aids is similar in concept to the fitting of a cochlear implant in that the output range of each frequency channel is designed to lie within the audible and comfortable range of the listener. For a hearing aid, the audible and comfortable range corresponds to a range of acoustic levels. For a cochlear implant, the audible and comfortable range corresponds to a range of electrical stimulus levels.

When an ADRO device is fitted, the audiologist and listener work together to determine acoustic or electric output levels that produce comfortable and audible sounds. The most direct way to do this is to use *in situ* measures, based on stimuli generated within the device itself.

For cochlear implants, thresholds and maximum comfortable levels are established by using behavioral or objective measures (such as neural response telemetry) for electric pulse trains delivered to one electrode at a time. To streamline the procedure, threshold and comfortable levels may be established on five or more electrodes and interpolated for the remaining electrodes. Often, loudness matching across electrodes is used to check that the original settings and interpolations are acceptable.

For hearing aids, the recommended process is quite similar. Suggested stimuli are narrow-band noises (1/6 octave) at seven audiometric frequencies (500, 750, 1,000, 1,500, 2,000, 3,000, and 4,000 Hz). Initial values for comfortable levels are estimated from the audiogram, and then one sound (usually at 1,000 Hz) is adjusted to a comfortable level according to the listener's reports. The remaining six sounds are matched in loudness to the first sound, and then the initial ADRO fitting is estimated from the matched comfort levels and the audiogram.

ADRO fittings may be predicted from audiometric information alone when behavioral measures are not appropriate, for example, when fitting hearing aids to an infant. Recently, predicted ADRO targets have been compared with desired sensation level input/output (DSL i/o) fittings (Scollie *et al.*, 2004). As a result of this research, ADRO can easily be fitted to achieve the output level goals of the DSL fitting rationale.

The final stages of fitting an ADRO hearing aid or cochlear implant are similar to other fitting

procedures. The listener should be asked whether the loudness of the fitting is appropriate under a variety of conditions that the listener is likely to encounter in everyday life. These include such things as loud and soft speech in quiet, loud speech in background noise, environmental sounds with sudden onset, sustained loud sounds, soft background noises, the sound of the listener's own voice, and high frequency sounds.

When a compression hearing aid is fit, a clear understanding of the interactive effects of compression ratios, kneepoints, gain, time constants, cross-over frequencies, and additional features such as noise reduction is required to fine-tune the fitting. For example, to obtain greater audibility of low-intensity sounds, one may increase the gain in each channel. Changing the gain for soft sounds will also increase the loudness of high-intensity sounds unless the compression ratio is increased in each channel. The amount of increase in the compression ratio required to keep intense sounds comfortable depends on the kneepoint. With ADRO, the audibility of low-intensity sounds can also be boosted by increasing the maximum gain values. This change will allow lower level sounds to reach the audibility target. In contrast with compression schemes, no other ADRO parameters need to be changed to keep high-intensity sounds comfortable, because the comfort and hearing protection rules operate independently of the other rules in each channel.

These examples illustrate the independent nature of the ADRO processing rules and the intuitive nature of the fitting process.

Bimodal and Hybrid Fittings for Hearing Aids and Cochlear Implants

One of the most exciting fields of research in audiology today is the treatment of severe hearing impairment. As the performance of hearing aids and cochlear implants improves, there is effectively a competition for the population with severe hearing impairment. Several studies have compared speech perception and other outcomes with hearing aids and cochlear implants for adults (Flynn *et al.*, 1998) and children (Boothroyd and Oran, 1997; Blamey *et al.*, 2001a; Blamey and Sarant, 2002) in this group. The conclusions indicate that on average, children and adults with cochlear implants perform more like people with

severe hearing loss than people with profound hearing impairment.

For people with a severe hearing impairment, the choice between a cochlear implant and a hearing aid can be difficult, but there are alternatives. For example, a "bimodal" fitting uses a hearing aid in one ear and a cochlear implant in the other, while a "hybrid" fitting uses a hearing aid and a cochlear implant together in a single ear. Both bimodal and hybrid fittings show promise.

It is well established that a hearing aid and a cochlear implant together can provide improved speech perception compared with either device on its own (Armstrong *et al.*, 1997; Dooley *et al.*, 1993; Ching *et al.*, 2004). The combination of devices is particularly effective in background noise (Armstrong *et al.*, 1997). The published results for bimodal speech perception have all been obtained under controlled conditions that provide approximately balanced loudness from the two devices. A listener in these conditions has access to information from both ears, and the information will be combined with optimum effect (Blamey *et al.*, 2001b).

One of the challenges in using acoustic and electric stimuli together is the difference in the dynamic range of the acoustic and electric stimuli. Blamey *et al.* (2000) showed that conventional hearing aid and cochlear implant amplification schemes and fitting procedures often resulted in large mismatches in loudness between the ears of bimodal listeners. Under these circumstances, the optimal combination of information from the two signals is unlikely to occur over a wide range of input levels because one signal will be much louder than the other. Thus, it is important that both the hearing aid and the cochlear implant provide audible and comfortable outputs corresponding to a wide range of input levels. To facilitate the combination of the two signals, it is also important that the temporal fluctuations of the amplitude envelope be similar in the two ears so that sounds are perceived as one rather than two streams of information.

Research by Ching *et al.* (2004) indicates that bimodal speech perception scores can be improved if the hearing aid is fitted taking into account the cochlear implant, rather than in the manner that would normally be prescribed for the hearing aid alone. If both devices are fitted to comfort levels with the ADRO strategy, then they will automatically produce compatible signals

that are reasonably closely matched for loudness across frequency and input dynamic range. The most information-rich part of the signal will be presented to both ears in the most effective part of the hearing range.

Similar considerations apply to hybrid stimulation where the low-frequency acoustic components need to be compatible with the electric components.

The ideal fitting software for bimodal and hybrid ADRO fittings would allow comfortable levels to be established and balanced for both acoustic and electric components in a single procedure, followed by simultaneous adjustment of the implant(s) and hearing aid(s) for both ears. The ideal fitting software does not yet exist, but the similarity of the ADRO processing and fitting procedures for the two devices makes it a possibility for the future.

Clinical Evaluation of ADRO in Cochlear Implants and Hearing Aids

Although ADRO was initially developed for bimodal listening, it was first implemented and tested separately in two cochlear implant studies (Dawson *et al.*, 2004; James *et al.*, 2002) and three hearing aid studies (Martin *et al.*, 2001a, 2001b; Blamey *et al.*, 2004a, 2005). Although the five studies used a diverse range of subject groups, devices, and comparison conditions, they showed a consistent pattern of benefits in the ADRO condition. The studies are summarized in Table 1, and the interested reader is referred to the original publications for further details.

An interesting feature of this set of studies is that they were all conducted with the ADRO and non-ADRO processing on the same hardware platform. The cochlear implant studies both used the

Table 1. Summary of Experimental Conditions and Results in Published ADRO Studies

Study	No of Subjects	Device Hardware	Comparison Amplification Scheme	Speech in Quiet at Low Input Level	Speech in Noise	Preferences Assessed by Questionnaire
James <i>et al.</i> , 2002	9 adults	SPRINT CI	ACE and SPEAK	16% improvement	No significant difference	Eight subjects preferred ADRO
Dawson <i>et al.</i> , 2004	15 children	SPRINT CI	ACE	8.6% improvement	6.9% improvement	11 children preferred ADRO in most situations
Martin <i>et al.</i> , 2001a, 2001b	15 adults with moderate to profound loss	Laboratory HA	Linear NAL-RP	36.4% improvement	7.0% improvement	
Blamey <i>et al.</i> , 2005	19 adults with moderate to profound loss	BTE HA	3-channel compression NAL-NL1	14.2% improvement	7.3% improvement	ADRO preferred in 74% of situations
Blamey <i>et al.</i> , 2004a	22 adults with mild to moderate loss	ITE HA	9-channel compression NAL-NL1	7.9% improvement	7.3% improvement	

SPRINT processor, in which ADRO was an option that could be switched on or off. The hearing aid studies used three different instruments (a laboratory processor, a behind-the-ear, and an in-the-ear aid), but in each case, it was possible to implement ADRO and the alternative amplification scheme on the same hardware. In all cases except for the laboratory study with linear hearing aid, the subjects were given at least 4 weeks of acclimatization experience in each condition before evaluation. In most cases, the evaluation was blind so that subjects were not aware which was the ADRO condition. None of the studies produced any result that was statistically significant in favor of the non-ADRO device.

A common question asked about the study results is whether it is the fitting procedure or the ADRO processing that accounts for the benefits observed. In the implant case, the answer is easy—the same fitting was used for both ADRO and non-ADRO conditions, so the processing accounts for the improvements. It is not so easy to answer this question for hearing aids on the basis of the published results, but two additional (unpublished) studies have now been completed. These studies attempted to match the long-term average output levels of the ADRO hearing aid using wide dynamic range compression for speech at two input levels.

In one study, the fitting software did not allow compression ratios higher than 3.3. A good matching of output levels was achieved for speech at the 50-dB SPL input level. The compression scheme was rated as uncomfortably loud by nine of ten subjects for speech at a 70-dB input level even though the maximum compression ratio was used. All ten subjects in the study scored better on the hearing in noise test (HINT) with ADRO, with a mean advantage of a 1.8-dB signal-to-noise ratio.

In the other study, the output levels were not successfully matched, and the subjects rated ADRO significantly better than the “matched” compression scheme in all conditions. Hearing instrument test box measures indicated that ADRO gave a significant audibility advantage in this second study.

These two studies show that ADRO processing is more effective at providing both audibility and comfort than conventional compression schemes in hearing aids unless very high compression ratios are used. When fittings were matched as closely as possible, ADRO still provided a significant advantage in background noise.

The published and unpublished data indicate that ADRO has significant advantages over compression. The improvement in intelligibility in background noise appears to be independent of the fitting used for the alternative device. The advantage has been demonstrated in cochlear implants with matched fittings, and in hearing aids in comparison with linear NAL-RP, 3-channel NAL-NL1, 9-channel NAL-NL1, and with compression using “matched” fittings. It seems obvious that compression (or linear) amplification could be used to match the audibility improvement for soft speech in quiet by an appropriate volume setting, but unless compression ratios are higher than 3.3, comfort will be compromised at this volume. Similarly, equal comfort for loud sounds could be achieved with compression by using an appropriate volume adjustment, but then audibility would be compromised unless compression ratios were higher than 3.3. In summary, wide dynamic range compression can match ADRO on either audibility or comfort but not both unless very high compression ratios are used. High compression ratios are likely to reduce speech intelligibility in background noise and adversely affect sound quality in quiet.

Summary

The ADRO processing strategy is the next step beyond compression for both hearing aids and cochlear implants. The statistical rules that control the ADRO process are able to optimize the audibility, comfort and intelligibility of the signals without compromising sound quality. To achieve the same results with compression would require high compression ratios that are known to reduce sound quality. ADRO also has the flexibility required to improve hearing for people with all degrees of hearing loss, including normal-hearing people using telephones and headsets, hard-of-hearing people using hearing aids, and deaf people using cochlear implants. The fitting of an ADRO processor for an individual user and its adaptation to a particular purpose or environment is straightforward because of the intuitive nature of the ADRO rules and fitting parameters. The advantages and the versatility of ADRO processing have shown robust performance benefits in

clinical trials for hearing aids and cochlear implants, and studies are in progress to show similar benefits in bimodal stimulation and for headsets in call centers.

Acknowledgment

The ADRO research and development reported here is the product of many years work by many people. It has been my privilege to work with talented people, and I sincerely appreciate their help and encouragement. Those who have been most closely involved with ADRO include Lois Martin, Chris James, Brett Swanson, Konrad Wildi, David Macfarlane, Elaine Saunders, and Hayley Fiket. I also greatly appreciate the time and insights freely given by the hundred or more listeners who have helped with the development and evaluation of ADRO. Privacy laws prevent me from naming you here, but your personal contributions have been great.

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