SLOW CORTICAL POTENTIAL MEASURES OF AMPLIFICATION

by

SUSAN LINDSAY MARYNEWICH

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ABSTRACT

With the advent of Universal Newborn Hearing Screening programs, it has become increasingly common for infants to be fit with hearing aids by six months of age. Due to the inability of infants to actively participate in the hearing aid fitting, there is a need for a reliable and objective measure of hearing aid validation in this population. Slow cortical potentials (SCP) are currently being marketed for the purpose of validating infant hearing aid fittings; however, there is a lack of evidence to support use of SCPs for this purpose. In the current thesis, two studies were carried out: Study A investigated N1-P2 amplitudes and N1 latencies in response to a 60-ms duration tonal stimulus (1000 Hz) presented at three intensities (30, 50 and 70 dB SPL) in aided and unaided conditions using three hearing aids (Analog, DigitalA, DigitalB) with two gain settings (20- and 40-dB). Study B investigated the effects of hearing aid processing on acoustic measures of the stimuli, under the same conditions as Study A, with an additional 757-ms tonal stimulus (1000 Hz). Overall, it was predicted that N1-P2 amplitudes would be larger and N1 latencies shorter in the aided compared with unaided conditions; however, the results showed response amplitudes were smaller for the digital hearing aids compared with the analog hearing aid and none of the hearing aids resulted in a reliable increase in response amplitude relative to unaided across conditions. Additionally, N1 response latencies in analog conditions were not significantly different from unaided N1 latencies; however, both digital hearing aids resulted in significantly delayed N1 peaks. Acoustic recording results obtained in Study B indicate that gain achieved by the hearing aids for both the short and long SCP stimuli was less than real-ear insertion gain measured with standard hearing aid test signals. The effect was more pronounced for the short stimulus. These results suggest that the typical stimuli used for SCP testing may be too brief for the processing time of hearing aids, especially those with digital processing.
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<td>Auditory brainstem response</td>
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<tr>
<td>A/D</td>
<td>Analog-to-digital</td>
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<td>AEP</td>
<td>Auditory evoked potential</td>
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<tr>
<td>ANOVA</td>
<td>Analysis of variance</td>
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<td>ANSD</td>
<td>Auditory Neuropathy Spectrum Disorder</td>
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<tr>
<td>ASSR</td>
<td>Auditory steady state response</td>
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<tr>
<td>CAEP</td>
<td>Cortical auditory evoked potential</td>
</tr>
<tr>
<td>CL</td>
<td>Compression limiting</td>
</tr>
<tr>
<td>DSL [i/o]</td>
<td>Desired Sensation Level, input-output prescription formula</td>
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<tr>
<td>EEG</td>
<td>Electroencephalogram</td>
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<tr>
<td>EOG</td>
<td>Electro-oculogram</td>
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<tr>
<td>HL</td>
<td>Hearing level</td>
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<tr>
<td>IHC</td>
<td>Inner hair cell</td>
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<tr>
<td>ISI</td>
<td>Interstimulus interval</td>
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<tr>
<td>L-I function</td>
<td>Latency-intensity function</td>
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<tr>
<td>MPO</td>
<td>Maximum power output</td>
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<tr>
<td>NAL-NL1</td>
<td>National Acoustics Laboratory’s non-linear prescription formula</td>
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<tr>
<td>OHC</td>
<td>Outer hair cell</td>
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<tr>
<td>PC</td>
<td>Peak clipping</td>
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<tr>
<td>PEACH</td>
<td>Parent’s Evaluation of Aural/Oral Performance in Children</td>
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<tr>
<td>ppeSPL</td>
<td>Peak-to-peak equivalent sound pressure level</td>
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<td>REAG</td>
<td>Real-ear aided gain</td>
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<td>Abbreviation</td>
<td>Description</td>
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<td>--------------</td>
<td>--------------------------------------------</td>
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<tr>
<td>REAR</td>
<td>Real-ear aided response</td>
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<td>Real-ear unaided response</td>
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<td>RMS</td>
<td>Root-mean-square</td>
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<td>SCP</td>
<td>Slow cortical potential</td>
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<tr>
<td>SNR</td>
<td>Signal-to-noise ratio</td>
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<td>SPL</td>
<td>Sound pressure level</td>
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<td>SPT</td>
<td>Swept pure tone</td>
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<tr>
<td>SWCN</td>
<td>Speech-weighted composite noise</td>
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Chapter 1

Literature Review: Hearing Aids and Objective Validation Methods
INTRODUCTION

With the advent of Universal Newborn Hearing Screening programs, it has become increasingly common for infants to be diagnosed with hearing loss and fit with hearing aids by six months of age, which requires that reliable methods are in place for hearing aid fitting in this young population (Ching & Dillon, 2003; Joint Committee on Infant Hearing, 2007 Seewald & Scollie, 2003; Tharpe, Sladen, Huta, & Rothpletz, 2001; Yoshinaga-Itano, 2003). The hearing aid fitting involves several stages, which include: assessment of hearing loss, and selection, fitting, verification, and validation of the hearing aid (Joint Committee on Infant Hearing, 2000; Scollie & Seewald, 2001; Valente, Abrams, Benson, Chisolm, Citron, Hampton, et al., 2006). Test procedures used in the hearing aid fitting fall into two general categories: (i) behavioural measures requiring active participation by the patient (e.g., pure tone and speech testing) and (ii) objective measures, which require no feedback from the patient (e.g., real-ear electroacoustic measures). Due to the inability of infants to actively participate in the hearing aid fitting, objective methods are required across all stages of the fitting; however, there is a lack of valid objective validation measures. For this purpose, auditory evoked potentials (AEP) such as the Auditory Brainstem Response (ABR), the Auditory Steady-State Response (ASSR) and the Slow Cortical Potential (SCP) have been considered and varying degrees of success have been reported.

The goal of the following discussion is to shed light on why further research is needed on SCPs for the purpose of hearing aid validation. At the outset, hearing aid settings and measurement techniques that are relevant to methods used in the current study will be discussed. Next, the effects of stimulus choice for the purpose of electroacoustic measures will be discussed. Of primary importance to the current study is the effect of different stimuli on
hearing aids, particularly those with digital processing. These stimulus effects may help to explain why different AEPs have found limited success for the purpose of hearing aid evaluation. Of particular interest is the SCP, about which there is conflicting evidence in the literature regarding its validity for measuring hearing aids.

HEARING AIDS

Adjustable Hearing Aid Settings

When selecting and fitting a hearing aid, there are choices that must be made regarding hearing aid settings that may ultimately affect the measurement of hearing aids using certain stimuli. Of particular interest for the current thesis research are the effects of amplification strategies (e.g., linear vs. non-linear) found in both analog and digital hearing aids and the processing features (e.g., noise reduction and feedback management) specific to digital hearing aids.

Amplification Strategies

The defining characteristic of linear amplification is that equivalent gain is provided across a wide variety of input levels for any particular frequency. In other words, as input is increased, there is an equivalent increase in output. Until maximum power output (MPO) is reached, the gain provided by a linear amplifier for any particular frequency is not dependent on the number of signals being amplified simultaneously; therefore, when a hearing aid is operating in its linear phase, the frequency-gain response should be the same regardless of the stimulus used (Frye, 1987; Kuk & Ludvigsen, 2003). In support of this, Stelmachowicz, Lewis, Seewald, & Hawkins (1990) described a case of linear amplification in which a swept pure tone (SPT)
stimulus was compared with a speech-weighted broadband signal. They found no significant difference between the frequency-gain responses of the two stimuli when the pure-tone level and root-mean-square (RMS) level (i.e., calculation of the average decibel level of a stimulus with variable amplitude) were equivalent at 60 dB SPL.

The suitability of linear amplification, however, is dependent on the type and degree of hearing loss. For example, cochlear hearing loss typically results in damage to outer hair cells (OHC) and in cases of severe or profound hearing loss damage to inner hair cells (IHC) is also common (Moore, Vickers, Plack, & Oxenham, 1999; Patuzzi, Yates & Johnstone, 1989a, 1989b). The effects of OHC damage are loss of the active mechanism and non-linear properties of the cochlear amplifier (Moore, 1996; Moore et al., 1999; Patuzzi et al., 1989a, 1989b), resulting in a loss of sensitivity for soft sounds, loss of natural compression, and a broadening of auditory tuning curves (Moore, 1996; Moore et al., 1999; Patuzzi et al., 1989a, 1989b). In consequence, the thresholds for detecting sounds are higher and loudness perception grows more rapidly in individuals with cochlear hearing loss (Moore, 1996; Steinberg & Gardner, 1937). This difference between an individual’s threshold and loudness discomfort levels is referred to as their dynamic range and it is difficult to provide individuals who have reduced dynamic ranges with speech that is audible without exceeding loudness discomfort (Dillon & Storey, 1998; Moore, 1996, 2004, 2007; Moore et al., 1999; Souza, 2002; Stone & Moore, 1997). In some cases of profound hearing loss, the dynamic range can be as small as 10 dB (Souza, 2002; Stone & Moore, 1997). Non-linear (i.e., compression) amplification allows gain to be adjusted independently across low, medium and high input levels so that soft sounds can be amplified to a comfortable level and loud sounds can be compressed so as not to exceed loudness discomfort (Dillon & Storey, 1998; Kuk, 1996; Moore, 2008; Souza, 2002; Souza, Jenstad, & Folino, 2005;
Stone & Moore, 1997). This is accomplished through the adjustment of several compression parameters including the ratio, threshold, and attack and release times.

The compression ratio determines how much gain reduction will occur above the compression threshold. For example, a 2:1 compression ratio indicates that the hearing aid output (in dB) will only increase in level by half as much as the input level (in dB); thus, for every 2-dB increase in input level, the output level will increase by 1 dB. The compression ratio can be increased so that a greater range of input levels can be compressed into the dynamic range of the hearing aid wearer.

Compression threshold refers to the lowest input level at which gain reduction is applied (Souza, 2002). Below this level, linear gain is often applied; however, in some cases hearing aids may apply expansion (Souza, 2002). Expansion functions in a complementary way to compression, such that in order to reduce circuit and low-level environmental noise, less gain is applied to signals as input level decreases (Bentler & Chiou, 2006; Bray & Ghent, 2001; Ghent, Nilsson & Bray, 2000; Souza, 2002).

Compression attack time refers to the speed at which gain for high-level signals is reduced (Moore, 2008; Souza, 2002), and release time refers to the speed at which the hearing aid recovers from compression (i.e., returns to linear processing) following a period of lower intensity input (Moore, 2008; Souza, 2002). The attack time is typically fixed and short (i.e., <5 ms), to reduce the likelihood of sounds exceeding the client’s discomfort levels (Hickson, 1994; Moore, 2008; Moore & Glasberg, 1988; Souza, 2002; Stone & Moore, 1992). Release time, on the other hand, can vary between a few milliseconds to several seconds in duration and clinicians
have the option of choosing a hearing aid dependent on this parameter (Souza, 2002; Stone & Moore, 1992). The advantage of choosing fast release times (e.g., <150 ms) is they adjust quickly to changes in input level, thus minimizing intensity differences between high- and low-level speech sounds and improving speech audibility (Hickson, 1994; Jenstad & Souza, 2005; Moore & Glasberg, 1988; Moore, Glasberg & Stone, 1991; Souza, 2002; Stone & Moore, 1992); however, the resultant smoothing of spectral peaks and valleys may remove important cues for place and manner of articulation, and voicing (Drullman, Festern & Plomp, 1994; Healy & Warren, 2003; Jenstad & Souza, 2005; Moore & Glasberg, 1988; Moore et al., 1991; Rosen, 1992; Shannon, Zeng, Kamath & Wygonski, 1995; Souza, 2002; Van der Horst, Leeuw & Dreschler, 1999). Also, as the hearing aid goes in and out of compression, the hearing aid wearer can experience a “pumping” sensation, which may be perceived as unpleasant (Moore, 2008; Moore & Glasberg, 1988; Moore et al., 1991). Furthermore, if the release time is shorter than 50 ms, gain may be increased during pauses between words, thus increasing the level of unwanted background noise (Moore, 2008; Stone & Moore, 2008). Alternatively, the goal of longer release times is to minimize manual volume adjustments on the part of the hearing aid wearer by maintaining a relatively constant output level (Moore, 2008; Moore & Glasberg, 1988; Moore et al., 1991; Souza, 2002); however, this processing may cause the wearer to miss out on auditory signals at low input levels during the period of time in which the hearing aid is releasing from compression (Moore, 2008).

For the purpose of estimating performance for speech, compression hearing aids should be measured using broadband test signals because they most closely approximate the spectral characteristics of speech (Souza, 2002), which will be explained in further detail below. Stelmachowicz, Kopun, Mace, & Lewis (1996) analyzed differences between three types of
hearing aid circuitry: (i) fast-acting compression, (ii) compression limiting (CL) and (iii) peak clipping (PC). In general, fast-acting compression is defined by fast attack and release times (Souza, 2002), peak-clipping refers to the type of processing whereby signals larger in voltage than some pre-set limit or the maximum electric output of a hearing aid cause the amplifier to clip (i.e., remove) the peaks of the signal (Agnew, 1998; Crain & van Tasell, 1994; Hawkins & Naidoo, 1993; Kates & Kozma-Spytek, 1994; Kuk, 1996; Larson, Williams, Henderson, Luethke, Beck, et al., 2000) and compression limiting employs a high compression ratio and high compression threshold to limit output without creating distortion (Souza, 2002).

Stelmachowicz et al. (1996) found that both PC and CL hearing aid circuits showed no significant gain differences between aided SPT (constant level) and speech stimuli at low input levels; however, at input levels greater than 70 dB SPL, gain was underestimated by the compared with the speech stimulus for both PC and CL circuits. The maximum difference between stimuli was 11 dB for the CL and 6 dB for the PC hearing aids at maximum input level. These findings suggest that both PC and CL hearing aids processed the SPT linearly for input levels up to 70 dB SPL, beyond which non-linear processing affected the tonal and speech stimuli differently. Furthermore, the SPT significantly underestimated gain for speech across all input levels for the fast-acting compression circuitry, which employs non-linear processing across all input levels. Smaller differences between aided speech-weighted composite noise (SWCN) and speech signals were reported for all three hearing aids compared with differences between aided SPT and speech stimuli. In combination, these findings suggest that complex stimuli may provide more valid measures of hearing aids with compression circuitry than tonal stimuli.
**Digital Signal Processing Strategies**

The main difference between the analog and digital hearing aids is the way in which they encode the incoming acoustic signal. In analog hearing aids, the incoming auditory signal is transformed into an analog electrical voltage by the microphone of the hearing aid and these voltages represent an analog of the frequencies and intensities present in the original stimulus (Levitt, 2007; Schweitzer, 1997). Digital hearing aids, on the other hand, transform analog electrical voltages into a very simple digital code (i.e., binary code) by an analog-to-digital (A/D) converter (Levitt, 2007; Schweitzer, 1997). This digital representation of auditory signals can be manipulated via algorithms, which allow digital hearing aids to use more advanced signal processing techniques than their analog counterparts (Levitt, 2007). Some examples include noise reduction algorithms and feedback management algorithms.

Many digital hearing aids employ noise reduction algorithms that make additional gain adjustments based on stimulus type in an attempt to reduce noise and improve listening comfort. Different manufacturers generally employ different algorithms; however, the ideal outcome is generally for signals with a relatively constant amplitude over time to be identified as noise because speech has more amplitude fluctuations than noise (including speech-shaped and speech in babble noise) (Bentler & Chiou, 2006). Theoretically, once some predetermined signal-to-noise ratio (SNR) had been exceeded, the noise reduction algorithm would reduce the gain for noise so that speech becomes maximally audible (Bentler & Chiou, 2006); however, in practice, when noise reduction algorithms detect noise, the gain for all signals entering the hearing aid is reduced (Bakke, Neuman & Toraskar, 1987; Hochberg, Boothroyd, Weiss, & Hellman, 1992; Levitt, 1991; Levitt, Bakke, Kates, Neuman, Schwander, & Weiss, 1993; Levitt, Neuman, Mills, & Schwander, 1986) and as a result, noise reduction algorithms serve to improve comfort in
noise compared with hearing aids that do not employ digital noise reduction (Boymans & Dreschler, 2000; Boymans, Dreschler & Shoneveld, 1999; Bray & Nilsson, 2001; Chung, 2004a; Levitt, 2001; Mueller, 2002; Mueller, Weber & Hornsby, 2006; Valente, Fabry, Potts, & Sandlin, 1998; Walden, Surr, Cord, et al., 2000). Stimulus choice is important when measuring noise reduction algorithms, as electroacoustic test stimuli with constant input levels over time (e.g., composite noise and pure tones) will be identified as noise, resulting in gain reduction (Groth, 2001; Kuk & Ludvigsen, 2003). As a result, real-ear measures of gain and hearing aid output using stimuli with constant input level would underestimate gain for speech (Kuk & Ludvigsen, 2003). The duration of the stimulus, however, may play a role in whether or not gain is reduced because noise reduction algorithms require analysis time (ranging from 1-10 s) to determine whether the temporal and spectral characteristics of the signal are noise-like (Bentler & Chiou, 2006; Chung, 2004a; Kuk & Ludvigsen, 2003). As a result, it is possible that stimuli shorter than one second may not be vulnerable to gain reduction by the noise reduction algorithm (Kuk & Ludvigsen, 2003); however, as a precaution, many manufacturers provide an option to disable digital noise reduction (Groth, 2001).

Another advanced feature available in digital hearing aids is feedback management. Once again, different manufacturers use different algorithms to manage feedback, including frequency shifting, adaptive notch filtering and adaptive gain reduction (Chung, 2004b). During real-ear verification measures, insertion of the probe tube is problematic for feedback because it changes the fit of the earmold or shell and consequently can create a feedback pathway (Groth, 2001); however, more important, is the effect of feedback management strategies on stimuli used for electroacoustic measures. Specifically, because the basic goal of feedback detection algorithms is to detect tonal stimuli and reduce or cancel their production in the hearing aid
output, the choice of tonal stimuli such as the SPT may not be appropriate for hearing aids with this feature (Chung, 2004b; Groth, 2001). Specifically, if the gain of a pure-tone stimulus is reduced because of a feedback management algorithm, hearing aid output and gain using this stimulus would not provide an accurate representation of gain for speech (Chung, 2004b; Groth, 2001). Some manufacturers, however, will provide an option to disable or not run the feedback management software so that feedback management will not occur during real-ear measures (Groth, 2001). Alternatively, a broadband stimulus can be used, which will not trigger the feedback reduction/cancellation algorithm (Groth, 2001).

The purpose of discussing these digital processing strategies was to outline the potential pitfalls of using certain stimuli to measure hearing aids with these features. In the current thesis, these features were disabled to ensure that they did not affect results. For instance, a result indicating that gain was lower in a particular condition could be attributed to feedback management if it were engaged during the test stimulus.

It is of additional importance that as a result of digital signal processing, digital hearing aids require additional processing time compared with analog hearing aids (Agnew & Thornton, 2000; Chung, 2004a; McGrath & Summerfield, 1985; Stone & Moore, 1999, 2002, 2003). Processing delay refers to the time it takes for a signal to pass from the microphone to the receiver of a hearing aid and depends on the amount of processing required by each algorithm. Processing delays can have many effects, including: (i) disturbing perception of the hearing aid wearers’ own voice (Agnew & Thornton, 2000; Stone & Moore, 1999, 2002) (ii) disturbing the perception of hearing aid processed speech (Chung, 2004a; Stone & Moore, 2002, 2003), and (iii) disrupting speech-reading cues due to a mismatch in timing between the auditory and visual...
cues (McGrath & Summerfield, 1985; Stone & Moore, 1999). Delays can be as short as one millisecond and as long as several hundred milliseconds (Chung, 2004a; Stone & Moore, 1999, 2002, 2003); however, even delays as small as 6 ms can cause perceptual disturbances (Agnew, 1997; Chung, 2004a; Stone & Moore, 2002, 2003) and 40-ms delays can negatively affect speech perception and production (McGrath & Summerfield, 1985; Stone & Moore, 1999, 2002, 2003). Additionally, across-frequency (“group”) delay refers to the relative delay across frequency channels of a hearing aid (Chung, 2004a; Stone & Moore, 1999) and it is believed that low-frequency channels have longer delays than high-frequency channels (Stone & Moore, 2002, 2003); therefore, disruptions caused by delays may be more objectionable for hearing aid wearers with good low-frequency hearing (Stone & Moore, 1999, 2002, 2003). Research has shown that larger overall processing delays are preferable to small across-frequency delays (Stone & Moore, 2002, 2003). Processing delays may lead to increased AEP waveform latencies in aided conditions relative to unaided, which will become important to consider when interpreting AEP waveforms for hearing aid validation purposes (Beauchaine, Gorga, Reiland, & Larson, 1986; Davidson, Wall & Goodman, 1990).

Electroacoustic Measures

The fitting and verification stages of hearing aid fitting involve setting the hearing aid to match prescriptive targets and verifying that targets have been met. Prescription procedures such as the NAL-NL1 (Byrne, Dillon, Ching, Katsch, & Keidser, 2001) or DSLm [i/o] (Scollie, Seewald, Cornelisse, Moodie, Bagatto, et al., 2005) specify the amplification targets for speech and maximum output necessary to provide loudness comfort, audibility of speech, and speech intelligibility (Alcantara, Moore & Marriage, 2004; Byrne & Dillon, 1986; Ching, Newall & Wigney, 1997; McCandless & Lyregaard, 1983; Snik & Stollman, 1995).
There are several options available for verifying fit to targets, including: (i) real-ear aided gain (REAG), (ii) real-ear insertion gain (REIG), (iii) real-ear aided response (REAR), and (iv) coupler measures. For all real-ear measures, a probe tube of 3-mm diameter is placed in the subject’s ear canal, close to the tympanic membrane. The probe tube insertion depth is a known source of variability (Burkhard & Sachs, 1977; Caldwell, Souza & Tremblay, 2006; Dirks & Kincaid, 1987; Gilman & Dirks, 1986; Hawkins & Mueller, 1986; Ringdahl & Leijon, 1984; Zemplenyi, Gilman & Dirks, 1985). Specifically, as the tip of the probe tube is moved further from the tympanic membrane, high-frequency measurements become less accurate due to standing waves (Caldwell et al., 2006; Dirks & Kincaid, 1987; Gilman & Dirks, 1986; Ringdhal & Leijon, 1984). In order to achieve measurement accuracy up to 6 kHz, the probe tube should be placed within 5 mm of the tympanic membrane (Caldwell et al., 2006; Hellstrom & Axelsson, 1993; Khanna & Stinson, 1985). This can be accomplished in several different ways: (i) the probe tube can be inserted 5 mm beyond the medial tip of the earmold (Burkhard & Sachs, 1977; Dirks & Kincaid, 1987; Scollie, Seewald, Cornelisse & Miller, 1998); (ii) the probe tube can be inserted until it touches the tympanic membrane and then withdrawn a few millimetres (Gerling & Engman, 1991; Storey & Dillon, 2001); or (iii) an average adult ear canal length can be assumed (Zemplenyi, Gilman & Dirks, 1985) and the probe tube can be inserted 30 mm beyond the tragal notch for males (Hawkins, Alvarez & Houlihan, 1991; Hawkins & Mueller, 1986; Zemplenyi et al., 1985), 28 mm for females (Zemplenyi et al., 1985), and 15-25 mm for children (Moodie, Seewald & Sinclair, 1994; Tharpe et al., 2001). In the current study, probe tubes were inserted to within 5 mm of the tympanic membrane using the third method listed above and subsequently verified by visual otoscopy.
The real-ear unaided response (REUR) is conducted with an unoccluded ear canal (Hawkins, 1987; Souza & Tremblay, 2006) and the real-ear aided response (REAR) is conducted with the hearing aid in place (Hawkins, 1987; Souza & Tremblay, 2006). The REAR is the measurement corresponding to hearing aid output, because it is the sum of input level and gain. Gain measures, such as REAG and REIG, are both derived from the REAR. Specifically, measurement of REAG involves subtracting the sound pressure level measured at the reference microphone from the REAR (Hawkins, 1987; Souza & Tremblay, 2006). The main difference between REAG and REIG measures is that REIG accounts for the gain provided by pinna and outer ear canal characteristics prior to insertion of the hearing aid (Hawkins, 1987; Souza & Tremblay, 2006). In other words, as opposed to subtracting the measurement value at the reference microphone, the REUR is subtracted from the REAR (Hawkins, 1987; Souza & Tremblay, 2006). As a result, the REAG is greater than the REIG by the amount of the individual's unaided response (Hawkins, 1987; Souza & Tremblay, 2006); therefore, if one is interested in comparing aided and unaided measures, REIG is a direct measure of the aided/unaided difference.

There is an important distinction between electroacoustic properties that are specified in terms of output (e.g., REAR) and those that are specified in terms of gain (e.g., REAG and REIG), as depicted in Figure 1.1. Hearing aid output is the sum of the input level and the gain provided by the hearing aid (i.e., output will increase with stimulus intensity), whereas hearing aid gain refers solely to the amount of gain provided at any given frequency. As an example, a linear hearing aid (which provides equivalent levels of gain across input levels) would show an increase in output as the stimulus level increased, regardless of the fact that gain setting was constant across input levels. It is important that these measures not be confused because
researchers may use either to report electroacoustic findings and may lead to different interpretations.

Stimuli Used to Measure Hearing Aid Characteristics

There is a significant amount of diversity among stimuli used to measure hearing aid characteristics, which can lead to different conclusions regarding the performance of hearing aids, particularly those with digital processing (Gorga, Beauchaine & Reiland, 1987; Preves, Beck, Burnett, & Teder, 1989; Scollie & Seewald, 2002; Stelmachowicz et al., 1990, 1996). In particular, stimuli can be distinguished according to their spectral characteristics, modulation, and crest factor; this information is provided in Table 1.1. Crest factor refers to the difference between peak amplitude and RMS level over a given integration time (e.g., 125 ms is a standard integration time for hearing aid fitting) (Amlani, Punch & Ching, 2002; Frye, 1987, Valente, 2002). The crest factor of a stimulus is important because it will determine the measured gain. For instance, even at very low stimulus intensities, hearing aids can be forced into saturation (i.e., limit in some way) if the crest factor is large (Frye, 1987; Stelmachowicz et al., 1990). This

Figure 1.1. Output (a) and gain (b) as a function of input.
occurs because even at low RMS levels there may be instantaneous peaks that exceed the compression limiting (CL) or peak clipping (PC) processing of the hearing aid when the crest factor of a stimulus is high. This would not predict the gain provided by that hearing aid for another signal with a low crest factor.

In addition to the spectrum, bandwidth, modulation, and crest factor, one may also manipulate stimulus duration, which may actually be a crucial characteristic when determining the appropriateness of stimuli for hearing aid evaluation measures as longer duration stimuli may result in more accurate hearing aid measurements. This will be covered in further detail when auditory evoked potential measures of hearing aids are discussed below.

Importantly, depending on their characteristics, not all stimuli are optimal for measuring each hearing aid parameter (Henning & Bentler, 2005; Scollie & Seewald, 2002; Stelmachowicz et al., 1990, 1996). For the purpose of hearing aid fitting and verification, it is important to consider: (i) which signal will provide a valid measure of the mechanism or algorithm to be evaluated, and (ii) which signal will provide a valid estimate of speech.

Tonal stimuli

There are two commonly used types of SPT stimuli, one in which the series of tonal stimuli are presented at a constant input level and another in which the series of tonal stimuli are speech-weighted. Because tonal stimuli (e.g., pure tones and pure tone sweeps) do not have high crest factors, they can be presented at higher RMS levels compared with complex stimuli before causing the hearing aid to saturate (Stelmachowicz et al., 1990). As a result, tonal stimuli provide a better measure of MPO than complex stimuli (Stelmachowicz et al., 1990).
Table 1.1. The stimulus characteristics of various commercially available test stimuli and their common usage in the hearing aid fitting process.

<table>
<thead>
<tr>
<th>Stimulus Characteristics</th>
<th>Spectrum</th>
<th>Modulation</th>
<th>Crest Factor</th>
<th>Common Usage</th>
</tr>
</thead>
<tbody>
<tr>
<td>Continuous Tone</td>
<td>Narrow</td>
<td>No</td>
<td>Low</td>
<td>Threshold Measures</td>
</tr>
<tr>
<td>Swept Pure Tone</td>
<td>Narrow</td>
<td>No</td>
<td>Low</td>
<td>Electroacoustic Measures</td>
</tr>
<tr>
<td></td>
<td>Continuous</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>level or</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Speech-weighted</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Warble Tone</td>
<td>Narrow</td>
<td>Yes</td>
<td>6 dB (low &gt;SPT)</td>
<td>Threshold and Electroacoustic Measures</td>
</tr>
<tr>
<td></td>
<td>Continuous</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>level or</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Speech-weighted</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Composite Noise</td>
<td>Broad</td>
<td>Yes</td>
<td>12 dB (moderate)</td>
<td>Threshold and Electroacoustic Measures</td>
</tr>
<tr>
<td></td>
<td>Continuous</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>level or</td>
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</tr>
<tr>
<td></td>
<td>Speech-weighted</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ICRA Noise</td>
<td>Broad</td>
<td>Yes</td>
<td>Low</td>
<td>Electroacoustic Measures</td>
</tr>
<tr>
<td></td>
<td>Speech-weighted</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Speech</td>
<td>Broad</td>
<td>Yes</td>
<td>12 -15 dB (moderate)</td>
<td>Threshold and Electroacoustic Measures</td>
</tr>
</tbody>
</table>

On the other hand, even when speech-weighted, swept pure tone signals are unlike speech in several ways: (i) each tone is presented in sequence which is unlike continuous discourse, (ii) the duration of each tone may be longer than individual components of speech, and (iii) the crest factor is 3 dB, which is much lower than that of speech and speech-weighted complex stimuli (e.g., 15 dB). As a result, hearing aid gain measured using a SPT signal does not provide a valid estimate of hearing aid gain for speech in hearing aids with non-linear amplification (Henning & Bentler, 2005; Scollie & Seewald, 2002; Stelmachowicz et al., 1990, 1996). Stelmachowicz et al. (1996) compared hearing aid gain for speech and constant level SPT stimuli. Stimuli were presented at RMS levels of 50, 60, 70, and 80 dB SPL. They tested 20 different hearing aids (8 analog) with a variety of processing capabilities, including: (i) linear,
(ii) fast-acting compression, (iii) PC, (iv) CL, and several other compression circuits that varied according to threshold, ratio and attack and release time. Comparisons between aided speech and aided SPT (constant level) pooled across hearing aid types revealed that the SPT significantly underestimated gain for speech in the high frequencies and had a tendency to overestimate gain in the low frequencies, similar to the “blooming effect” described by Preves, Beck, Burnett, and Teder (1989).

Scollie and Seewald (2002) also conducted a study in which they compared speech and SPT stimuli; however, they described their results in terms of hearing aid output as opposed to gain. They found that when pooled across hearing aid types, aided SPT (constant level) tended to overestimate output for aided speech at all frequencies, particularly above 3000 Hz. These findings are consistent with previous research by Preves et al. (1989) and Stelmachovicz et al. (1996).

*Composite noise*

Composite noise is a broadband signal that contains energy across the entire frequency spectrum. As with the SPT stimulus, composite noise can be constant in level across the frequency range or it can be speech-weighted. Speech-weighted composite noise (SWCN) has been designed to imitate the properties of speech in three ways: (i) it approximates the spectral characteristics of the long-term average speech spectrum (LTASS) (Frye, 1986; Scollie & Seewald, 2002), (ii) the bandwidth is larger than that of a pure-tone stimulus (Frye, 1986; Scollie & Seewald, 2002), and (iii) it is composed of many different frequencies that are presented simultaneously with a randomized phase pattern and because signal components are not in phase, the result is a crest factor of 12 dB; whereas, if the signal components were not randomized, the
crest factor would be much higher (e.g., ≥19 dB) (Frye, 1986; Scollie & Seewald, 2002). As a result, the SWCN has potential to more closely approximate gain for speech compared with tonal stimuli.

In the same study described above, Stelmachowicz et al. (1996) compared gain for SWCN and speech. Their results showed that when pooled across hearing aids, the SWCN stimulus more closely approximated gain for speech in the low frequencies than the aided SPT signal; however, both the SWCN and SPT tended to underestimate gain for real speech in the high frequencies. Additionally, variability among hearing aids for the SWCN signal was quite high. Specifically, the SWCN showed significantly higher variability at high input levels than at low input levels, as well as showing significantly higher variability than the SPT stimulus at high input levels. Stelmachowicz and colleagues proposed that this may be due to intermodulation distortion, which is the result of two or more signals mixing together. They concluded that SWCN performed only slightly better at predicting gain for real speech compared with the SPT stimulus.

Scollie and Seewald (2002) also examined the difference in hearing aid output measured using speech and composite noise. They found that aided output for composite noise was a better match with aided speech output than was aided SPT output, which tended to overestimate output for speech. In cases of high input levels or high-power instruments, however, the composite noise signal tended to underestimate gain for speech. This imperfect match at high input levels between composite noise and speech was similarly discussed by other researchers (e.g., Cox & Flamme, 1998; Henning & Bentler, 2005; Stelmachowicz et al., 1996), although
only Scollie and Seewald (2002) determined that this result was due to noise reduction
algorithms that could not be deactivated in some hearing aids.

*Warble tone*

More correctly referred to as a frequency modulated tone, the warble tone is a
narrowband stimulus that modulates slightly in frequency at a very fast rate (Erulkar, Butler &
Gerstein, 1967). This tone is used behaviourally as an adjunct to pure tone stimuli, particularly
for threshold testing in the sound field (Dockum & Robinson, 1975; Morgan, Dirks & Bower,
1979). The swept warble tone, on the other hand, is used for electroacoustic testing and as with
the SPT and composite noise signals, the swept-warble tone can have a constant input level
across frequencies or be speech-weighted. The crest factor for this stimulus is lower than for
speech but higher than for SPT at approximately 6 dB (Seewald & Scollie, 2002). As a result,
aided thresholds for validation purposes using the speech-weighted warble-tone could potentially
provide a better estimation of speech than pure-tone stimuli.

Stelmachowicz et al. (1996) reported that the remaining three stimuli in their study: (i)
the speech-weighted swept warble, (ii) speech-modulated noise (similar to ICRA noise described
in Table 1.1), and (iii) simulated speech, all closely approximated the gain measured for real
speech. Similarly, Scollie and Seewald (2002) found that speech-weighted swept warble tones
were a much better match for speech output and showed significantly smaller levels of error
compared with the SPT stimulus. As with the composite noise signal, differences increased
between the swept warble tone and speech stimulus at high input levels. As previously
mentioned, these differences can be attributed to noise reduction algorithms that could not be
disabled in several hearing aids used in this study. It is interesting that although the SWCN
stimulus tended to underestimate speech (as noted above), the swept warble tone tended to overestimate speech. These findings suggest that the swept warble tone did not activate the noise reduction algorithm because it is not a continuous signal.

All of the findings described above help to illustrate the importance of choosing stimuli based on the purpose of conducting hearing aid measures. Specifically, speech and non-speech signals will result in different levels of gain or output, leading to very different conclusions regarding hearing aid performance (Henning & Bentler, 2005). If the difference in gain measured using speech and non-speech signals is sufficiently large, gain for speech may be under or overestimated (Henning & Bentler, 2005; Scollie & Seewald, 2002; Stelmachowicz et al., 1996). For this reason, complex stimuli provide a better estimate of gain for speech compared with tonal stimuli. On the other hand, complex stimuli are not necessarily optimal for measuring other hearing aid parameters, such as MPO, which is measured more accurately using tonal stimuli. Consequently, it is possible that the stimuli used to elicit AEPs may not be best for measuring certain hearing aid parameters.

Hearing Aid Validation

The final stage in hearing aid fitting is the validation stage, which begins immediately after verification and continues for many years afterwards (Valente et al., 2006). During this stage of the fitting process it is often necessary to adjust hearing aid settings in accordance with individual preferences for sound quality and comfort (Byrne, 1986; Leijon, Eriksson-Mangold, & Bech-Karlson, 1984; Leijon, Lindkvist, Ringdahl, & Israelsson, 1990). In adults, this is accomplished using a variety of functional measures, including: aided thresholds, communication with the patient, verbal feedback, and written feedback through questionnaires.
such as the Hearing Handicap Inventory for the Elderly (Ventry & Weinstein, 1982) and the Hearing Aid Performance Inventory (Walden, Demorest & Hepler, 1984). Unfortunately, there is a limited array of validation measures for infants, who cannot participate in conventional audiometric and/or speech recognition testing and, as a result, it is difficult to ascertain that hearing aids are providing benefit for speech recognition, while maintaining a level below loudness discomfort (Damarla & Manjula, 2007; Golding, Pearce, Seymour, Cooper, Ching, & Dillon, 2007). Questionnaires such as the Parent’s Evaluation of Aural/Oral Performance in Children (PEACH) are often relied upon in these instances to provide information regarding sound audibility and comfort levels (Ching & Hill, 2007; Harrison, 2000). Unfortunately, caregiver questionnaires are dependent on caregiver presence and recollection of specific behaviours, and provide little information regarding the perception and/or discrimination of individual speech sounds (Golding et al., 2007; Purdy, Katsch, Dillon, Storey, Sharma, & Agung, 2005). As a result, there is a strong need for reliable objective measures of hearing aid validation in infants that can provide information regarding comfort and speech perception. Several auditory evoked potentials have been considered as potential objective validation measures, including the auditory brainstem response (ABR), the auditory steady-state response (ASSR) and the slow cortical potential (SCP).
ELECTROPHYSIOLOGICAL METHODS OF HEARING AID EVALUATION

For the purpose of hearing aid evaluation, AEPs have been considered in two primary ways: (i) to determine whether aided AEPs can be measured at input levels where unaided AEPs could not, and (ii) to determine whether changes in hearing aid settings and/or stimuli can be measured using aided AEPs and if so, whether these measures have behavioural correlates such as loudness discomfort.

Using the Auditory Brainstem Response (ABR) for Hearing Aid Evaluation

The ABR is an evoked potential consisting of a series of vertex-positive peaks which occur within 15-20 ms of stimulus onset, numbered according to their temporal position in the waveform (Jewett, Romano & Williston, 1970; Jewett & Williston, 1971). Waves I and V are the most commonly evaluated ABR peaks as they are the least variable in amplitude and morphology (Amadeo & Shagass, 1973; Burkard & Don, 2007; Jewett & Williston, 1971). The proposed generators of the ABR waveform are the spiral ganglion (wave I), the auditory nerve at the exit of the internal auditory meatus (wave II), the ventral cochlear nucleus (wave III), the superior olivary complex and initial portion of the lateral lemniscus (wave IV), and the ascending portion of the lateral lemniscus and inferior colliculus (wave V) (Burkard & Don, 2007; Jewett & Williston, 1971; Lev & Sohmer, 1972; Maurizi, Altissimi, Ottaviani, & Paludetti, 1982). The tone-evoked ABR has proven successful in hearing diagnostics as it is a response that is not affected by sleep or sedation and therefore, can be used for hearing assessment in even very small infants (Burkard & Don, 2007; Hyde, 1985; Stapells, 1989; Stapells, Gravel & Martin, 1995; Stapells, Picton & Durieux-Smith, 1994). Routine use of the click as a stimulus for measuring auditory thresholds is strongly discouraged (Stapells & Oates, 1997; Stapells et al., 1994, 1995); however, it is still used clinically for ABR neurological (i.e.,
suprathreshold) testing. This stimulus is a very short duration or transient signal, 0.1-ms square wave that when transduced by earphone results in an asymmetric stimulus typically 1-2 ms in duration and contains components across the entire frequency range (Burkard & Secor, 2002; Durrant & Boston, 2007). As previously mentioned, the crest factor for click stimuli is very high and for this reason it may not be an appropriate stimulus for electroacoustic measures of hearing aid processing. Regardless, several click-evoked ABR measures have been assessed for their application in the hearing aid validation process; these include wave V thresholds (Kileny, 1982; Mahoney, Condie & Snyder, 1980; Mokotoff & Krebs, 1976) and wave V latencies (Beauchaine et al., 1986; Hecox, 1983).

Aided wave V thresholds

Several early studies reported ABR wave V thresholds for subjects while wearing their personal hearing aids at the same input level where no unaided wave V thresholds existed (Mahoney et al., 1980; Mokotoff & Krebs, 1976). These studies generated a lot of interest regarding the feasibility of using this measure for hearing aid validation. For instance, Kileny (1982) reported a series of case studies using analog hearing aids in which wave V thresholds were used successfully to measure hearing aid benefit in a group of children. He found lower (better) aided wave V thresholds compared with unaided wave V thresholds when the subject was fitted with a hearing aid appropriate for their level of hearing loss. These results implied that wave V thresholds may be an appropriate measure of hearing aid benefit; however, there were several problems associated with the procedure used by Kileny (1982). For instance, only four cases were discussed in which aided thresholds were measured; therefore, these findings are difficult to generalize to a larger population. Additionally, threshold measures only provide information about how the hearing aid functions at threshold levels. Functional gain measures
have several limitations, including failure to assess non-linear hearing aid performance at conversational and loudness discomfort levels, and an inability to assess the dynamic features of the hearing aid, such as compression (which requires a dynamic input level) (Seewald, Hudson, Gagne, & Zelisko, 1992; Stelmachowicz & Lewis, 1988). More information would be required before successful validation of the hearing aid could be achieved; including gain provided at conversational levels and loudness discomfort levels.

Aided wave V latency

Hecox (1983) proposed using ABR wave V latency as a correlate of loudness perception. He hypothesized that the slope of the latency-intensity (L-I) function would improve the success of hearing aid fitting by predicting a subjects’ ability to tolerate loudness. This procedure was suggested as an adjunct to more traditional measures of hearing aid fitting in cases where those measures failed to provide the subject with a comfortable fitting. Hecox tested his theory using the slope of the L-I function and absolute wave V latencies for subjects with sensorineural hearing loss in aided and unaided conditions. In a normal hearing individual, wave V latencies become shorter as intensity is increased; however, based on aided and unaided ABR findings, this pattern was different in individuals with hearing loss. Hecox believed that the observed patterns could be used to determine what type of hearing aid would best suit a hearing loss. For instance, (i) if the slope was normal but the whole function was shifted up on the intensity scale, then the subject would do well with a linear fitting, (ii) if there were no aided responses or if the subject had a central dysfunction, then hearing aids would not help, and (iii) a steeper than normal L-I function was a sign of recruitment (i.e., reduced dynamic range) and that subject would benefit from compression circuitry. Benefits from compression circuitry reported by Hecox (1983) were based on improved acceptance of hearing aids by the hearing aid wearer as
well as improved dynamic range, evidenced by lower wave V threshold and larger wave V amplitude at 60 dB HL while wearing compression hearing aids compared with responses while wearing linear hearing aids.

Results of more recent studies have questioned the validity of Hecox’s hypothesis that the L-I function is related to recruitment. In particular, studies have suggested that the slope of the L-I function may at least be partially related to the configuration of the hearing loss and not to the perception of loudness (Eggermont, 1982; Gorga, Reiland & Beauchaine, 1985; Gorga, Worthington, Reiland, Beauchaine, & Goldgar, 1985). In partial agreement, Serpanos, O’Malley and Gravel (1997) reported that the wave V L-I function (in response to click stimuli) can be a useful index of loudness perception and recruitment, but only in groups of individuals with normal hearing or flat sensorineural hearing loss. Slope of the L-I function was not a useful measure of loudness perception in groups of individuals with sloping sensorineural hearing loss.

Additionally, several studies have reported variable temporal delays imposed by different hearing aids (Beauchaine et al., 1986; Davidson et al., 1990) which adds an additional confound to any aided measure of latency. Beauchaine et al. (1986) reported that ABR waveforms were delayed while subjects were wearing hearing aids. This indicates that the previously discussed temporal delays imposed by digital hearing aids may be measurable in evoked potential waveforms. Beauchaine et al. (1986) concluded that a correction factor must be applied to aided ABR wave V latencies before they could be compared with unaided latencies. This was not addressed by Hecox (1983); therefore, Beauchaine and colleagues advised that the results of his study must be interpreted with caution as the L-I function may have underestimated high-frequency gain predictions by as much as 15 dB without the correction factor.
Limitations of the click-ABR for hearing aid evaluation

The most important limitation of click-evoked ABRs is that click stimuli are broadband and thus do not represent accurate measures of hearing thresholds for any specific frequency, and may completely miss or underestimate hearing loss in particular frequency regions (Eggermont & Don, 1980; Picton, 1978; Picton & Durieux-Smith, 1988; Picton, Ouellette, Hamel, & Durieux-Smith, 1979; Stapells, 1989; Stapells et al., 1994). This makes the click-evoked ABR an inappropriate stimulus for distinguishing between hearing aids with different low-frequency responses and for validating hearing aid settings for subjects with good low-frequency hearing thresholds.

The tone-ABR uses a more frequency-specific stimulus (Oates & Stapells, 1997; Stapells et al., 1994) with potential to overcome this stimulus problem; however, there is research to suggest that all brief stimuli may be problematic for measuring hearing aids. In particular, studies have suggested that transient stimuli used to evoke ABRs may be too brief to effectively measure hearing aids (Brown, Klein & Snydee, 1999; Gorga et al., 1987; Kiessling, 1982; Levillain, Garcon & Le Her, 1979). Gorga et al. (1987) used tone-evoked ABR stimuli (2000-Hz; 0.5-ms rise time; 10-ms duration) to examine hearing aid output in non-linear hearing aids. They were particularly interested in whether the compression circuitry would be activated by a tone of such short rise-time (0.5-ms chosen to approximate the short rise-time of a click stimulus) and whether there would be an interaction between stimulus rate and hearing aid attack/release time. Interestingly, they found that there were significant differences between the hearing aid responses to onset and steady-state portions of the waveform and this effect was dependent on the stimulus rate and intensity. Specifically, differences between hearing aid responses to onset and steady state portions of the brief tone became smaller as the intensity and
stimulus rate increased. Additionally, there was considerable variability across hearing aids, such that some hearing aids showed larger onset outputs, while others showed larger steady-state outputs. In particular, one hearing aid showed an onset output that was 20-dB less intense than its steady-state output. Although Gorga et al. (1987) found that there was a tendency for some hearing aids to disagree less between onset and steady-state responses at high stimulus rates (e.g., 50/s), this trend was very small and overall, Gorga et al. (1987) concluded that the observed processing discrepancies could not be explained by the attack/release time, compression type, or compression ratio of the hearing aid. Brown et al. (1999) similarly reported that brief-tone stimuli did not activate any of the hearing aid compression systems used in their study (i.e., WDRC, super-compression, super-compression with adaptive release time, soft peak-clipping, and peak clipping). As a result, the hearing aid output for brief tones always exceeded the output for continuous tones. Brown et al. (1999) also proposed that differences in performance between brief and continuous stimuli cannot be attributed to attack time alone. They found that brief tones resulted in larger output levels than did continuous stimuli, even when fast-acting, peak-clipping circuits with less than 1-ms attack time were tested.

Finally, some researchers have reported difficulties with measuring aided ABRs for normal hearing subjects. Beauchaine et al. (1986) found that they could not use wave V threshold shifts in normal hearing subjects to estimate hearing aid gain because wave V thresholds were not significantly different in aided and unaided conditions. Conversely, hearing-impaired subjects showed significant differences between aided and unaided wave V thresholds. These findings could be due to several factors: (i) the hearing aid circuit noise may have been audible to normal-hearing subjects (Agnew, 1997), (ii) alterations to the stimulus resulting from hearing aid processing (e.g., compression, noise reduction, or feedback reduction) may have
been audible to listeners with an normal peripheral auditory system, or (iii) noise levels in the sound booth may have been audible to subjects with normal hearing (Pittman, Stelmachowicz, Lewis, & Hoover, 2003).

Current status of ABR use for hearing aid evaluation

There has not been much success in discovering valid ABR measures to assess hearing aids using the click-evoked ABR, and although little research has been carried out measuring hearing aids using the tone-evoked ABR, the research by Gorga et al. (1987) and Brown et al. (1999) suggest that all the transient stimuli for the ABR may be too brief to accurately measure hearing aids. This ABR stimulus problem led researchers to examine other auditory evoked potentials to use for hearing aid measures. In particular, because the SCP can employ somewhat longer stimuli and the 80-Hz auditory steady-state response uses continuous stimuli, both AEP measures may show promise for hearing aid evaluations, and may solve some of the problems associated with brief transient stimuli.

Using the Auditory Steady-State Response (ASSR) for Hearing Aid Evaluation

The ASSR is a brainstem-evoked potential that occurs in response to a steady-state, amplitude and/or frequency modulated continuous tone (Campbell, Atkinson, Francis, & Green, 1977; Galambos, Makeig & Talmachoff, 1981; Lins & Picton, 1995; Lins, Picton, Boucher, Durieux-Smith, Champagne, & Moran, 1996; Picton, Durieux-Smith, Champagne, Wittingham, Moran, Giguère, et al. 1998). The 80-Hz ASSR is quickly becoming a commonly used threshold test for children (Picton, Dimitrijevic, van Roon, John, Reed, & Finkelstein, 2002; Stapells, Herdman, Small, Dimitrijevic, & Hatton, 2005). There are several reasons for the

Studies have shown that ambient noise levels in a typical sound-attenuated booth range from 17-45 dB SPL in the low frequencies and 10-20 dB SPL in the high frequencies (Frank & Williams, 1993).
increasing popularity of this measure: (i) it can be recorded in infants, whether sleeping or
awake (Aoyagi, Kiren, Kim, Suzuki, Fuse, & Koike, 1993; Cohen, Rickards & Clark, 1991;
Levi, Folsom & Dobie, 1993; Lins & Picton, 1995; Lins, Picton, Picton, Champagne, &
Durieux-Smith, 1995; Rance, Dowell, Rickards, Beer, & Clark, 1998; Rance, Rickards, Cohen,
De Vidi, & Clark, 1995; Rickards, Tan, Cohen, Wilson, Drew, & Clark, 1994; Small, Hatton &
Stapells, 2007), (iii) it can be measured down to near-threshold in normal-hearing as well as
hearing-impaired individuals (Aoyagi, Suzuki, Yokota, Furuse, Watanabe, & Ito, 1999;
Dimitrijevic, John, van Roon, Purcell, Adamonis, & Ostroff, 2002; Herdman & Stapells, 2001,
2003; Lins & Picton, 1995; Lins et al., 1995, 1996; Picton, Dimitrijevic, Perez-Abalo, & van
Roon, 2005; Picton et al., 1998; Rance et al., 1995, 1998; Rickards et al., 1994; Small et al.,
2007; Swanepoel, Hugo & Roode, 2004; Van Maanen & Stapells, 2005), (iv) at least four
simultaneous stimuli can be presented monaurally with little loss of response amplitudes
compared with single stimulus presentations (Dimitrijevic et al., 2002; Herdman & Stapells,
2001, 2003; John, Lins, Boucher, & Picton, 1998; Lins & Picton, 1995) and (v) most importantly
for the purpose of the current study, ASSR measures typically use continuous, steady-state
stimuli with low crest factors which would allow a hearing aid to settle into its steady-state and
are less likely to result in distortion from the sound source (e.g., sound field speaker) or in a
hearing aid (Picton et al., 1998). On the other hand, the longer duration stimulus used for ASSR
measures may be susceptible to reduction by digital noise reduction and feedback reduction
algorithms available in digital hearing aids.

Aided 80-Hz ASSR

Picton et al. (1998) provided the first examination of aided 80-Hz single and multiple-
stimulus ASSR thresholds. They compared aided ASSR thresholds and behavioural thresholds
obtained via conventional audiometry in a group of normal hearing and hearing impaired subjects. For subjects with normal hearing, average differences between behavioural and ASSR thresholds were within 18-26 dB, which is a larger difference than reported in previous studies (Tlumak, Rubinstein & Durrant, 2007). Also, differences across individual subjects revealed that aided ASSR thresholds were 5-25 dB higher than behavioural thresholds. Although not ideal, according to Picton et al. (1998), this 20-dB range is acceptable when no other information exists regarding aided thresholds.

The introduction of multiple-stimulus ASSRs to the hearing aid evaluation process was a natural progression as it has the potential to significantly reduce evaluation time. As mentioned previously, Picton et al. (1998) also examined multiple-stimulus ASSRs in individuals with normal hearing and in individuals with hearing loss while wearing their personal hearing aids. They found that presenting four simultaneous stimuli in normal-hearing subjects did not significantly reduce ASSR amplitude compared with single-stimulus presentations. These findings parallel those described in other studies (Dimitrijevic et al., 2002; Herdman & Stapells, 2001, 2003; John et al., 1998; Lins & Picton, 1995). Interestingly, however, when stimuli were processed by a hearing aid, ASSR thresholds were significantly better for the single-stimulus presentation compared with multi-stimulus presentation in some subjects with hearing loss (although information regarding the number of subjects was not given). This finding suggests that although the multiple-stimulus ASSR is more efficient and equally accurate as the single-stimulus ASSR for determining unaided thresholds, it may not be more efficient when measuring ASSRs with hearing aids in place; however, other researchers (e.g., Herdman & Stapells, 2003) have suggested that Picton et al. (1998) overstated this potential problem. Further research on aided multiple-stimulus ASSRs has not yet been conducted.
A more recent study by Stoebel, Swanepoel and Groenewald (2007) sought to determine whether the aided single-stimulus ASSR findings of Picton et al. (1998) could be applied to infants. They tested six infants between the ages of three and six months using unaided and aided ASSRs and compared these findings with aided behavioural thresholds obtained via visual reinforcement audiometry at a later date. They found that across all measured frequencies, aided single-stimulus ASSR thresholds were higher than aided behavioural thresholds with average differences of 13 dB. For individual subjects, they found that in 63% of cases, aided ASSR thresholds were within 15 dB of aided behavioural thresholds, and in 79% of cases they were within 20 dB of aided behavioural thresholds. Average differences in this study are comparable to those of Picton et al. (1998); however, the standard deviation was much larger (13 dB) compared with those observed in the Picton et al. (1998) study (7-9 dB). Similar to Picton and colleagues, they concluded that aided ASSR thresholds provide useful information when no other aided threshold information is available. There were several problems associated with this study, however, which should be noted. First of all, the sample size was very small (n = 6); therefore, it is difficult to generalize these results to a larger population. Secondly, because the subjects were infants, the only available method of behavioural testing was visual reinforcement audiometry; therefore, the reported thresholds may have been minimum response levels rather than true thresholds. This means that actual differences between ASSR and behavioural thresholds may have been larger than reported.

Instead of comparing aided ASSR and behavioural thresholds, a recent study by Damarla and Manjula (2007) reported greater success by comparing ASSR functional gain (i.e., aided minus unaided ASSR thresholds) with REIG in a group of 37 adult subjects with mild to moderately-severe sensorineural hearing loss of undisclosed configuration. Subjects were fitted
with one of two digital hearing aids based on best match to prescription targets. ASSR gain was calculated at 500, 1000, 2000, and 4000 Hz. They reported a positive correlation between ASSR gain and REIG (e.g., 0.63, 0.67, 0.63, and 0.62 at 500, 1000, 2000, and 4000 Hz, respectively), such that there was no significant difference between the two estimates at any of the four frequencies tested, although they did not report the processing types used across hearing aids (i.e., linear vs. non-linear). At first glance, these findings seem to indicate that aided ASSR thresholds may be a useful measure for hearing aid evaluation; however, as with all functional gain measures, aided ASSR thresholds are susceptible to masking by circuit and ambient noise, and do not provide information regarding gain at supra-threshold levels.

**Limitations of the ASSR for hearing aid evaluation**

Upon reviewing the literature, it is immediately obvious that there is a lack of research associated with the ASSR and hearing aid evaluation. In order to determine whether the ASSR is feasible for selecting, setting and verifying hearing aids, further studies are required. Also, all of the studies described above used ASSR threshold measures and thus are susceptible to problems associated with functional gain. Additionally, studies have suggested that the 80-Hz ASSR and the ABR may be different measures of the same response, because ASSR frequency specificity is very similar to that of the ABR, despite using more frequency-specific acoustic stimuli (Herdman, Picton & Stapells, 2002; Herdman & Stapells, 2003). As a result, the ASSR may not reflect the entire stimulus, but rather only a portion of each modulation cycle (Herdman et al., 2002). In any case, the 80-Hz ASSR is still a brainstem-generated response and regardless of the fact that it may have addressed the transient stimulus problem associated with ABRs, it has limited predictive ability for higher-level processing and perception.
Using the Slow Cortical Potential (SCP) for Hearing Aid Evaluation

There are many characteristics of the P1-N1-P2 complex that make it potentially desirable for use as a hearing aid evaluation tool. For instance, (i) there is a strong relationship between SCP and behavioural thresholds (e.g., Beagley & Kellogg, 1969; Kaf, Durrant, Sabo, Boston, Taubman, & Kovacyk, 2006; Lightfoot & Kennedy, 2006; Tomlin, Rance, Graydon, & Tsialios, 2006; Van Maanen & Stapells, 2005), (ii) cortical potentials can be recorded in infants (Golding et al., 2007; Gravel, Kurtzberg, Stapells, Vaughan and Wallace, 1989; Pearce, Golding & Dillon, 2007; Purdy et al., 2005; Rapin & Graziani, 1967; Wunderlich, Cone-Wesson & Shepherd, 2006), (iii) cortical potentials can be readily evoked by a variety of stimuli, including longer stimuli such as speech (Billings, Tremblay, Souza, & Binns, 2007; Golding et al., 2007; Korczak, Kurtzberg & Stapells, 2005; Oates, Kurtzberg & Stapells, 2002; Pearce et al., 2007; Purdy et al., 2005; Tremblay, Billings, Friesen, & Souza, 2006), and (iv) differences among speech stimuli are evident by changes in the response waveform (Agung, Purdy & Kitamura, 2005; Aiken & Picton, 2008; Ostroff, Martin & Boothroyd, 1998; Sharma & Dorman, 1999; Tremblay, Friesen, Martin, & Wright, 2003; Tremblay et al., 2006), which some researchers have suggested may indicate a link between cortical responses and behavioural measures of auditory discrimination (Purdy et al., 2005). At the very least, it might be possible improve perception by making adjustments to hearing aid settings so that neural response patterns are optimized (Tremblay et al., 2006). The SCP thus could address limitations of the ABR and ASSR methods, discussed above, in the assessment of dynamic hearing aid characteristics.

The P1-N1-P2 complex was one of the very first auditory evoked potentials used in the hearing aid evaluation process. In fact, the first experiment examining aided cortical potentials was reported in 1967 by Rapin and Graziani. In this study, clicks and brief tones were presented
in the sound field to evoke aided SCPs. They tested eight infants (with various types and
degrees of developmental disability) using their personal hearing aids (all analog) and hearing
aid effectiveness was determined by shifts in N1 threshold (i.e., aided N1 presence at an input
intensity which did not result in an unaided N1 response). They found that most infants
demonstrated improved SCP thresholds while wearing their hearing aids. In particular, five of
the eight subjects showed larger and more defined aided waveforms than their unaided
waveforms. One subject who had only recently been fitted with a hearing aid had insufficient
data to make a valid threshold comparison; however, aided cortical thresholds did seem to be
improved and neural responses were better defined compared with unaided responses. The final
two subjects did not show any detectable change in SCP thresholds while wearing their hearing
aids. Of these two, one subject had only recently been fitted with a hearing aid and the other
would not accept the hearing aid and was deriving no observable benefit from the aid according
to his mother. Rapin and Graziani (1967) concluded that these preliminary aided SCP threshold
measures showed promise as a clinical tool for predicting hearing aid benefit.

A series of case studies reported by Gravel et al. (1989) examined aided SCPs in children
with various cognitive and developmental disabilities. They used speech-evoked (/ta/ and /da/)
SCP measures as part of an audiologic test battery to determine whether hearing aids were
providing any measurable benefit in these children. Overall, they determined that aided SCPs
were able to measure benefit, as determined by threshold shifts with hearing aids in place. In
one case of a child with profound sensorineural hearing loss, they found that although no
identifiable SCP waveforms were visible for either speech stimulus in the unaided condition,
measurable responses were obtained in the aided condition for both stimuli. According to these
researchers, these aided SCP responses were lower in voltage than normal SCP responses;
however, they suggested that some degree of residual hearing was present in this individual. Another subject with bilateral aural atresia and normal cochlear hearing (as determined by bone-conduction ABR testing), showed good aided SCP responses to the /da/ stimulus and borderline responses for the /ta/ stimulus measured through a bone-conduction hearing aid. These findings suggest that aided SCPs can be useful in determining whether hearing aids are providing usable gain for subjects with sensorineural and/or conductive hearing loss.

More recently, a study by Korczak et al. (2005) compared the speech-evoked cortical responses of subjects with sensorineural hearing loss with those of normal-hearing subjects by examining SCPs and behavioural measures of sensitivity and reaction time. The speech stimuli for this study were /ba/ and /da/, chosen because they differ in place of articulation, which is a difficult articulatory feature to discriminate for individuals with hearing loss (Boothroyd, 1984; Dubno, Dirks & Ellison, 1989; Dubno, Dirks & Langhofer, 1982; Gordon, 1987; Walden, 1984). Although Korczak and colleagues examined several different cortical potentials (N1, MMN, N2b, P3b), the following discussion will focus on their SCP N1 results as it shows the most potential for measuring hearing aids. Subjects with various degrees of hearing loss (i.e., moderate to profound) used their personal hearing aids, which consisted of both linear and non-linear processing types (although specifics of non-linear processing was not mentioned).

Overall, Korczak et al. (2005) found that aided SCP responses were larger than unaided responses when stimuli were presented at the same input level. Averaged across all subjects with hearing loss, aided N1 amplitudes were 50% larger and latencies were 30 ms shorter than the unaided N1 results for both 65 and 80 dB SPL input levels. These differences were greatest in subjects with severe-to-profound sensorineural hearing loss and no measurable unaided SCP thresholds; thus, there seems to be an effect of degree of sensorineural hearing loss on the
magnitude of improvement in aided SCP responses. Compared with normal-hearing subjects, however, participants with hearing loss showed significantly longer aided N1 latencies at 80 dB SPL and a trend for longer aided N1 latencies at 65 dB SPL. These results were interpreted as suggesting that, despite their overall improvement in SCPs with hearing aids, individuals with sensorineural hearing loss still do not process speech stimuli with the same efficiency as their normal-hearing counterparts.

Researchers at the National Acoustic Laboratories in Australia have also conducted several aided SCP studies on infants with various degrees (moderate to profound) of hearing loss (Golding et al., 2007; Pearce et al., 2007; Purdy et al., 2005); however, there is little information regarding their aided SCP measures available in the literature. For example, in a review article by Purdy et al. (2005), a study on aided SCPs in infants with sensorineural hearing loss was described; however, the details of the study were not specified. They reported that aided SCPs were consistently present in infants with moderate sensorineural hearing loss and only present in half of the cases with profound hearing loss. Although these findings provide further support that SCPs can be elicited in aided conditions, these results show an opposite pattern to those described by Korczak et al. (2005). This suggests that aided SCPs may be variable across different hearing aids. Additionally, detailed reporting of only one subject with severe sensorineural hearing loss was described, although improvements in aided SCP waveform morphology positively correlated with behavioural testing conducted at a later date.

Although the ability to measure aided SCPs has been well documented, other researchers have sought to determine whether changes to hearing aid settings could be measured using SCPs. Tremblay et al. (2006) studied normal-hearing subjects because they believed that it would be
difficult to separate the effects of hearing loss and amplification. The purpose of their study was to determine whether 20-dB of hearing aid gain would alter SCP waveforms in response to the speech stimuli /si/ and /ʃi/ (with 756.30-ms and 654.98-ms durations, respectively; presented at 64 dB peak-to-peak equivalent (ppe) SPL). These specific speech stimuli were chosen because individuals with high-frequency hearing loss often have difficulty with high-frequency fricatives (Tremblay et al., 2003; Tremblay et al., 2006) and /si/ and /ʃi/ stimuli produce different neural responses in individuals with normal hearing (e.g., N1 response to /s/ is larger in amplitude than response to /ʃ/) (Tremblay et al., 2003). The digital hearing aids used in this study were omnidirectional, set to provide 12-15 dB of gain at 1000 Hz, 24-26 dB of gain at 2000 Hz and 24-26 dB of gain at 3000 Hz (as verified in a 2-cc coupler). The hearing aids had output limiting compression circuitry with a compression threshold of 75 dB SPL, 1-ms attack time, 50-ms release time, and were coupled with stock foam earmolds. Although Tremblay et al. (2006) found that distinct neural patterns elicited by different speech stimuli were evident in aided conditions they had also expected that adding 20-dB of hearing aid gain would have the same effect as increasing the stimulus intensity by 20 dB. On the contrary, their results showed no significant differences between aided and unaided SCPs in response to either speech stimulus. Specifically, adding 20-dB of gain resulted in no significant increase in N1 amplitude or decrease in N1 latency. They concluded that further research on the effects of amplification on neural response patterns was needed before slow cortical potentials could be used reliably in the hearing aid evaluation process.

A follow-up study by Billings et al. (2007) used a 1000-Hz tone with a total duration of 757-ms (7.57-ms rise/fall times) in lieu of speech stimuli to compare aided and unaided SCPs in normal hearing subjects. The digital hearing aid used in this study was omnidirectional, set to
provide 20-dB of gain from 250-5000 Hz (as programmed by the manufacturer’s module using NOAH software and verified using coupler measures). The hearing aid had a compression threshold of 65 dB, a compression ratio of 2:1, a 5-ms attack time, and a 30-ms release time. SCPs were measured over a series of input levels (30, 40, 50, 60, 70, 80, and 90 dB ppe SPL). Overall, Billings and colleagues found that as stimulus intensity was increased, the expected increase in amplitude and decrease in latency was observed in both aided and unaided conditions. Nevertheless, they also found that similar to the Tremblay et al. (2006) findings, adding 20-dB of hearing aid gain did not result in an amplitude increase or latency decrease compared with unaided SCPs at the same input level. The results from these two studies led this group of researchers to conclude that aided SCPs should be approached with caution as hearing aids seem to be changing more than just the intensity of the auditory stimulus. Unfortunately, neither Tremblay et al. (2006), nor Billings et al. (2007) reported any acoustic measures of hearing aid processed stimuli. Additionally, Billings et al. (2007) used a stimulus that is atypical for SCP recordings (which will be discussed in further detail below), and it is possible that this may have negatively affected SCP peak amplitudes.

An attempt to explain the lack of observable differences between unaided and aided SCP amplitudes in response hearing aid gain was made by Billings, Tremblay, Stecker, and Tolin (2009), who examined the role of the SNR on amplitude and latency of SCP components. They found that, in fact, the SNR and not the absolute signal level determined N1 amplitudes and latencies such that larger amplitudes and shorter latencies were recorded as the SNR increased. Conversely, N1 amplitudes and latencies did not differ across conditions if the absolute signal intensity and background noise level were increased together with no resulting change in SNR. Billings et al. (2009) believed that if the SNR was maintained across aided and unaided
conditions in their previous research (e.g., Billings et al., 2007; Tremblay et al., 2006), this might be the reason N1 amplitudes were not larger in the aided conditions compared with unaided.

Despite conflicting results regarding the validity of using SCPs to measure hearing aids, the National Acoustic Laboratories, in conjunction with Frye Electronics (Tigard, OR) has developed the “NALHEARLab Cortical Evoked Potential Analyzer” (Frye Electronics, 2007). This device is equipped with threshold evaluation and aided SCP modules which test with low-, mid-, and high-frequency speech stimuli (/m/, /g/ and /t/) with the expectation/belief that if a child’s brain shows different responses to different speech sounds which span the frequency range at a conversational level, they have the capability to develop speech and language (Agung et al., 2005; Ling, 1976, 2002). Despite a lack of normative data and empirical evidence to support the use of the SCP for hearing aid validation, this device is currently in production.

Due to the need for a reliable objective validation measure SCPs have become the current AEP of interest for this purpose (Dillon, 2005). As this is the focus of the current study, a more detailed review of SCPs is necessary.
THE SLOW CORTICAL POTENTIAL

Discovery

The slow cortical potential was first detected by P.A. Davis (1939) in the ongoing electroencephalographic recordings of waking humans. The similarity of response morphology and scalp distribution across modalities (Picton, Hillyard, Krausz & Galambos, 1974) and interactions between stimuli across modalities (Davis, Osterhammel, Wier & Gjerdingen, 1972; Walter, 1964), led some to believe that the response was not modality specific (i.e., it did not arise in brain areas specific to processing of auditory or visual (or other modality); rather it was suggested to originate in multi-sensory areas; it subsequently became known as the “vertex potential”. Possible generator sources were believed, at that time, to be (i) the frontal cortex (Picton et al., 1974; Walter, 1964), (ii) regions of the thalamus (Fruhstorfer, 1971), (iii) the brainstem (Saletu, Itil & Saletu, 1971), and/or (iv) the cingulate gyrus (Chatrian, Canfield, Knauss, & Lettich, 1975). In 1970, Vaughn and Ritter were the first to propose that the generators of the auditory vertex response were specific to auditory system and were located bilaterally on the superior-temporal lobe in the auditory cortices (Vaughn & Ritter, 1970). Since then, it has largely been shown that: (i) similarities between scalp distributions may not indicate common generators (Scherg & von Cramon, 1985, 1986a), (ii) the auditory vertex response is generated within the auditory cortex (Scherg & von Cramon, 1985, 1986a, 1986b; Vaughn & Ritter, 1970), and (iii) the vertex responses which occur for stimuli of different modalities have generator sources in areas of the brain other than the auditory cortex (e.g., visual evoked potentials arise from generators in the visual areas of the occipital lobe) (Adrian & Matthews, 1934; Ciganek, 1961; Hirsch, Pertuiset, Calvert et al., 1961). The term vertex potential is therefore not an appropriate term as it refers to an erroneous belief that the response is not specific to the auditory system. Since then, this cortical potential has been called by several
names, including the cortical auditory evoked potential (CAEP), the slow cortical potential (SCP), the late vertex response (LVR), the long latency response (LLR), and the P1-N1-P2 complex. The term cortical auditory evoked potential (CAEP) is a broad term which has been used to encompasses both slow cortical potentials (e.g., the P1-N1-P2 complex) and late cortical evoked potentials (e.g., mismatch negativity) (Damaschke, Riedel & Kollmeier, 2005; Picton & Hillyard, 1988; Sharma & Dorman, 1999); however, it has also been used interchangeably with the term slow cortical potential, especially when referring to the infant slow cortical potential (which shows different waveform morphology than the adult slow cortical potential) (Golding et al., 2007; Gravel et al., 1989; Pearce et al., 2007). A more commonly accepted name for this potential is the slow cortical potential, which is named thus because the pattern of responses is generated by the cortex and takes longer following stimulus onset (50 to 300 ms) compared with the auditory brainstem response (ABR) which occurs between 0 and 20 ms and the middle latency response (MLR) which occurs between 10 and 100 ms (Stapells, 2009).

The SCP can be elicited by a variety of signals, including tonal and speech stimuli (Billings et al., 2007; Golding et al., 2007; Korczak et al., 2005; Oates et al., 2002; Pearce et al., 2007; Purdy et al., 2005; Rapin & Graziani, 1967; Tremblay et al., 2006), and is composed of a series of positive (P) and negative (N) waves numbered according to their position in the response waveform. P1 is a positive peak approximately 50 ms post-stimulus onset, N1 is a negative peak approximately 80 to 100 ms post-stimulus onset, and P2 is a positive peak approximately 180 to 120 ms post-stimulus onset (Davis, Mast, Yoshie, & Zerlin, 1966). As a result, the SCP is also referred to as the P1-N1-P2 complex. Alternatively, these peaks and troughs have been named for their polarity and the latency at which they normally occur. In this case, the N100 (i.e., negative wave at 100 ms) and P175 (i.e., positive wave at 175 ms) would be
equivalent to the N1 and P2 waveforms respectively (Naatanen & Picton, 1987, Scherg & von Cramon, 1985, 1986a; Vaughan & Ritter, 1970; Wolpaw & Wood, 1982; Wood & Wolpaw, 1982). For the purposes of this thesis, the term slow cortical potential (SCP) will be used to refer to the P1-N1-P2 complex and when appropriate, the specific components of the complex will be discussed directly.

**Generator and Lesion Studies**

During the time of the debate regarding the specificity of the auditory vertex response, many researchers sought to discover the generators of this response. Research involving intracranial recordings in animal (Arezzo, Pickoff & Vaughan, 1975, Hardin & Castellucci, 1970) and human subjects (Celesia, 1976; Celesia & Puletti, 1971) revealed that several areas of the cortex were active at the time when the N1 response occurs, including: (i) the superior aspect of the temporal lobe, (ii) the reticular formation, (iii) the temporal lobe, and (vi) the ventral lateral nucleus of the thalamus. This information led researchers to question whether there were multiple generators of the N1 response. Using multi-electrode scalp recordings to conduct dipole source analyses, Scherg and von Cramon (1985, 1986a, 1986b) reported two generator sources in each hemisphere. The first source was generated by a tangentially (i.e., vertically) oriented dipole in the auditory cortex corresponding to the N1 first described by Vaughn and Ritter (1970). This waveform is optimally recorded at the vertex and shows a negative peak at 100 ms (N1) followed by a positive peak at 180 ms (P2) which inverts for electrode locations below the auditory cortex (Scherg & von Cramon, 1985, 1986a; Vaughan & Ritter, 1970). The second source, also known as the T-complex, was generated by a radial (i.e., horizontal) dipole in the secondary auditory cortex. This waveform is optimally recorded in the midtemporal region.
and shows a positive peak at 100 ms (Ta) and a negative peak at 150 ms (Tb) (Scherg & von Cramon, 1985, 1986a; Wolpaw & Penry, 1970, 1977).

Further evidence of these four generator sources has been provided by studies involving temporal lobe damage. More specifically, research has shown that bilateral lesions to the primary auditory cortex result in absent N1 responses (Adams, Rosenberger, Winter, & Zollner, 1977; Albert, Sparks, Stockert, & Sax, 1972; Graham, Greenwood & Lecky, 1980; Jerger et al., 1969; Lechevalier, Rossa, Eustache, Schupp, Boner, & Bazin, 1984; Michel, Perronet & Schott, 1980; Özdamar, Kraus & Curry, 1982; Shindo, Kaga & Tanaka, 1981) and unilateral lesions affecting the primary or secondary auditory cortex result in significantly attenuated P1-N1-P2 source potentials on the damaged side (Knight et al., 1980; Scherg & von Cramon, 1986a).

Additionally, Scherg and von Cramon (1986a) found that unilateral lesions affecting the auditory radiation fibres but not the primary auditory cortex resulted in no attenuation of P1-N1-P2 dipole source potentials. These three findings provide support for the location of an N1 generator in the primary auditory cortex. In this same study, Scherg and von Cramon (1986a) reported that lesions affecting the secondary auditory cortex but not the primary auditory cortex resulted in normal tangential components and significant attenuation of radial components. This finding provides further support for the location of the N1 generator in the primary auditory cortex as well as for the location of the N1 generator in the secondary auditory cortex. One limitation of the dipole modelling technique used by Scherg and von Cramon (1985, 1986a, 1986b) is that it is difficult to distinguish between different generator sources if the scalp distributions of these sources overlap. As a result, although the bulk of the research on temporal lobe lesions support the presence of two generator sources, there are some that believe in the existence of a third generator. More specifically, several clinical studies have reported that bilateral lesions
involving the primary auditory cortex did not cause the expected decrease in N1 amplitudes mentioned above (Parving, Solomon, Elberling, Larsen, & Lassen, 1980; Rosati, Bastiani, Paolino, Prosser, Arslan, & Artioli, 1982; Woods, Knight & Neville, 1984). These findings suggest that either: (i) a third component of the N1 response has generator sources outside of the primary auditory cortex, or (ii) the lesions may not have been complete (i.e., not affecting the entire auditory cortex). Recordings of magnetic fields that occur in unison with scalp recorded potentials have revealed that an additional component is active only when interstimulus intervals greater than 4 ms are used (Hari, Kaila, Katila, Tuomisto, & Varpula, 1982). This third component is most likely generated in the frontal cortex and shows a negative peak at 100 ms (Hari et al., 1982; Naatanen & Picton, 1987; Velasco, Velasco & Olvera, 1985; Velasco & Velasco, 1986).

**Functional Significance**

Although the N1 response is composed of three underlying components and generator sources, all three components respond to abrupt changes in auditory stimuli and are thought to indicate the arrival of stimulus energy at the cortical level or, in other words, the presence of an audible stimulus (Naatanen & Picton, 1987). In order to detect a change in stimulus energy, any preceding energy must have remained constant for some duration of time (Naatanen & Picton, 1987). Although some have suggested that differential neural response patterns elicited by different stimuli may give some indication of discrimination (Pearce et al., 2007; Purdy et al., 2005; more), presence of the N1 response only indicates that the stimulus is detectable by the auditory cortex. This is necessary for subsequent discrimination and the two may be highly correlated (Martin, Kurtzberg & Stapells, 1999; Martin, Sigal, Kurtzberg, & Stapells, 1997). Nevertheless, the presence of N1 is not a direct indication of discrimination.
Stimulus Factors

Many stimulus parameters have marked effects on SCP amplitudes and latencies; therefore, these will be discussed in further detail in order to understand how the previously described characteristics of electroacoustic test signals will affect SCPs.

Stimulus rise/fall times and duration

The most important determinant for N1 response presence is the rise time of the stimulus, as the N1 has been found to occur in response to change in stimulus energy from some baseline value (Alain, Woods & Covarrubias, 1997; Kodera, Hink, Yamada & Suzuki, 1979, Onishi & Davis, 1968). As a result, the N1 response is often described as an “onset” response (Goldstein, Hall & Butterfield, 1968), and characteristics of the stimulus onset will affect N1 latency and amplitude. The largest increases in N1 amplitude will occur as rise time increases up to 30 ms (Alain et al., 1997; Kodera et al., 1979, Onishi & Davis, 1968). When stimulus rise times are increased beyond 30 ms, N1 amplitude shows a gradual decline until the rise time is 50 ms, beyond which significant decreases in amplitude occur (Onishi & Davis, 1968). Furthermore, increasing duration of the stimulus will result in increased N1 amplitudes up to total durations of 30-50 ms (Gage & Roberts, 2000; Joutsiniemi, Hari & Vilkman, 1989; Kodera et al., 1979; Onishi & Davis, 1968). It follows that any changes to the stimulus as a result of hearing aid processing during this time could potentially affect N1 amplitude or may not be reflected at all in the SCP if hearing aid processing takes longer than the duration of the stimulus.

Frequency

The effect of frequency on the N1 response depends on the intensity of the stimulus (Antinoro & Skinner, 1968; Antinoro, Skinner & Jones, 1970; Davis et al., 1966). Near
threshold or when loudness is maintained at a constant level across frequencies (i.e., different frequencies will have different dB SPL levels (Hyde, 1997)), N1 amplitude will not change as a function of frequency (Antinoro & Skinner, 1968; Davis et al., 1966); however, if the intensity of the stimulus is kept constant across frequencies (i.e., all frequencies have same dB SPL level), there will be a significant effect of frequency on the N1 amplitude, such that N1 amplitude decreases as stimulus frequency increases, particularly for frequencies above 2000 Hz (Antinoro et al., 1970).

**Interstimulus interval (ISI)**

The rate of stimulus presentation has a significant effect on N1 amplitude. Studies have shown that N1 amplitude increases with increasing ISI, rapidly between 0.5 and 3 s and then more gradually up to ISIs of 10 s (Appleby, 1964; Davis et al., 1966; Lightfoot & Kennedy, 2006; McCandless & Best, 1964; Milner, 1969; Picton, Woods, Baribeau-Braun, & Healey, 1977; Ritter, Vaughan & Costa, 1968; Zerlin & Davis, 1967). This effect seems to be less pronounced at low stimulus intensities (Naatanen & Picton, 1987; Picton, Goodman & Bryce, 1970). Furthermore, it has been well documented that the N1 response shows a rapid decline in amplitude with stimulus repetition (Bourbon, Will, Gary, & Papanicolaou, 1987; Budd, Barry, Gordon, Rennie, & Michie, 1998; Davis et al., 1966; Fruhstorfer, 1971; Fruhstorfer, Soveri & Jarvilehto, 1970; Lightfoot & Kennedy, 2006; Prosser, Arslan & Michelini, 1981; Ritter, Vaughan & Costa, 1968). More specifically, when subjects are presented with a train of identical stimuli, most studies have reported a 50% reduction in N1 amplitude by the second stimulus. There have been two different explanations for this rapid decline in N1 response amplitude. The first possibility is habituation (i.e., a reduction in response amplitude that results from a loss of stimulus novelty). The second possibility is sensory refractoriness, which has...
been supported by the fact that the N1 response is sensitive to ISI. In other words, the cells in the cortex are simply not allowed enough time to recover when the ISI is too short (Budd et al., 1998). As a result, N1 amplitudes are smaller when the ISI is short and as the ISI becomes longer, N1 amplitudes increase because the cortical neurons have longer to recover. This evidence is further supported by the fact that presenting a new stimulus of a different frequency will result in large N1 amplitudes as the cortical units activated by the new and old stimuli belong to different neural populations (Butler, 1968). Although it has been shown that N1 response amplitudes increase up to ISIs of 10 s, the optimal ISI when both amplitude and duration of test time are considered is between 1-2 s (Davis & Zerlin, 1966; Davis et al., 1966; Lightfoot & Kennedy, 2006; Picton et al., 1977).

**Stimulus intensity**

Increasing stimulus intensity results in increased N1 amplitudes and decreased N1 latencies (Beagley & Knight, 1967; Davis et al., 1966; Picton et al., 1977; Rapin, Schimmel, Tourk, Krasnegor, & Pollak, 1962). N1 amplitude tends to saturate at high intensities (Picton et al., 1970; Picton et al., 1974). However, when blocks of stimuli with longer ISIs (>4 s) are presented, much less saturation occurs at high intensity levels (Gilles, Bottcher & Ullsperger, 1986). It is possible that the non-specific component of the N1 potential is responsible for these differential findings because it occurs only for stimuli with long ISIs (>4 s) (Hari et al., 1982). This theory is at least partially supported by the fact that Budd et al. (1988) found that the three N1 components have different refractory periods, with the non-specific component showing the longest recovery time and therefore requiring the longest ISI.
Current Clinical Applications of the SCP

The SCP (with ideal stimulus, subject, and recording conditions) is a sensitive measure of auditory thresholds; SCP thresholds typically occur within 5-10 dB of behavioural audiometric thresholds (Beagley & Kellogg, 1969; Kaf et al., 2006; Lightfoot & Kennedy, 2006; Tomlin, Rance, Graydon, & Tsialios, 2006). SCP amplitude and latency become variable with sleep, thus subjects are normally required to remain awake but quiet during testing. As it can be difficult to keep infants awake and quiet, use of the SCP for threshold testing in infants has largely been abandoned; however, it remains the gold-standard AEP measure in adults (Martin, Tremblay & Stapells, 2007; Stapells, 2002). More recently, researchers have studied suprathreshold applications of slow cortical potentials in children, including children with sensorineural hearing loss or auditory neuropathy spectrum disorder (ANSD) (Kumar & Jayaram, 2005; Pearce et al., 2007; Rance, Cone-Wesson, Wunderlich, & Dowell, 2002). Additionally, as discussed above, the P1-N1-P2 complex is being considered for hearing aid fitting and evaluation in infants.
RATIONALE FOR THIS STUDY

A system for measuring slow cortical potentials is currently being marketed for the purpose of validating infant hearing aid fittings (Frye Electronics, 2007). This is problematic because SCP studies related to this purpose have produced mixed results; thus it is not known whether SCPs provide an accurate measure of cortical response to signals processed by hearing aids, particularly hearing aids with digital signal processing. Several studies have shown that SCPs can be reliably recorded in aided conditions (Billings et al., 2007; Golding et al., 2007; Gravel et al., 1989; Korczak et al., 2005; Purdy et al., 2005; Rapin & Graziani, 1967; Tremblay et al., 2006), and some have reported that stimuli can reliably be differentiated in the SCP response in aided conditions (Purdy et al., 2005; Tremblay et al., 2006); however, there is evidence to suggest that changes to hearing aid settings (e.g., gain) cannot be measured by changes in the SCP response (Billings et al., 2007; Tremblay et al., 2006). The stimulus used by Billings et al. (2007) was atypical for SCP testing, which may have affected the outcome of their study, and acoustic recordings of their hearing aid processed stimuli were not published.

Clearly, it is important to quantify the acoustic effects of the hearing aid processing on the test signal; otherwise, the stimulus used to evoke the SCP in aided conditions is not known. This idea is supported by studies which found that due to hearing aid processing, stimuli used to measure some auditory evoked potentials may not result in valid measurements of hearing aid gain or output (Brown et al., 1999; Gorga et al., 1987). Overall, this is problematic because using SCPs for hearing aid validation requires that changes to gain settings can be measured, particularly in infants who cannot verbally acknowledge whether a change was audible.

Due to conflicting evidence regarding the reliability of SCPs for hearing aid validation purposes, the primary aim of the current thesis was to determine whether the SCP accurately
reflected changes to the acoustic stimulus caused by hearing aid processing. This was accomplished by examining the results obtained in two studies: (i) Study A compared N1-P2 amplitudes and N1 latencies in unaided and aided conditions, using a stimulus designed to elicit optimum N1 amplitudes in accordance with previous research (e.g., Alain et al., 1997; Kodera et al., 1979; Onishi & Davis, 1968); and (ii) Study B assessed whether hearing aid processed stimuli were altered in such a way that is consistent with/explains the cortical results. This was accomplished by investigating the acoustic effect of hearing aid processing on the stimulus used for SCP testing, as well as on the longer stimulus used by Billings et al. (2007).
Chapter 2: Slow Cortical Potential Measures of Amplification
INTRODUCTION

With the advent of Universal Newborn Hearing Screening programs, it has become increasingly common for infants to be fit with hearing aids by six months of age, which requires that reliable methods are in place for fitting hearing aids in this young population (Ching & Dillon, 2003; Joint Committee on Infant Hearing, 2007; Seewald & Scollie, 2003; Tharpe, Sladen, Huta, & Rothpletz, 2001; Yoshinaga-Itano, 2003). The hearing-aid fitting process involves several stages, including first the assessment of hearing loss and relevant acoustic variables, and then selection, fitting, verification, and validation of the hearing aid fitting (Joint Committee on Infant Hearing, 2007; Scollie & Seewald, 2001; Valente, Abrams, Benson, Chisolm, Citron, Hampton, et al., 2006). Test procedures used in the hearing aid fitting process are categorized in two ways: (i) behavioural measures are those that require active participation by the patient (e.g., pure tone, speech testing and self-report questionnaires) and (ii) objective measures, which require no subjective responses from the patient (e.g., real-ear electroacoustic and evoked potential measures). The final stage in hearing aid fitting is the validation stage, which begins immediately after verification and continues for many years afterwards (Valente et al., 2006). During this stage of the fitting process it is often necessary to adjust hearing aid settings in accordance with individual preferences for sound quality and comfort (Byrne, 1986; Leijon, Eriksson-Mangold, & Bech-Karlson, 1984; Leijon, Lindkvist, Ringdahl, & Israelsson, 1990). In adults, this is often accomplished behaviourally using verbal feedback and/or written feedback through questionnaires such as the Client Oriented Scale of Improvement (Dillon, James, Ginis, 1997) and the Hearing Aid Performance Inventory (Walden, Demorest & Hepler, 1984). Unfortunately, there is a limited array of validation measures for infants who cannot participate in conventional audiometric and/or speech recognition testing and as a result, it is difficult to ascertain that hearing aids are providing benefit for speech recognition, while
maintaining loudness comfort (Damarla & Manjula, 2007; Golding, Pearce, Seymour, Cooper, Ching, & Dillon, 2007). Questionnaires such as the Parent’s Evaluation of Aural/Oral Performance in Children (PEACH) are often relied upon in these instances to provide information regarding sound audibility and comfort levels (Ching & Hill, 2007; Harrison, 2000). Caregiver questionnaires are dependent on caregiver presence and recollection of specific behaviours, and provide little information regarding the perception and/or discrimination of individual speech sounds (Golding et al., 2007; Purdy, Katsch, Dillon, Storey, Sharma, & Agung, 2005). As a result, there is a strong need for reliable, objective measures of hearing aid validation in infants that can provide information regarding listening comfort and speech perception. For this purpose, auditory evoked potentials (AEP) have been considered in two primary ways: (i) to determine whether aided AEPs can be obtained at input levels where unaided AEPs could not (Gravel, Kurtzberg, Stapells, Vaughan, & Wallace, 1989; Kileny, 1982; Rapin & Graziani, 1967) and/or (ii) to determine whether changes in hearing aid settings and/or stimuli can be measured using AEPs (Billings, Tremblay, Souza, & Binns, 2007; Golding et al., 2007; Korczak, Kurtzberg & Stapells, 2005; Pearce, Golding & Dillon, 2007; Picton, Durieux-Smith, Champagne, Whittingham, Moran, Giguère, & Beauregard, 1998; Purdy et al., 2005; Stoebel, Swanepoel & Groenewald, 2007; Tremblay, Billings, Friesen, & Souza, 2006) and if so, whether these measures have behavioural correlates such as loudness growth (Hecox, 1983; Kiessling, 1982, 1983). Several AEPs have been assessed as potential objective validation measures, including the auditory brainstem response (ABR), the auditory steady-state response (ASSR) and the slow cortical potential (SCP).

Research on AEPs measured with hearing aids in place has yielded varying degrees of success. One reason for this variability may be due to the stimuli used to measure them. More
specifically, although there are many commercially available test signals that may be used to assess hearing aids electroacoustically and/or via AEPs (including tonal and complex stimuli), not all stimuli are appropriate for measuring hearing aid processing. For instance, complex stimuli such as speech-weighted composite noise provide a better estimate of gain for speech compared with tonal stimuli (Stelmachowicz, Lewis, Seewald, & Hawkins, 1990; Stelmachowicz, Kopun, Mace, & Lewis, 1996; Scollie & Seewald, 2002), whereas tonal stimuli provide a better measure of maximum power output (MPO) than do complex stimuli (Stelmachowicz et al., 1990). However, neither of these stimuli may be best for eliciting a given AEP and the stimuli used to elicit an AEP may not be best for measuring hearing aids (Brown, Klein & Snydee, 1999; Gorga, Beauchaine & Reiland, 1987; Levillain, Garcon & Le Her, 1979). For example, research on hearing aid processed stimuli has revealed that the click and brief-tone stimuli used in ABR testing are too short to activate the compression processing and steady-state response of the hearing aid (Brown et al., 1999; Gorga et al., 1987). In light of these findings, researchers have considered the 80-Hz ASSR as a solution to the stimulus problem (e.g., Damarla & Manjula, 2007; Picton et al., 1998; Stoebel, Swanepoel & Groenewald, 2007). For the purpose of hearing aid validation, the potential advantage of ASSR stimuli over clicks and brief tones is that they are continuous, steady-state stimuli with low crest factors, which allow the hearing aid to settle into its steady state and are less likely to result in distortion from the sound source (Picton et al., 1998). However, several studies suggest that although continuous stimuli are used to elicit the 80-Hz ASSR, the responses likely reflect only the initial portion of the stimulus, much like the ABR (Mo & Stapells, 2008). Thus, the 80-Hz ASSR may be subject to the same limitations as the ABR (Stapells, Herdman, Small, Dimitrijevic, & Hatton, 2005) and shows poorer frequency specificity than indicated by the acoustics (Herdman, Picton & Stapells, 2002; Herdman & Stapells, 2003). Both the ABR and 80-Hz ASSR share the limitation of being
brainstem responses and, as a result, do not give any indication of higher-level (cortical) processing, which is likely important for hearing aid validation (Korczak et al., 2005). Slow cortical potentials have an advantage over brainstem AEPs because they originate in the auditory cortex (Scherg & von Cramon, 1985, 1986a, 1986b) and can be elicited by a variety of signals, including tonal stimuli longer than those used for the ABR and speech stimuli (Martin, Tremblay & Stapells, 2007; Stapells, 2009). As a result, the SCP is now of greatest interest for hearing aid validation purposes (Dillon, 2005).

Even though there are many potential advantages of using SCPs for hearing aid validation, some studies have supported its use for this purpose, whereas others have questioned it. For example, several studies using subjects with various degrees, types, and configurations of hearing loss and their personal hearing aids have shown that SCPs show some promise in the hearing aid validation process using a variety of stimuli, including tonal and complex (Dillon, 2005; Golding et al., 2007; Gravel, Kurtzberg, Stapells, Vaughan and Wallace, 1989; Korczak et al., 2005; Pearce et al., 2007; Purdy et al., 2005; Rapin & Graziani, 1967). Findings such as these have led a group of researchers at the National Acoustic Laboratories to develop a new device being marketed as a hearing aid evaluation tool for infants and children (Frye Electronics, 2007; Purdy et al., 2005; Golding et al., 2007; Pearce et al., 2007). In contrast, some recent studies on SCPs in subjects with normal hearing using speech and tonal stimuli have found that hearing aid gain cannot accurately be measured using SCPs, such that there were no significant differences between unaided and aided N1-P2 amplitudes (Billings et al., 2007; Tremblay et al., 2006). Billings, Tremblay, Stecker, and Tolin (2009), have since examined the effect of SNR and as a result, believe the SNR may have been maintained across aided and unaided conditions.
in their previous research (e.g., Billings et al., 2007; Tremblay et al., 2006) so that N1 amplitudes were not larger in the aided conditions compared with unaided.

The stimuli used by Billings et al. (2007) were atypical for SCP stimuli in that they used a more rapid rise time than is optimal for eliciting the SCP and were much longer in duration than can be reflected by the SCP. Research has shown that N1 does not reflect stimulus changes beyond the first 50 ms (Gage & Roberts, 2000; Joutsiniemi, Hari & Vilkman, 1989; Kodera, Hink, Yamada & Suzuki, 1979; Onishi & Davis, 1968). Also, rise times between 20 and 30 ms result in the largest N1 amplitudes (Alain, Woods & Covarrubias, 1997; Kodera et al., 1979; Onishi & Davis, 1968). Perhaps different results may have been obtained using a more typical stimulus duration, such as 60 ms (Lightfoot & Kennedy, 2006; Martin et al., 2007; Stapells, 2002; Van Maanen & Stapells, 2005). This possibility was addressed by this paper’s Study A, which examined whether hearing aid gain could be measured by SCPs in normal-hearing subjects using a 60-ms duration tonal stimulus with a 20-ms rise time. Additionally, it is not clear what effect hearing aid processing had on the stimulus used by Billings et al. (2007) as they did not provide any acoustic recordings of the hearing-aid processed stimuli. Study B addressed the effects of hearing aid processing on the 60-ms duration tonal stimulus used in Study A, as well as on the longer duration tonal stimulus used by Billings et al. (2007).
METHODS AND RESULTS

The following section is divided into: (i) methodology that is common to both studies, and (ii) methodology and results that are specific to each particular study.

Methods Common to Studies A and B

Subjects

A total of 18 subjects between the ages of 19 and 59 participated in these studies and four subjects participated in both. Subjects were briefed on the study procedures and provided informed written consent prior to participating. All subjects were screened for normal hearing by behavioural audiometry (Study A) and for normal middle/outer ear function by immittance audiometry (Studies A and B). Normal hearing was defined by pure-tone behavioural thresholds equal to or better than 15 dB HL from 500 to 4000 Hz and equal or better than 20 dB HL at 250 and 8000 Hz (ANSI, 1996). Normal tympanograms were defined by a single-peak static admittance between ±50 daPa in response to a 226-Hz probe tone (Fowler & Shanks, 2002). In study A, subjects were excluded if: (i) SCPs were absent in any of the unaided conditions, or (ii) SCPs were absent in three or more aided conditions. Two subjects (not included in the above numbers) were excluded from the study on the basis of these criteria.

Hearing aids

The same three behind-the-ear stock hearing aids, coupled with Comply snap tip 9-mm foam earmolds, were used for each participant: (i) Oticon E27 (“Analog”), (ii) Phonak Savia 211 dAZ (“DigitalA”) and (iii) Siemens Acuris S (“DigitalB”). Two digital hearing aids were selected because digital signal processing is currently the most commonly used technology; therefore, SCP results using digital hearing aids are the most clinically relevant. An analog
A hearing aid was selected to account for possible discrepancies between the more-recent studies using digital hearing aids (e.g., Billings et al., 2007; Golding et al., 2007; Purdy et al., 2005; Tremblay et al., 2006) and earlier studies primarily using analog hearing aids (e.g., Gravel et al., 1989; Korczak et al., 2005; Rapin & Graziani, 1967).

The digital hearing aids were programmed using Noah 3 software and the NoahLink programming assistant. Two gain settings (20 and 40 dB) were required; therefore, for each subject, two programs were created and the gain settings were verified by real-ear insertion gain (REIG) measures. The 20-dB gain setting was chosen to approximate the hearing aid setting used by Billings et al. (2007) and the 40-dB gain setting was added to assess whether additional gain would result in a significant difference between unaided and aided SCPs. Both programs were set with a 1:1 compression ratio across the frequency range and were verified for linear processing using input/output coupler measures. All additional hearing aid features such as digital noise reduction and feedback management were disabled. Settings for the digital instruments were saved in the Noah 3 software for each subject so that hearing aid programs could be recalled in follow-up sessions. Gain settings for the analog hearing aid were achieved by setting the volume control to one (minimum) and turning the dB SPL trim-pot until the REIG was 20 dB at 1000 Hz. To achieve the 40-dB gain setting, the volume control wheel was turned up until REIG equaled 40 dB at 1000 Hz. The volume control wheel was then marked for that setting. Unlike the digital hearing aids, gain settings for the Analog hearing aid had to be re-measured in follow-up sessions.

REIG was determined for each individual using the Fonix 7000 real-ear system. REIG was chosen because it equals the difference between unaided and aided responses in the ear
canal, which most closely approximates the comparisons made in these studies (i.e., SCPs and acoustic measures were conducted in both unaided and aided conditions). A small probe-tube (3-mm in diameter) was placed in the ear canal of the participant within 5 mm of the eardrum (verified by otoscopy). The tube was then marked at the tragal notch to ensure identical probe-tube placement across hearing aids. The gain control setting at 1000 Hz was adjusted until the appropriate REIG level was achieved for each program; other frequencies were set to provide the least amount of gain possible.

A swept pure tone with constant input level across the frequency range was used to measure REIG. A pure-tone stimulus was used, as this is the same type of stimulus used to elicit the SCPs. A 50 dB SPL input level was used to program both 20- and 40-dB REIG settings and a 70 dB SPL input level was used to verify the 20-dB REIG settings at higher input levels. These measures provided further verification that all three hearing aids were providing linear processing.

**Stimuli**

A 1000-Hz stimulus of 60-ms total duration (including a 20-ms rise/fall time) was used for both studies. A 20-ms rise/fall time was chosen because it is suitable for generating a large N1 response and the 60-ms total duration was chosen because it has been shown that stimuli of longer duration do not result in increased N1 amplitudes (Alain et al., 1997; Kodera et al., 1979, Onishi & Davis, 1968). An additional 757-ms stimulus, which was similar to that used by Billings et al. (2007), was used in Study B.² Stimuli were presented with offset-to-onset interstimulus intervals (ISI) of 940 ms. Stimuli generated by Neuroscan’s Stim2 software were

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² The 757-ms stimulus used in the Billings et al. (2007) study was not used for Study A because pilot testing revealed that the 757-ms duration stimulus used in their study and the 60-ms duration stimulus used in this study results in similar N1 amplitudes.
further amplified by a Wavetek Rockland 852 filter (providing 20-dB of amplification below 3000 Hz), and routed through a Tucker Davis Technologies (TDT) PA5 attenuator and HB7 headphone driver, and finally to a speaker in the sound field placed at 1.5 meters from the subject at 0˚ azimuth. The stimulus output at 80 dB SPL was calibrated with a Larson-Davis sound level meter by measuring the level of a longer-duration 1000-Hz tone (2-s duration, 20-ms rise/fall time; equal in peak-to-peak amplitude to the 60-ms 1000-Hz stimulus) at the head height of the subject, 1.5 m from the speaker. Stimuli were presented at three intensities (30, 50 and 70 dB SPL). A maximum stimulus intensity of 70 dB SPL was chosen to limit the maximum output to 90 dB SPL, after hearing aid gain. For all conditions, the subject’s left ear was plugged with a deeply seated foam plug in order to reduce any contributions of responses resulting from stimulation to the non-test ear.

Procedure

Subjects were asked to complete two test sessions for each study in which they participated, lasting no longer than three hours each, and were given the choice of completing the sessions sequentially or on separate days. Procedures were approved by the University of British Columbia Behavioural Research Ethics Board. Subjects were screened for normal hearing (Study A) and for normal outer- and middle-ear function (Studies A and B). Immittance audiometry was also conducted in the second test session to ensure no changes across test sessions.

Following hearing aid programming, all testing was conducted in a double-walled sound-attenuating booth. Average noise levels in the sound-attenuated booth at .5, 1, 2, and 4 kHz were 12, 10, 10, and 12 dB SPL, respectively. There were 18 test conditions in Study A and 36 test
conditions in Study B (i.e., 18 for each of the short and long stimuli) and presentation order for each subject was randomly assigned prior to the test date. During testing, participants were asked to sit as still as possible while watching a movie of their choice in closed-captioning and no sound. Subjects sat in a reclining chair set in the upright position so that each participant was seated with their head above the chair back.

Study A: Slow Cortical Potentials Measured With and Without Hearing Aids

The purpose of Study A was to record slow cortical potentials to stimuli presented in the sound field with and without hearing aids in place. Of particular interest was whether there would be a difference between unaided and aided response amplitudes when hearing aids were set for 20 or 40 dB of gain and whether unaided and aided response amplitudes would be comparable for equivalent nominal output levels.

Subjects and Recording

Thirteen normal-hearing subjects participated in Study A (mean age: 25±5.5 years; 5 females). One electroencephalogram (EEG) channel was recorded from electrodes placed at Cz and M1. A second channel to monitor vertical eye movements and eye blinks (EOG) was recorded from electrodes over the left supraorbital ridge of the frontal bone and over the zygomatic bone under the left eye. A fifth electrode on the nape of the neck served as ground. Electrode impedances were maintained below 5000 Ohms. Recordings were made using Neuroscan Synamps2 and Scan 4.3 software. The EEG and EOG channels were amplified, filtered (1-30 Hz), and digitized (5000 Hz), using a 700-ms analysis time (including a 100-ms prestimulus baseline). Single-trial epochs were saved for offline processing, including: baseline

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3 Extra care was taken to ensure consistent head position as pilot testing revealed that ear-level acoustic recordings varied in intensity with small changes to subject position in the sound-field.
correction across the total sweep duration, artifact rejection (±100 µV in any of the channels, and ±75µV in the EOG channel), and averaging. The stimulus was presented in each test condition until at least 200 accepted trials were obtained. Epochs were averaged separately for each condition, and average data were divided into odd and even trials to serve as replications. Averages were baseline-corrected using the pre-stimulus interval.

**Data Analysis**

SCP measures of interest were N1-P2 peak-to-peak amplitude and N1 latency. N1 peak amplitude measures were determined at the largest negativity occurring before 200 ms and P2 peak amplitude measures were determined at the largest positivity within 100 ms of N1; N1 latency was measured at the centre of the peak. In cases of multi-peaked waveforms, amplitude measures were taken at the largest amplitude and N1 latency was taken as the midpoint of the two negative peaks. For a response to be “present”, N1 was required to be replicable across odd and even average waveforms. If responses were “absent”, a value of 0 µV was substituted as a reasonable estimate of amplitude (e.g., Korczak et al., 2005; Martin, Sigal, Kurtzberg, & Stapells, 1997; Martin, Kurtzberg & Stapells, 1999; Oates & Stapells, 1997; Oates, Kurtzberg & Stapells, 2002; Whiting, Martin & Stapells, 1998). Latencies were not estimated for no-response results. Due to absent responses for 5 of 13 subjects in some of the 30-dB SPL conditions, latency results for this input level were excluded from statistical analyses.

**Statistical analysis**

For amplitude measures, four repeated-measures analyses of variance (ANOVA) were conducted: (i) to measure the effect of the 20-dB gain setting, a two-way repeated-measures ANOVA was conducted comparing four levels of hearing aid type (unaided, Analog, DigitalA,
and DigitalB) and three input levels (30, 50 and 70 dB SPL); (ii) to measure the effect of the 40-
dB gain setting, a two-way repeated-measures ANOVA was conducted comparing four levels of
hearing aid type (unaided, Analog, DigitalA, and DigitalB) and two input levels (30 and 50 dB
SPL); (iii) to compare 20- and 40-dB gain settings, a three-way repeated measures ANOVA was
conducted comparing two levels of gain (20 and 40 dB), three levels of hearing aid type
(Analog, DigitalA and DigitalB) and two input levels (30 and 50 dB SPL), and (iv) to compare
results for a 70 dB SPL “nominal output” (i.e., input intensity plus gain), a one-way repeated-
measures ANOVA with seven “conditions” was conducted (i.e., unaided 70 dB SPL input level
condition, 50 dB SPL input level plus 20-dB gain condition for the Analog, DigitalA and
DigitalB hearing aids, and 30 dB SPL input level plus 40-dB gain condition for the Analog,
DigitalA and DigitalB hearing aids).

For latency measures, the following three repeated-measures ANOVAs were conducted:
(i) to measure the effect of the 20-dB gain setting, a two-way repeated-measures ANOVA was
conducted comparing four levels of hearing aid type (unaided, Analog, DigitalA, and DigitalB)
and two input levels (50 and 70 dB SPL), (ii) to measure the effect of the 40-dB gain setting, a
one-way repeated-measures ANOVA was conducted (because the 30 dB SPL input level was
excluded) comparing four levels of hearing aid type (unaided, Analog, DigitalA, and DigitalB) at
one input level (50 dB SPL), and (iii) to make comparisons between the 20- and 40-dB gain
settings, a one-way repeated-measures ANOVA was conducted comparing three levels of
hearing aid type (Analogue, DigitalA and DigitalB) at 50 dB SPL. Analysis of “nominal output”
was not conducted for latency measures due to the absence of latency results for the 30 dB SPL
input level and thus the exclusion of the 40-dB gain condition.
For all analyses in Study A, main effects and interactions were considered significant if \( p < .05 \). Huyn-Feldt epsilon (\( \varepsilon \)) correction factors for repeated-measures were applied to the degrees of freedom and are reported where appropriate. Neuman-Keuls post-hoc analyses were performed for significant main effects or interactions. Post-hoc analyses were considered statistically significant if \( p < .05 \).

**Results**

Grand-mean waveforms for unaided compared with 20- and 40-dB of hearing aid gain for three hearing aids (Analog, DigitalA, Digital B) across three input levels (30, 50, 70 dB SPL) are provided in Figure 2.1. Mean amplitude and latency results (±1 standard deviation) for each condition are presented in Figure 2.2.

**Unaided compared with 20-dB gain condition.** In Figures 2.1 and 2.2, it is apparent that any differences between unaided response amplitudes and aided response amplitudes in the 20-dB gain condition are quite small; nevertheless, results of the ANOVA revealed a significant interaction between hearing aid type and input level \([F(6, 72) = 2.48, \varepsilon = .78, p = .046]\). Neuman-Keuls post-hoc analysis showed that N1-P2 amplitudes in the DigitalB condition were significantly smaller at 50 and 70 dB SPL input levels compared with the other aided conditions as well as the unaided; however, there were no significant differences between N1-P2 amplitudes at 30 dB SPL with or without hearing aids. At 70 dB SPL, the only hearing aid condition that resulted in larger N1-P2 amplitudes (compared to unaided) was the Analog aid. There were also significant main effects for hearing aid type \([F(3,36) = 6.99, \varepsilon = .95, p < .001]\) and for input level \([F(2, 24) = 74.97, \varepsilon = 1.00, p < .001]\). These findings indicate that the only 20-dB gain condition in which N1-P2 amplitudes were larger than unaided N1-P2 amplitudes occurred in the analog...
condition; even this result was not consistent across input levels. Additionally, response amplitudes in the DigitalB condition were significantly smaller than in all other aided and unaided conditions.

Figure 2.2 shows that N1 latencies in both digital hearing aid conditions appear longer compared with analog and unaided conditions. This was confirmed by ANOVA results, which revealed a significant main effect of hearing aid type \([F(3, 36) = 15.33, \varepsilon = 1.00, p < 0.001]\). Post-hoc analysis showed that both digital hearing aids resulted in significantly delayed N1 latencies compared with unaided responses. In contrast, there was no significant difference between unaided and aided N1 latencies in Analog hearing aid conditions. There was also a significant main effect of input level \([F(1, 12) = 47.88, \varepsilon = 1.00, p < .001]\), such that response latencies were longer for the 50 compared with the 70 dB SPL input level. There was a non-significant trend for latencies to be the same in unaided and Analog hearing aid conditions for both input levels, and for latencies to be longer for both digital hearing aids compared with Analog and unaided conditions for the interaction between hearing aid type and input level \([F(3, 36) = 2.42, \varepsilon = .817, p = .096]\).
Figure 2.1. Grand-mean waveforms (n=13) from electrode Cz for unaided and aided conditions (Analog, DigitalA and DigitalB) with two gain settings (20- and 40-dB) at three input levels (30, 50 and 70 dB SPL).
Figure 2.2. Mean amplitude and latency data (n = 13) for unaided and aided conditions (Analog, DigitalA and DigitalB) with two gain settings (20- and 40-dB) at three input levels (30, 50 and 70 dB SPL).
Unaided compared with 40-dB gain condition. In Figures 2.1 and 2.2, N1-P2 amplitudes are larger in the 40-dB gain conditions compared with unaided N1-P2 amplitudes at 50 and 70 dB SPL input levels; however at 30 dB SPL, there appears to be no difference between N1 amplitudes across conditions. Results from the ANOVA revealed a significant main effect of hearing aid type \([F(3,36) = 7.58, \varepsilon = 1.00, p < .001]\) and input level \([F(1, 12) = 115.26, \varepsilon = 1.00, p < .001]\), as well as a significant interaction between hearing aid type and input level \([F(3, 36) = 5.42, \varepsilon = .96, p = .004]\). Post-hoc analysis showed that N1-P2 amplitudes were significantly smaller in DigitalB aided conditions compared with Analog and DigitalA. Additionally, post-hoc analysis confirmed that N1-P2 amplitudes were larger for all three hearing aids at 50 dB SPL compared with unaided; however, no significant differences existed between N1-P2 amplitudes in aided and unaided conditions at the 30 dB SPL input level and no significant differences existed between hearing aids at 30 dB SPL. Thus, despite being set with 40-dB of gain, N1-P2 amplitudes were never significantly larger for any of the aided conditions compared with unaided at 30 dB SPL.

Similar to findings for the 20-dB gain setting, N1 latencies for the 40-dB gain setting appear longer in the digital hearing aid conditions relative to the analog and unaided conditions. This seems to be the case for both 30 and 50 dB SPL input levels; however, as mentioned previously, the latencies for the 30 dB SPL input level was not included in the statistical analysis. The ANOVA results revealed a statistically significant difference between hearing aid conditions \([F(3, 36) = 5.57, \varepsilon = 1.00, p = .003]\). As is evident in Figure 2.2, there was a trend for both digital hearing aid conditions to be delayed compared with unaided and analog hearing aid conditions; however, N1 latencies were only significantly longer for the DigitalA condition and
there was no significant difference between response latencies for the unaided or other hearing aid conditions.

20-dB compared with 40-dB gain condition. In order to determine whether hearing aid gain had an effect on response amplitudes and/or latencies, results for the two gain settings (20- and 40-dB) were compared across input level (30 and 50 dB SPL for amplitude and 50 dB SPL for latency). It is evident in Figure 2.2 that N1-P2 amplitudes are larger in the 40-dB compared with the 20-dB gain setting for all hearing aid types. Results from the ANOVA revealed a significant main effect for gain [F(1, 12) = 73.18, ε = 1.00, p < .001] and for input level [F(1, 12) = 118.13, ε = 1.00, p < .001], as well as a significant interaction between gain setting and input level [F(1, 12) = 7.32, ε = 1.00, p = 0.019]. Post-hoc analysis showed significantly larger N1-P2 amplitudes for the 40-dB gain setting compared with the 20-dB gain setting at 50 dB SPL; however, no significant difference existed between N1-P2 amplitudes for the two gain settings at the 30 dB SPL input level. The main effect for hearing aid type was also significant [F(2, 24) = 6.01, ε = 1.00, p = .008], such that N1-P2 amplitudes were significantly smaller for DigitalB hearing aid condition compared with Analog and DigitalA hearing aid conditions. There was no significant interaction between hearing aid type and gain setting [F(2, 24) = 1.02, ε = 1.00, p = .37], hearing aid type and input level [F(2, 24) = 1.73, ε = 1.00, p = .20], or hearing aid type by gain setting by input level [F(2, 24) = .41, ε = 1.00, p = .67]. These results are consistent with findings reported above that there is little to no effect of gain for the 30 dB SPL input level, but a significant effect of gain for the 50 dB SPL input level.

Latency results for the 20- and 40-dB gain settings in Figure 2.2 show that, once again, there is an obvious difference between N1 latencies across digital and analog aided conditions,
such that N1 latencies are longer in the digital hearing aid conditions (particularly for the 20-dB gain setting); however, there does not seem to be an effect of gain on latency for any hearing aid. The ANOVA confirmed a significant main effect for hearing aid type \[ F(2, 24) = 13.66, \epsilon = 1.00, p < .001 \], such that significantly shorter N1 latencies were obtained in Analog hearing aid conditions compared with either digital hearing aid. No significant main effect of gain setting existed for N1 latencies \[ F(1, 12) = .46, \epsilon = 1.00, p = .51 \], indicating that latencies for the 40-dB gain setting were not significantly shorter than latencies for the 20-dB gain setting. There was a non-significant trend for latencies to be shorter in the Analog hearing aid conditions compared with the digital hearing aid conditions across gain settings for the hearing aid type and gain interaction \[ F(2, 24) = 2.92, \epsilon = 1.00, p = .07 \].

Equivalent nominal output levels. A key question in this study was whether N1-P2 amplitudes would be the same when compared across unaided and aided conditions for the same nominal output level. More specifically, the 70 dB SPL input level for the unaided condition was compared with the 30 dB SPL input level for the 40-dB gain condition and the 50 dB SPL input level for the 20-dB gain condition, because all three combinations would be expected to yield a 70 dB SPL output in the ear canal. Grand-mean waveforms for these three combinations of unaided and aided conditions are presented in Figure 2.3. N1-P2 amplitudes are larger in the unaided compared with the aided conditions, which is confirmed by ANOVA results that revealed a significant main effect across conditions \[ F(6, 72) = 14.75, \epsilon = .89, p < .001 \]. Post-hoc analysis showed that N1 response amplitudes were significantly smaller for all hearing aid conditions (i.e., 30 dB SPL input level + 40-dB hearing aid gain and 50 dB SPL input level + 20-dB hearing aid gain) compared with the unaided condition (i.e., 70 dB SPL). Although nominal output levels should have been equal, N1-P2 amplitudes obtained with the 50 dB SPL + 20-dB
gain conditions were significantly larger than those obtained with the 30 dB SPL + 40-dB gain conditions for all hearing aid conditions except for DigitalB.

Figure 2.3. Grand-mean waveforms for 70 dB SPL equivalent nominal output for unaided and aided conditions (Analog, DigitalA and DigitalB).
Study B: Measures of Hearing Aid Processed 60- and 757-ms tonal stimuli

The purpose of Study B was to measure, in the ear canals of subjects, stimuli used for SCP testing before and after hearing aid processing to determine how this processing affected the stimuli. Of particular interest was whether there would be a difference between the gain measured with standard hearing aid test system stimuli and that measured with the stimuli used for the cortical measures. Also of interest was whether there would be a differential effect of analog and digital processing, even with all advanced features disabled.

Subjects and Recording

Five subjects participated in Study B (mean age: 23±2.1 years; 4 females). An ER7C probe-tube output (set to provide 20-dB of attenuation) was routed through a second attenuator to Channel 1 of the Neuroscan recording system. The second attenuator ensured that input was not clipped by the recording system. The recording channel was amplified, filtered (0.05-3500 Hz), and digitized (20,000 Hz), using a 204.75-ms analysis time for the short stimulus (including a 70-ms pre-stimulus baseline) and a 960-ms analysis time for the long stimulus (including a 100-ms pre-stimulus baseline). The stimulus was recorded in the ear canal for each test condition until at least 100 accepted trials were obtained. Single-trial epochs were saved for offline processing, including: baseline correction across the stimulus duration and averaging.

Data Analysis

Acoustic measures of interest were: (i) gain at 30 ms post-stimulus onset and (ii) maximum gain. Gain values were calculated for 20- and 40-dB hearing-aid gain conditions by determining the relative amplitude differences between aided and unaided stimulus waveforms from averaged recordings in the ear canal. A measurement point of 30 ms post-stimulus onset
was chosen because several studies indicate this is the most effective rise time and evokes the largest N1-P2 amplitudes, beyond which SCP amplitudes show little or no increase (Alain et al., 1997; Kodera et al., 1979; Onishi & Davis, 1986); maximum gain was calculated to determine the maximum gain produced at any time during the stimulus and whether hearing aid processing resulted in delays that would cause stimulus rise time to be longer than 30 ms.

**Statistical Analysis**

For the short-duration (60-ms) stimulus, two repeated-measures analyses of variance (ANOVA) were conducted for the gain determined for each of the 30-ms and maximum amplitude measurement points: (i) to measure the effects of the 20-dB gain setting, a two-way repeated-measures ANOVA was conducted comparing three levels of hearing aid type (Analog, DigitalA and DigitalB) and three input levels (30, 50 and 70 dB SPL), and (ii) to measure the effects of the 40-dB gain setting, a two-way repeated-measures ANOVA was conducted comparing three levels of hearing aid type (Analog, DigitalA and DigitalB) and two input levels (30 and 50 dB SPL). The same four repeated-measures ANOVAs were performed for the long duration (757-ms) stimulus.

Due to the exploratory nature of study B, main effects and interactions for all analyses were considered significant if p < .10. Huyn-Feldt epsilon (ε) correction factors for repeated-measures were applied to the degrees of freedom and reported where appropriate. Neuman-Keuls post-hoc analyses were performed for significant main effects or interactions. Post-hoc analyses were considered statistically significant if p < .10.
Results

The following section is divided into: (i) results for the short stimulus, and (ii) results for the long stimulus. Mean data for gain measured at 30-ms and maximum amplitude are provided in Table 2.1.

Acoustic waveforms for the short stimulus (60-ms duration) in both unaided and aided conditions (Analog, DigitalA and DigitalB) are presented for 30, 50, and 70 dB SPL input levels in Figures 2.4, 2.5, and 2.6, respectively. Both 20- and 40-dB gain settings are depicted where appropriate (e.g., 30 and 50 dB SPL) and 30-ms and maximum amplitude measurement points are depicted by closed and open triangles, respectively. All figures in the following section illustrate the acoustic measures for a single subject representative of the overall pattern (subject 2). It is important to note that for optimum visual representation, the scale is different across stimulus waveform figures. As a result, waveform amplitudes for 30 and 50 dB SPL input levels do not appear different in Figures 2.4 and 2.5; however, they are on different scales.
<table>
<thead>
<tr>
<th>Gain (dB)</th>
<th>Short Stimulus (60 ms)</th>
<th>Long Stimulus (757 ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>30</td>
<td>50</td>
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<tr>
<td>Input (dB SPL)</td>
<td></td>
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<tr>
<td>Analog</td>
<td></td>
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<tr>
<td>30 ms</td>
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<td>17.0±0.9</td>
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<tr>
<td>Max</td>
<td>15.7±1.1</td>
<td>16.9±1.0</td>
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<tr>
<td>DigitalA</td>
<td>10.9±4.5</td>
<td>15.8±3.1</td>
</tr>
<tr>
<td>Max</td>
<td>15.4±3.2</td>
<td>17.1±2.7</td>
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<tr>
<td>Digital B</td>
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<td>0.9±2.9</td>
</tr>
<tr>
<td>Max</td>
<td>4.5±3.8</td>
<td>7.6±2.3</td>
</tr>
</tbody>
</table>
Figure 2.4. The short stimulus presented at 30 dB SPL as measured in the ear canal of one subject who is representative of the overall pattern. The unaided condition is centered at the top of the diagram, and aided conditions are marked according to hearing aid type and gain setting. On all stimulus waveforms, the 30-ms and maximum amplitude measurement locations are marked by closed and open triangles, respectively.
Figure 2.5. The short stimulus presented at 50 dB SPL as measured in the ear canal of one subject who is representative of the overall pattern. The unaided condition is centered at the top of the diagram, and aided conditions are marked according to the hearing aid type and gain setting. On all stimulus waveforms, the 30-ms and maximum amplitude measurement locations are marked by closed and open triangles, respectively.
Figure 2.6. The short stimulus presented at 70 dB SPL as measured in the ear canal of one subject who is representative of the overall pattern. The unaided condition is centered at the top of the diagram, and aided conditions are marked according to the hearing aid type and gain setting. On all stimulus waveforms, the 30-ms and maximum amplitude measurement locations are marked by closed and open triangles, respectively.
Comparison among hearing aids: 20-dB gain condition for the short stimulus at 30 ms post-stimulus onset. Mean gain values presented in Table 2.1 indicate that at the 30-ms measurement point for the short stimulus, all hearing aids provided less than 20-dB gain. The DigitalB hearing aid provided less gain than the Analog or DigitalA hearing aids. Results from the ANOVA showed significant main effects of hearing aid type \([F(2, 8) = 99.97, \varepsilon = .89, p < .001]\) and input level \([F(2, 8) = 17.80, \varepsilon = 1.00, p < .001]\), as well as a significant interaction between the two \([F(4, 16) = 3.49, \varepsilon = .68, p = .06]\). Neuman-Keuls post hoc analysis revealed that the Analog hearing aid provided equal gain across all input levels and although all three hearing aids were set to provide linear amplification, both digital hearing aids provided less gain for the 30 compared with the 50 dB SPL input and equivalent gain for the 50 and 70 dB SPL inputs. Post-hoc analysis confirmed that the DigitalB hearing aid provided significantly less gain than the DigitalA and Analog hearing aids across all three input levels and the Analog hearing aid provided significantly more gain than the DigitalA aid for the 30 dB SPL input. There was no significant difference between gain provided by Analog and DigitalA aids for 50 and 70 dB SPL input levels.

Comparison among hearing aids: 20-dB gain setting for the short stimulus measured at maximum amplitude. Mean gain values presented in Table 2.1 show that, when measured at the maximum amplitude measurement point, once again, all of the hearing aids provided less than 20-dB gain. The DigitalB aid, once again, provided less gain than Analog and DigitalA hearing aids. The ANOVA revealed a significant main effect of hearing aid type \([F(2, 8) = 60.81, \varepsilon = 1.00, p < .001]\) and post-hoc analysis confirmed that the Analog hearing aid did not produce significantly more gain than the DigitalA hearing aid; however, the DigitalB hearing aid provided significantly less gain than both DigitalA and Analog hearing aids. The main effect of
input level was also statistically significant \([F(2, 8) = 6.81, \varepsilon = .54, p = .05]\), such that less gain was provided across hearing aids for the 30 dB SPL input level compared with 50 and 70 dB SPL input levels. The interaction between hearing aid type and input level was not statistically significant \([F(4, 16) = 2.27, \varepsilon = .45, p = .17]\). The difference in measured gain between DigitalB and both Analog and DigitalA hearing aids is also evident in Figures 2.4, 2.5 and 2.6 for the 30, 50, and 70 dB SPL input levels, respectively; however, the difference between gain provided by the Analog and DigitalA hearing aids in this individual’s data are much larger than mean values listed in Table 2.1. These results indicate that DigitalA produced nearly as much gain as the Analog hearing aid at some point during the 60-ms stimulus used for SCP testing in Study A. However, none of the three hearing provided 20-dB gain for the short stimulus and less gain was provided for the 30 dB SPL input. These findings are reflected in the results of Study A for SCP measures of the 20-dB gain setting, where amplitudes were not significantly larger in aided compared with unaided conditions.

*Comparison among hearing aids: 40-dB gain setting for the short stimulus measured 30 ms post-stimulus onset.* Mean gain values indicate that, when measured 30 ms post-stimulus onset, all of the hearing aids provided less than 40-dB gain. The DigitalB hearing aid provided less gain than both Analog and DigitalA hearing aids. The ANOVA revealed a significant main effect of hearing aid type \([F(2, 8) = 39.10, \varepsilon = .78, p < .001]\) as well as a significant main effect of input level \([F(1, 4) = 8.26, \varepsilon = 1.0, p = .05]\). There was also a significant interaction between hearing aid type and input level \([F(2, 8) = 5.17, \varepsilon = .80, p = .05]\), such that post-hoc analysis revealed the DigitalB hearing aid provided significantly less gain than the DigitalA hearing aid for the 30 dB SPL input level, and the DigitalA hearing aid, in turn, provided significantly less gain than the Analog hearing aid. For the 50 dB SPL input, however, the DigitalB hearing aid
provided significantly less gain than both other hearing aids but there was no longer any significant difference between gain provided by the Analog and DigitalA hearing aids. Additionally, DigitalB provided the same amount of gain for both 30 and 50 dB SPL input levels. These findings are similar to results for the 20-dB gain setting indicating that the DigitalB hearing aid is providing significantly less gain for these brief stimuli than either the DigitalA or Analog hearing aids.

Comparison among hearing aids: 40-dB gain setting for the short stimulus measured at maximum amplitude. Mean gain values follow much the same pattern at maximum amplitude for the short stimulus as it did at 30 ms. Once again, all of the hearing aids provided less than 40-dB gain and the DigitalB provided less gain than both Analog and DigitalA hearing aids. Results from the ANOVA revealed a significant main effect of hearing aid type, \( F(2, 8) = 37.58, \varepsilon = .67, p < .001 \), such that the DigitalB hearing aid provided significantly less gain than either the Analog or DigitalA hearing aids and there was no significant difference between the gain provided by Analog and DigitalA. There was also a significant main effect of input level \( F(1, 4) = 5.1, \varepsilon = 1.0, p = .09 \) and post-hoc analysis showed that the 50 dB SPL input level resulted in significantly higher gain compared with the 30 dB SPL input level. There was no significant interaction between hearing aid type and input level \( F(2, 8) = .01, \varepsilon = .70, p = .97 \).

Acoustic waveforms for the long stimulus (757-ms duration) in both unaided and aided conditions (Analog, DigitalA and DigitalB) are presented for 30, 50, and 70 dB SPL input levels in Figures 2.7, 2.8, and 2.9, respectively. Both 20- and 40-dB gain settings are depicted where appropriate (e.g., 30 and 50 dB SPL) and 30-ms and maximum amplitude measurement points
are depicted by closed and open triangles, respectively. Once again, all figures in the following section illustrate the acoustic measures for a single representative subject (subject 2).

Comparison among hearing aids: 20-dB gain setting for the long stimulus measured 30-ms post-stimulus onset. Mean gain values in Table 2.1 indicate that, when gain for the long stimulus was measured at 30 ms, all of the hearing aids provided less than 20-dB gain. Once again the DigitalB hearing aid provided less gain than either Analog or DigitalA hearing aid. The ANOVA revealed significant main effects for hearing aid type \([F(2, 8) = 47.96, \varepsilon = .98, p < .001]\) and input level \([F(2, 8) = 40.92, \varepsilon = .64, p < .001]\). There was also a significant interaction between hearing aid type and input level \([F(4, 16) = 3.91, \varepsilon = .84, p = .03]\), such that no significant difference was observed between gain provided by Analog and DigitalA hearing aids for the 50 and 70 dB SPL inputs; however, for the 30 dB SPL input, the Analog hearing aid provided significantly more gain than the DigitalA aid. Additionally, both DigitalA and Analog aids provided significantly more gain than the DigitalB aid across all input levels. Generally, this pattern of results is seen in Figures 2.7, 2.8, and 2.9; however, the difference between Analog and DigitalA hearing aid gain in the individual’s waveforms is larger than is indicated by the mean data presented in Table 2.1. Post-hoc analysis also revealed significant processing differences across the three hearing aids, with the Analog hearing aid producing equivalent gain across all three input levels, whereas the DigitalA hearing aid provided less gain for the 30 compared with the 50 dB SPL input and equivalent gain for 50 and 70 dB SPL inputs. Finally, the DigitalB hearing aid provided less gain for the 30 compared with the 50 dB SPL input and less gain for the 50 compared with the 70 dB SPL input. These results suggest that DigitalB hearing aid processes these relatively brief tonal stimuli differently from both other hearing aids, and both digital hearing aids, in turn, process stimuli differently from the Analog hearing aid.
Figure 2.7. The long stimulus presented at 30 dB SPL as measured in the ear canal of one subject who is representative of the overall pattern. The unaided condition is centered at the top of the diagram, and aided conditions are marked according to the hearing aid type and gain setting. On all stimulus waveforms, the 30-ms and maximum amplitude measurement locations are marked by closed and open triangles, respectively.
Figure 2.8. The long stimulus presented at 50 dB SPL as measured in the ear canal of one subject who is representative of the overall pattern. The unaided condition is centered at the top of the diagram, and aided conditions are marked according to the hearing aid type and gain setting. On all stimulus waveforms, the 30-ms and maximum amplitude measurement locations are marked by closed and open triangles, respectively.
Figure 2.9. The long stimulus presented at 70 dB SPL as measured in the ear canal of one subject who is representative of the overall pattern. The unaided condition is centered at the top of the diagram and aided conditions are marked according to the hearing aid type. On all stimulus waveforms, the 30-ms and maximum amplitude measurement locations are marked by closed and open triangles, respectively.
Comparison among hearing aids: 20-dB gain setting for the long stimulus measured at maximum amplitude. At maximum amplitude for the long stimulus, the mean gain values indicate that all three hearing aids nearly produced 20-dB gain, with the Analog and DigitalA hearing aids providing more gain than the DigitalB hearing aid. The ANOVA showed a significant main effect of input level [F(2, 8) = 10.77, ε = .82, p = .001] but not for hearing aid type [F(2, 8) = .45, ε = 1.0, p = .65]; however, there was a significant interaction between hearing aid type and input level [F(4, 16) = 3.56, ε = .42, p = .09]. Post-hoc analysis confirmed that the DigitalB aid provided less gain than the DigitalA aid for both 30 and 50 dB SPL inputs but only provided less gain than Analog hearing aid for the 30 dB SPL input. On the other hand, there was no significant difference between Analog and DigitalA aids for the 50 dB SPL input. At 70 dB SPL, there was no significant difference among gain provided by any of the three hearing aids. This pattern is not nearly as evident in the individual’s stimulus waveforms presented in Figures 2.7 through 2.9; however, it is clear that the DigitalB hearing aid produced significantly more gain at maximum amplitude than at the 30-ms measurement point. Post-hoc analysis also showed that, once again, the Analog hearing aid provided equivalent gain across all three input levels, the DigitalA aid produced less gain for the 30 compared with 50 dB SPL input and equivalent gain for 50 and 70 dB SPL inputs, and the DigitalB aid produced less gain for the 30 compared with the 50 dB SPL input and less gain for the 50 compared with the 70 dB SPL input. There was no significant main effect of hearing aid type for this comparison [F(2, 8) = .45, ε = 1.00, p = .65]. These results indicate that although none of the hearing aids reached 20-dB gain, all three came closest to producing 20-dB gain for this longer stimulus and later measurement point (compared to shorter stimulus at 30 ms or maximum amplitude or to longer stimulus at 30 ms). This suggests that hearing aids, particularly those with digital processing, require additional time to reach maximum gain. Interestingly, this occurred at
different times in the stimulus waveform for each hearing aid, as is evident in Figures 2.7 through 2.9. Although no statistical analysis was conducted on hearing aid processing delay, it is apparent that maximum gain was reached much later by the DigitalB hearing aid compared with both Analog and DigitalA hearing aids.

Comparison among hearing aids: 40-dB gain setting for the long stimulus measured 30-ms post-stimulus onset. Mean gain values presented in Table 2.1 indicate that similar to the 20-dB gain setting at 30 ms, all of the hearing aids provided less than 40-dB gain. In particular, the DigitalB provided less gain than either the Analog or DigitalA hearing aids. The ANOVA revealed a significant main effect of hearing aid type \( [F(2, 8) = 55.0, \varepsilon = .84, p < .001] \) and post-hoc analysis showed that once again the DigitalB hearing aid provided significantly less gain than both other hearing aids and there was no significant difference in gain provided by Analog and DigitalA. The main effect of input level was also statistically significant \( [F(1, 4) = 11.6, \varepsilon = 1.0, p = .03] \), such that less gain was provided overall for the 30 dB SPL input compared with the 50 dB SPL input. These findings are consistent with results for the 20-dB gain setting for the same stimulus and measurement point.

Comparison among hearing aids: 40-dB gain setting for the long stimulus measured at maximum amplitude. Mean gain data indicate that, at maximum amplitude for the long stimulus, all three hearing aids nearly reached 40-dB gain for all input levels. The ANOVA revealed no significant main effects for hearing aid type \( [F(2, 8) = 1.34, \varepsilon = 1.0, p = .32] \) or input level \( [F(1, 4) = 1.42, \varepsilon = 1.0, p = .30] \). Although the interaction between hearing aid type and input level was not statistically significant, \( [F(2, 8) = 2.99, \varepsilon = 1.0, p = .10] \), there was a trend for all three hearing aids to provide equivalent gain for the 50 dB SPL input and for DigitalA to provide more
gain than either of the other hearing aids for the 30 dB SPL input. Similar to the 20-dB gain setting, these results demonstrate that the DigitalB hearing aid nearly provided 40-dB gain if given enough time to reach maximum amplitude; however, as is evident in Figures 2.7 through 2.9, this occurs much later compared with both other hearing aids.
DISCUSSION

There has been much interest in using the slow cortical potential as an objective hearing aid validation measure (Dillon, 2005; Billings et al., 2007; Golding et al., 2007; Gravel et al., 1989; Korczak et al., 2005; Purdy et al., 2005; Rapin & Graziani, 1967; Tremblay et al., 2006); however, there is conflicting evidence in the literature regarding the accuracy of the SCP for this purpose. Although researchers agree that aided SCPs can be measured (Billings et al., 2007; Golding et al., 2007; Gravel et al., 1989; Korczak et al., 2005; Purdy et al., 2005; Rapin & Graziani, 1967; Tremblay et al., 2006) and that different cortical responses for different speech stimuli are maintained in aided conditions (Golding et al., 2007; Korczak et al., 2005; Purdy et al., 2005; Tremblay et al., 2006), this is only the third study assessing the accuracy of measuring changes in hearing aid settings using SCPs. The previous two studies have reported no significant differences between unaided N1-P2 amplitudes and N1-P2 amplitudes measured while subjects wore hearing aids providing 20-dB of gain (Billings et al., 2007; Tremblay et al., 2006). Importantly, the present research is the only study to assess changes to the acoustic waveforms of SCP stimuli resulting from hearing aid processing.

Comparing Results of Studies A and B

It is important to reiterate that verification of REIG measures using conventional hearing aid test procedures confirmed all three hearing aids provided 20 and 40 dB of gain at mid- and high-level inputs for all subjects. This is particularly noteworthy given (i) the general lack of difference between N1-P2 amplitudes in aided and unaided conditions in Study A, particularly for the 20-dB gain setting, and (ii) none of the hearing aids achieved these gain levels when measured using the same stimulus (Study B). This suggests that SCP amplitudes and latencies are consistent with the characteristics of stimuli following hearing aid processing. In addition,
there were similar relationships among hearing aids across Studies A and B. For instance, N1-P2 amplitude results from Study A for the 20-dB gain setting indicated that response amplitudes were much smaller in DigitalB hearing aid conditions compared with both other hearing aids. In Study B, the DigitalB hearing aid provided significantly less gain to the short stimulus than both DigitalA and Analog hearing aids across all input levels when measured at 30 ms. Recall that this is the same stimulus used in Study A, and the 30-ms measurement point is relevant due to the effect of rise-time on SCP peak amplitudes (Alain et al., 1997; Kodera et al., 1979, Onishi & Davis, 1968). Results from these studies are consistent with the view that the first 30 ms of stimulus onset largely determines N1 presence and amplitude. It appears that the amount of gain a given hearing aid has provided by 30 ms post-stimulus onset will determine the amplitude of N1 relative to unaided response amplitudes, because: (i) the cortical findings reported in Study A and the findings reported by Billings et al. (2007) were similar in that gain could not be accurately reflected by the SCPs, and (ii) the pattern of acoustic measures for the long and short stimuli were similar at the 30-ms measurement point, despite the different rise-times of the two stimuli (i.e., 20-ms and 7.57-ms rise-times for the short and long stimuli, respectively). The combined results of Studies A and B indicate that digital hearing aids, in particular, require additional time to reach maximum gain for short duration SCP stimuli and may in fact provide negative gain when measured 30-ms post-stimulus onset. As a result, N1 amplitudes in digital hearing aid conditions are smaller in some conditions than unaided N1 amplitudes.

Additional evidence that SCP results are reasonably consistent with the stimulus characteristics after hearing aid processing was provided by N1 latency measures. In particular, N1 latencies were longer in conditions involving digital hearing aids compared with Analog and unaided conditions, while there were no significant differences between N1 latencies for unaided
and Analog aided conditions. Similarly, electroacoustic measures of delay conducted on the Fonix 7000 system indicated that both digital hearing aids had longer delays (i.e., 6.8 ms and 2.3 ms for DigitalA and DigitalB, respectively) compared with the Analog hearing aid (i.e., 0.4 ms). Longer delays for digital compared with analog hearing aids are commonly reported in the acoustics literature (Agnew & Thornton, 2000; Chung, 2004a; Dillon, Keidser, O’Brien, & Silberstein, 2003; Henrickson & Frye, 2003; McGrath & Summerfield, 1985; Stone & Moore, 1999, 2002, 2003). Interestingly, prior studies involving SCP measures of hearing aids have reported shorter latencies in aided conditions, even when testing involved digital hearing aids (Billings et al., 2007; Korczak et al., 2005; Tremblay et al., 2006).

Stimulus Waveform Shape

Although there was no statistical analysis involving the shape of stimulus waveforms across conditions, the waveforms obtained for Study B show visually that shape varies across hearing aids for both short- and long-duration stimuli, particularly the onset. For instance, unaided and Analog hearing aid waveform shapes do not appear to be different. In contrast, DigitalA and DigitalB waveform shapes appear different from unaided waveforms. The initial portion of the DigitalB hearing aid waveform is rounded, such that it appears that maximum gain is reached more gradually in this hearing aid. This is reflected by the gain measured at 30 ms and at maximum amplitude. In particular, for the 30 dB SPL presentation level and short stimulus, the DigitalA waveform was shaped similar to DigitalB. This was less obvious for all other conditions and, interestingly, it was a consistent finding that less gain was provided by digital hearing aids for the 30 dB SPL input level. This may or may not have been due to expansion processing, which functions in a complementary way to compression by providing less gain for low level inputs than for high level inputs. The implication of these findings for
SCP research is that the same hearing aid may alter stimuli in different ways for different input levels despite being set to provide linear gain across the input range and disabling all advanced processing features.

The current studies have shown that SCP measures were consistent with the acoustic measures of hearing aid processed stimuli, such that N1 amplitudes were smaller and latencies longer if the stimulus was altered in such a way that its amplitude was small when measured 30-ms post-stimulus onset.
CONCLUSIONS

In light of findings from the present study, it is likely that prior studies using short-duration speech or tonal stimuli (Billings et al., 2006; Golding et al., 2007; Gravel et al., 1989; Korczak et al., 2005; Pearce et al., 2007; Purdy et al., 2005; Rapin & Graziani, 1967; Tremblay et al., 2006) were measuring SCPs to stimuli altered by the hearing aid in a way that was not quantified. For instance, different speech stimuli may not result in distinct neural response patterns if the hearing aid processed stimuli are altered in such a way that they are acoustically very similar. Likewise, the same stimulus may be altered in different ways by the same hearing aid, as was the case in the current studies.

Prior studies on hearing aid processed click and brief-tone stimuli (typically used for ABR testing), reported considerable variability among hearing aids in terms of gain provided to onset and steady-state portions of transient stimuli (Brown, Klein & Snydee, 1999; Gorga et al., 1987), thus, these stimuli were determined to be too short for measures of hearing aid processing. The longer duration stimuli used for SCP testing were thought to be long enough to overcome this problem; however, findings from the current studies indicate that a tonal stimulus with parameters appropriately set to elicit large unaided N1 amplitudes is still too brief to measure hearing aid gain, particularly those with digital processing.

The inability to reliably measure the effects of hearing aid gain using SCP recordings (Study A), and the less-than-expected measureable gain resulting from hearing aid processing in Study B, indicates that SCP stimuli presented in this manner do not provide appropriate measures of hearing aid gain. Future research might involve additional hearing aid measures to determine the source of alteration to rise time; however, the current research indicates that the
problem lies in the type of stimuli required for SCP testing, rather than a problem associated with hearing aid processing. As a result, there remains a need for a reliable objective hearing aid validation measure. Future research might explore different/more appropriate SCP stimuli for hearing aid measures.
REFERENCES


Study A

Appendix A: Amplitude and latency measures by condition and subject
Appendix A: Amplitude and Latency measures by condition and subject

Table A1. Amplitude and latency measures across unaided conditions for each subject plus mean and standard deviation values.

<table>
<thead>
<tr>
<th>Input Level (dB SPL)</th>
<th>N1-P2 Amplitude (µV)</th>
<th>N1 Latency (ms)</th>
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Table A2. Amplitude and latency measures across aided conditions for each subject plus mean and standard deviation values for the Analog hearing aid.

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Analog
Table A3. Amplitude and latency measures across aided conditions for each subject plus mean and standard deviation values for the Digital A hearing aid.

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Table A4. Amplitude and latency measures across aided conditions for each subject plus mean and standard deviation values for the Digital B hearing aid.

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<th>N1 Latency (ms)</th>
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<td>30  50  70</td>
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Study B

Appendix B: Gain measures by condition and subject
### Table B1. Real-ear insertion gain (REIG) measures for 20- and 40-dB gain settings across hearing aids.

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<td>19.80</td>
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<td>20.20</td>
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Table B2. Gain calculated for the short stimulus at 30-ms and maximum amplitude measurement points by condition and subject.

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<th>Measurement Point</th>
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<th>Maximum Amplitude</th>
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<th>50</th>
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<td>Gain Setting (dB)</td>
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<td>50</td>
<td>30</td>
<td>50</td>
<td></td>
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<tr>
<td>Input Level (dB SPL)</td>
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<td>50</td>
<td>70</td>
<td>30</td>
<td>50</td>
</tr>
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<td><strong>Subject</strong></td>
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<tr>
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<td>16.64</td>
<td>17.44</td>
<td>34.25</td>
<td>36.83</td>
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<td>18.56</td>
<td>18.92</td>
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<td>37.13</td>
</tr>
<tr>
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<td>14.70</td>
<td>16.80</td>
<td>17.11</td>
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Table B3. Gain calculated for the long stimulus at 30-ms and maximum amplitude measurement points by condition and subject.

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Appendix C: Behavioural Research Ethics Board (BREB) approval
The University of British Columbia
Office of Research Services
Behavioural Research Ethics Board
Suite 102, 6560 Agronomy Road, Vancouver, B.C. V6T 1Z3

CERTIFICATE OF APPROVAL - MINIMAL RISK

PRINCIPAL INVESTIGATOR: David R. Stabel

INSTITUTION / DEPARTMENT: UBC Medicine, Faculty of Audiology & Speech Sciences

UBC BREB NUMBER: H08-00546

INSTITUTION(S) WHERE RESEARCH WILL BE CARRIED OUT:

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Other locations where the research will be conducted:
The project will be conducted in the Human Auditory Physiology Laboratory in the Instructional Resources Centre on the UBC campus.

CO-INVESTIGATOR(S):

| Lorenae Jensen |
| Susan Lindsey Marynowich |

SPONSORING AGENCIES:

N/A

PROJECT TITLE:

Cortical event-related potential measures of amplification

CERTIFICATE EXPIRY DATE: May 9, 2009

DOCUMENTS INCLUDED IN THIS APPROVAL: Consent Forms:

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The application for ethical review and the document(s) listed above have been reviewed and the procedures were found to be acceptable on ethical grounds for research involving human subjects.

Approval is issued on behalf of the Behavioural Research Ethics Board and signed electronically by one of the following:

Dr. M. Judith Lynam, Chair
Dr. Ken Craig, Chair
Dr. Jim Rapport, Associate Chair
Dr. Laurie Ford, Associate Chair
Dr. Daniel Sathian, Associate Chair
Dr. Anita Ho, Associate Chair