Hearing-aid-processed Tone Pips: Electroacoustic and ABR Characteristics

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Abstract

The auditor brainstem response (ABR) recorded while wearing a hearing aid may supply supplemental information about the benefit and appropriateness of the hearing aid for certain infants. The purposes of this study were (1) to determine the effects of different output limiting circuits on the acoustics of tone-pip stimuli used for ABR recordings and (2) assess how changes in hearing-aid-processed stimuli affect ABR characteristics. Electroacoustic input/output functions to tone-pip stimuli were constructed for three different output limiting circuits (wide dynamic range compression, output compression, and linear with peak clipping) available in a programmable hearing aid. Wave V latency and amplitude functions were then measured to the same stimuli and hearing aid settings in five normal-hearing adults. Electroacoustic results showed that none of the output limiting circuits, including linear peak clipping, were effectively activated by tone pips compared to the hearing aid performance to continuous tones. Aided wave V latency and amplitude functions were asymptotic to high stimulus levels, suggesting that cochlear output was in saturation.

Key Words: Auditory brainstem response, hearing aids

Abbreviations: ABR = auditory brainstem response, I/O = input/output, MPO = maximum power output, OAE = otoacoustic emission, pe SPL = peak equivalent sound pressure level, SC = super compression, sPC = soft peak clipping, WDRC = wide dynamic range compression

As a result of the trend toward universal hearing screening of all newborns, there is an increased number of hearing-impaired infants identified and fit with hearing aids by 6 months of age or sooner. In current practice, infant hearing thresholds are primarily based on auditory brainstem response (ABR) results and supplemented with soundfield measurements when possible. When amplification is indicated, the appropriate hearing aid gain may be determined by soundfield aided thresholds, real-ear measurements, or, more recently, the predicted real-ear aided response based upon 2-cm³ coupler measurements (Moodie et al, 1994; Seewald et al, 1996).

Infants whose hearing thresholds cannot be established because there are no observable ABRs, otoacoustic emissions (OAEs), or behavioral responses undergo a less structured hearing aid fitting procedure. The hearing loss is presumed to be severe or profound and a high-power hearing aid is selected. If there is questionable benefit from the aid, a “watchful waiting” approach is taken consisting of periodic repeated testing and enlisting the care giver to report if the child adopts any new reactions to environmental sounds or increases vocalizations. This trial period usually proceeds from several months to years and if no benefits are seen, alternative habilitative actions such as consideration for a cochlear implant may be appropriate. A recordable ABR while the infant is wearing a hearing aid would be an important contraindicator of anacusis and could be helpful in planning the habilitative strategy.
Other infants difficult to manage are those with multiple problems precluding reliable behavioral testing but with recordable ABRs for threshold establishment. After wearing a hearing aid for many months, they seem to be deriving no benefit from the aid in either their response to sound or by positive changes in their vocalizing pattern. The ABR recorded while aided would be an important confirmation that the hearing aid is, in fact, lowering auditory thresholds as predicted.

Aided ABR assessment has been investigated as a method for infant hearing aid fitting, ABR criteria used to judge the hearing aid fitting include wave V latencies and/or thresholds (Kileny, 1982; Hecox, 1983; Mahoney, 1985). Studies have indicated that the brief stimulus used for ABR recordings may interact with the hearing aid electronics and earmold characteristics, producing unexpected effects on wave V latency and amplitude. For example, Mahoney (1985) demonstrated that wave V latency can actually increase, rather than decrease, as hearing aid gain is increased. Also, there appears to be an interactive effect on wave V latency between stimulus repetition rate, hearing aid gain, and compression characteristics that must be taken into account when using wave V latency for a fitting criterion (Gerling, 1991). Unaided and aided ABR soundfield thresholds to broadband clicks were measured by Beauchaine et al (1986) to assess the relation between ABR-based estimates of hearing aid gain and traditional, functional gain measurements. The decibel difference between aided and unaided wave V latency functions compared closely with functional gain averaged across several frequencies; however, these results were based on only four hearing-impaired subjects and frequency-specific tone-pip stimuli for ABR measurements were not used. Furthermore, the ABR stimulus level and repetition rate were shown to have a significant effect on hearing aid compression characteristics, suggesting that the brief tone-pip stimulus may not effectively activate the compression circuitry (Gorga et al, 1987).

While some question the usefulness of ABR measurements recorded while wearing a hearing aid (Bergman et al, 1992), the ABR still remains the only practical means to estimate hearing thresholds when behavioral responses are unavailable. As such, a better understanding of the technical aspects of aided ABR measurements is still warranted to better apply this technique for those circumstances when additional information is needed. This experiment is divided into two parts: Part I examines the electroacoustic effects of different hearing aid output limiting circuits on tone-pip stimuli typically used for ABR recordings; Part II is directed at assessing the effects of the different hearing aid output limiting circuits on ABR waveform characteristics.

**METHOD**

**Part I: Hearing Aid Electroacoustic Analysis**

The Biologic Navigator evoked response system was used to generate continuous-tone and tone-pip stimuli that were transduced by a loudspeaker (Realistic, Minimus). The loudspeaker was always positioned 7.6 cm away from the sound level meter, the hearing aid, or the subjects’ ears (90° azimuth, re: nose). This distance required a 0.23-msec travel time for the sound to reach the ear and was not subtracted from the wave V latencies. Tone pips at center frequencies at 1, 2, and 4 kHz were presented at a rate of 59/sec. This rate was chosen for two reasons. First, rapid data collection is essential in order to complete a complicated aided/unaided ABR clinical assessment in sedated infants, the population to which this procedure would be applied. A fast stimulus repetition rate facilitates data collection and does not adversely affect response identification (Klein et al, 1992). Second, a hearing aid’s response characteristics to transient stimuli are most similar to its behavior to long-duration tones at faster transient repetition rates (Gorga et al, 1987). Thus, the fastest feasible repetition rates were used to best activate the designed characteristics of the hearing aid. The rise-plateau-fall times were set at 2-0-2 msec for 1 kHz, 1-0-1 msec for 2 kHz, and .5-.5-.5 msec for 4-kHz signals. The system was calibrated using a sound level meter (B&K 2203), spectrum analyzer (HP 3561 Dynamic Signal Analyzer), and oscilloscope (Tektronix 2215). The tone-pip levels were converted to peak-equivalent SPL (pe SPL) by matching a continuous tone to the peak-to-peak amplitude of the tone pip and measuring SPL on the sound level meter.

The hearing aid was attached to a 2-cm³ coupler (B&K 4152 Artificial Ear), which was attached to the sound level meter. The output from the sound level meter was routed to the oscilloscope and spectrum analyzer. All measurements were made in a sound-treated chamber.

A Phonak Piconet2 P2AZ programmable behind-the-ear hearing aid was selected for this
study. This hearing aid system was chosen primarily for the range of output limiting options available. The Piconet2 features five output-limiting options: wide dynamic range compression (WDRC), super compression (SC), SC with adaptive release time (SC+aRT), soft peak clipping (sPC), and peak clipping (PC). The PiCS Phonak Reference Manual describes the output limiting options as follows: the WDRC circuit has a fixed compression threshold at 45 dB SPL input level and a variable compression ratio dictated by the gain and maximum power output (MPO) settings. The circuit decreases amplifier gain as input increases. The SC and SC+aRT circuits are output limiting compression. Output compression provides linear amplification over a wide range of input levels and then compresses (reduces amplifier gain) severely at an 8:1 ratio at the threshold of compression that is near the output saturation point. SC+aRT differs from SC by providing variable release times for long- and short-duration signals. PC circuits amplify linearly until the point of saturation is reached. Then the output is limited by "clipping" voltage peaks rather than reducing amplifier gain. PC circuits, therefore, are more likely to cause distortion of the input signal. The manufacturer claims that the sPC circuits reduce some of the distortion seen in traditional peak clipping circuits.

In a pilot study, stimuli spectra and waveforms were compared at each of the five output limiting options. Three limiting options, WDRC, SC, sPC, were selected to study the widest variety in signal limiting characteristics. WDRC showed the least distortion of signal. SC and SC+aRT were noted to be quite similar, and SC was empirically chosen for testing. PC and sPC were similar; however, unexpectedly, sPC showed the most distortion of the signal, and this mode was chosen to offer more diversity. The attack/release times measured for the different circuits were 2/40 msec (WDRC), 3/50 msec (SC), and <1 msec (sPC). This is consistent with the manufacturer's reported specifications of 5/30 msec.

The hearing aid was set and programmed during these experiments with Noah software and a HI-PRO interface box. The aid was fixed at a gain of 16 (32 maximum) arbitrary units and an MPO of 1 (5 maximum) arbitrary units. The 16 units produced about 20 dB of gain, and an MPO of 1 produced a peak saturation SPL of about 105 dB. This minimal output setting was used to prevent acoustic trauma in the normal hearing subjects. Low-, mid-, and high-frequency channels were set for minimal filter effect.

The hearing aid was analyzed to ensure that it was performing within manufacturer's specifications. Hearing aid performance was measured to continuous tones and tone pips presented at 40 to 100 dB SPL in 10-dB increments. Values were obtained at 1, 2, and 4 kHz using all three output limiting methods. The tone pips were presented first through the 2-cc coupler alone to note possible coupler effects on the stimuli. Then the aid was connected and the same series of stimuli were presented. The electroacoustics of the hearing-aid-processed tone pips for each output limiting method were evaluated for distortion of the time waveform, differences in spectra, and the effectiveness of output limiting relative to the continuous tones of the same frequency. Analysis of the time waveform revealed that the hearing aid transduction process consistently introduced a 0.4-msec delay. This factor was subtracted from wave V latency in all aided ABRs in Part II.

Part II: Unaided and Aided ABR Recordings

Five normal-hearing adults (five ears) aged 25 to 30 served as subjects. Hearing normalcy (≤20 dB HL) (ANSI, 1989) was determined by pure-tone audiometry and otoscopic examination of the ear.

Tone-pip stimuli, as described in Part I, were used for ABR recordings. Mean behavioral thresholds for the five subjects to the tone pips presented in sound field (90° azimuth) were 26, 22.8, and 20 dB pe SPL at 1, 2, and 4 kHz, respectively.

The same hearing aid described in Part I was used.

The evoked response system used to generate the stimuli was also used for the ABR recordings. ABRs were recorded with electrodes on each mastoid and on the forehead. Interelectrode impedance was less than 5 kohms. The ABR analysis time window was set at 16.9 msec. The raw EEGs were filtered at 30 to 1500 Hz for 1-kHz tone-pip data collection and at 100 to 3000 Hz for the 2- and 4-kHz tone-pip data collection. Different EEG filter settings were used for the 1-kHz stimuli to enhance wave V amplitude to low-frequency stimuli (Hayes and Jerger, 1982; Stapells, 1989). One thousand stimuli were averaged for each ABR recording.

The subject was in a recumbent position throughout testing. The contralateral ear was fitted with a foam noise-reduction plug to ensure monaural recordings. Recordings were repeated
three to five times to low-level stimuli to help identify replicable wave V peaks. The averaged latencies and amplitudes of the repeated trials were used for data analysis. Data collection was performed over two sessions of about 2 hours each. In session one, behavioral soundfield thresholds were determined and unaided ABRs were recorded to 1-, 2-, and 4-kHz stimuli presented at 40, 60, 80, 90, and 100 dB pe SPL. In the second session, aided ABRs were recorded for the three output limiting options at these same stimuli input levels.

RESULTS

Hearing Aid Electroacoustics

Input/output (I/O) data for continuous tones and tone pips at 1, 2, and 4 kHz for each of the three hearing aid output limiting circuits are shown in Figure 1. One point that this figure illustrates is that the maximum hearing aid output to tone pips always exceeds the output to continuous tones regardless of limiting circuitry. The second point is that the I/O functions for continuous tones and tone pips differ. The I/O functions to tones for SC and sPC are similarly linear until saturation output is approached while the I/O function for WDRC is highly nonlinear across a wide range of input levels. This behavior is in accordance with the manufacturer's described functions for these circuits. In contrast, the I/O functions to tone pips are all more linear and do not substantially differ between limiting circuits. Two slight trends are present in the I/O functions for the tone pips. There is a trend for the output to the WDRC to be lower for 90 and 100 dB SPL inputs for the 1-kHz signal and a second trend for the output for the sPC at 4 kHz to be lower at the 90 dB SPL input. The unique characteristic of the WDRC function was not activated by the tone-pip stimuli used in this study.

The electroacoustic responses of the hearing aid were further examined by observations of spectra and waveforms from the 2-cc coupler output. At low input levels, the waveforms for all output limiting circuits were similar, showing only slight distortion caused by ringing in the 2-cc coupler. However, at the higher input levels, the various output limiting circuits clearly produced different effects. Representative data for the 100 dB SPL inputs for the 1-, 2-, and 4-kHz tone pips are shown in Figures 2, 3, and 4, respectively. The top spectrum and waveform, displaying the WDRC, indicate a slight ringing of the waveforms. This ringing is primarily caused by the 2-cc coupler. The middle panels, displaying data for the SC circuit, show that more distortion has been introduced into the waveforms with visible ringing extending to about 6-msec post-stimulus onset. Accordingly, the side lobes of the spectra are not as reduced in relation to the primary energy peak. The bottom panels, displaying the sPC data, indicate substantial waveform distortion with ringing extending to 8 msec or more and numerous harmonics are prominent in the spectra. The least amount of ringing occurs for the 1-kHz signal, while the most ringing occurs for the 4-kHz signal.

Hearing Aid Effects on the ABR

Latency and amplitude functions for wave V were constructed for the unaided condition and
Figure 2  Waveform (upper) and corresponding fast Fourier transforms (FFT) (lower) pairs for the 100 dB peSPL, 1-kHz signal for each output limiting circuit. On the time waveform panel, vertical gridline scale is 4 msec/division. On the FFTs, vertical gridlines indicate 1 through 7 kHz, respectively. The full-scale vertical range on the FFTs is 80 dB.

For each of the three hearing aid output limiting conditions. Due to the small number of subjects, statistical testing of the differences between means was not appropriate. The latency functions shown in Figure 5 vary according to tone-pip frequency and recording condition. Unaided latencies to the 1-kHz pips are shorter than all aided conditions at the higher stimulus levels.

Figure 3  Waveform (upper) and corresponding FFT (lower) pairs for the 100 dB peSPL, 2-kHz signal for each output limiting circuit. See legend for Figure 2.

For the 2- and 4-kHz pips, all latencies are essentially equal at the highest level except at 4 kHz where the latencies in the sPC condition tend to be longest. At lower stimuli levels, the latencies in the unaided condition are consistently longer compared to the aided conditions.

Amplitude data in Figure 6 are characteristically more variable. There is a trend for lower amplitude wave Vs in the unaided condition for most input levels. However, at the highest stimulus level there is little or no difference in amplitude across all recording conditions.
DISCUSSION

Several points were demonstrated in this study. First, the brief-duration stimuli used for ABR recordings did not effectively activate any of the output limiting circuitry options examined in this study. This same observation was made by Gorga et al (1987) using older-technology hearing aids. It apparently still holds true for, at least, the current high-technology analog hearing aid used in this study. Surprisingly, this was the case for the simple, linear, peak clipping circuit with <1-msec attack time, as well as for the more sophisticated compression circuits. This suggests that attack time alone is not the reason for differences in hearing aid performance to brief stimuli as compared to long-duration stimuli. The present data indicate that hearing aid performance to longer duration stimuli, such as noise or tones used for real-ear, electroacoustic coupler, or soundfield behavioral threshold measurements, will probably not be accurately represented by brief tone pips, particularly at high-intensity levels. The error will be in the direction of overestimating the available gain/output performance of the hearing aid for high-intensity stimuli. The above findings may only apply to the hearing aids assessed by Gorga et al (1987) and to the one assessed here. Study of newly developed digital hearing aids might be performed to assess their performance with transient signals.

The second main point of this study was to assess the effects of stimulus distortion and compression introduced by the hearing aid output limiting circuitry on ABR parameters. The ABR latency and amplitude functions demonstrated some differences as a function of input level and tone-pip frequency compared to the unaided functions. The trend for shorter latencies and higher amplitudes in the aided conditions generally seen below the 90 dB SPL input level is attributed to the higher intensity level of the signal as a result of amplification from the hearing aid. These findings were expected and were similar to the results of Beauchaine et al (1986), who used a similar paradigm.

Although the acoustic output levels shown in Figure 1 were always greater for the hearing-aid-processed tone-pip signals compared to the speaker-only condition for a given input level, this was not reflected in the latency or amplitude data at the higher input levels. In other words, we would expect that ABR latencies would be shorter and amplitudes larger in all of the aided conditions compared to the unaided condition at a given input level. Rather, the amplitude and latency data were either the same, or, in the case of the 1-kHz data, latencies were actually shorter in the unaided versus aided conditions. This disparity between the higher acoustic output levels and the reduced physiologic output functions is not straightforward and is perhaps due to a combination of several factors. One possible reason is that the extensive ringing of the stimulus produced by the hearing aid effectively
reduces the interstimulus interval, causing forward masking or adaptation of the ABR. Nevertheless, the ABR waveforms demonstrated good resiliency to distorted stimuli, particularly in the sPC output limiting mode.

A second possible source of decreased physiologic output could involve attenuation of the stimulus by stapedial reflex contraction. The stapedial reflex, however, should have little effect on the 4-kHz stimulus.

Another possible, and perhaps more likely, reason for the lack of increased physiologic output is that the cochlear (neural) output was approaching saturation at high stimuli inputs. There is evidence for this even in the unaided condition, as suggested by some of the latency and amplitude data in Figures 5 and 6. Therefore, the additional acoustic gain provided by the hearing aid could not be accurately represented by changes in physiologic parameters because there were few additional neural elements available to contribute to the gross ABR. This raises an important question: can an ABR be recorded in someone with a severe to profound hearing loss by simply increasing the level of the stimulus through a hearing aid? While a high-gain hearing aid can lower the behavioral thresholds of a severely hearing-impaired individual by 60 dB or more, it is not yet known if an ABR is recordable under the same circumstances.

In summary, hearing aids will not necessarily perform as designed or in accordance with the ANSI testing standard (based on long-duration stimuli) to brief tone-pip stimuli. This fact complicates the problem of using ABR measurements for hearing aid assessments. Sub-
stantial pilot data are needed to define the idiosyncracies of a specific aid’s response to the ABR stimuli prior to the clinical evaluation. Another complication identified in this study was the apparent saturation of ABR parameters at high stimulus levels in normal-hearing subjects. Aided ABR data on subjects with severe/profound sensorineural hearing impairment are needed to determine the limits of physiologic measurements in this population. Finally, the present data were collected from normal-hearing adult subjects. The ABR may not be as resilient to distortion of the stimulus and/or may saturate in a different pattern in individuals with hearing loss or in infants. There is clearly a need for much additional data on the aided and unaided ABR in hearing-impaired adult and infant populations before the actual clinical usefulness of the aided ABR can be determined.

REFERENCES


