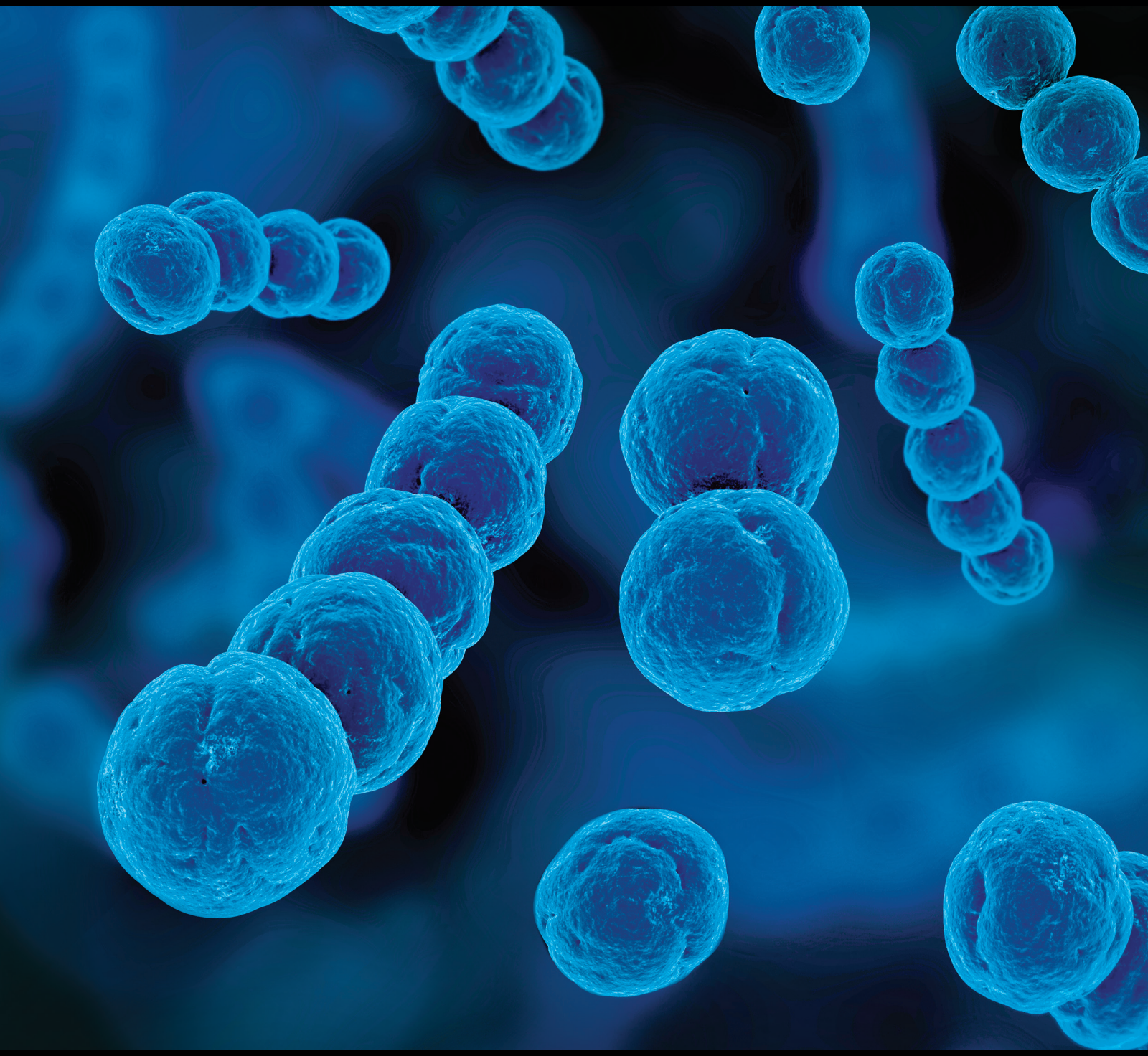
International Journal of Otolaryngology

Hearing Aids and the Brain

Guest Editors: K. L. Tremblay, S. Scollie, H. B. Abrams, J. R. Sullivan, and C. M. McMahon



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Editorial Board

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Editorial

Hearing Aids and the Brain

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At the heart of most rehabilitation programs for people with hearing loss is the use of amplification. The purpose of hearing aid amplification is to improve a person's access to sound. Depending on the degree and configuration of the individual's hearing loss, the hearing aid is tasked with increasing sound levels at different frequency regions to ensure that incoming speech frequencies are reaching the ear at sufficient levels to compensate for the individual's hearing loss. However, a perceptual event is dependent not only on the audibility of the signal at the level of the ear, but also on how that sound is biologically coded, integrated, and used. As described by Tremblay and Miller in this special issue, this complex ear-brain neural network system starts with sound leaving the hearing aid. At this stage the acoustics of the amplified signal has been altered by the hearing aid. It is this modified signal that is being encoded at subsequent stages of processing: the ear, brainstem, midbrain, and the cortex. The integrity of the signal, and the biological codes, are therefore assumed to contribute to the resultant perceptual event and it is for this reason that the brain can be considered an essential component to rehabilitation. Yet, little is known about how the brain processes amplified sound or how it contributes to perception and the successful use of hearing aid amplification (see Figure 1).

The intent of this IJO special edition is to integrate neuroscience and clinical practice to advance hearing health care. Such lines of inquiry can spawn new insight into stimulation-related brain plasticity that might, in turn, help explain why some individuals (but not others) report increased speech understanding while wearing their devices. From a clinical perspective, there is interest in determining if measures of brain activity might be of use to the clinician, during hearing aid selection and fitting, as well as to the engineers who are designing the instruments. However, to move forward, there are unresolved issues that can appear conflicting to the clinician and/or scientist who wish to embark in new research directions or clinical services. For this reason, a call for papers on the topic of Hearing Aids and the Brain was made in an effort to define converging evidence. What resulted is a collection of papers from different laboratories that identify caveats and concerns, as well as potential opportunities.

The collection of papers addresses two main questions: (1) Is it possible to use brain measures to determine a person's neural detection of amplified sound? (2) Can brain measures provide information about how the brain is making use of this amplified sound? Before we summarize the answers to the questions, it's important to review the framework for including objective measures to examine the neural representation of amplified sound.

When a person is fitted with a hearing aid, there are two categories of test procedures involved in the process: (i) behavioral measures, which require active participation by the patient (e.g., pure-tone threshold estimation, speech testing, and self-report questionnaires), and (ii) objective measures, where subjective responses are not required (e.g., probe microphone electroacoustics and unaided electrophysiology). Here we expand the use of electrophysiology to determine if brain measures can be used as an objective measure to estimate aided thresholds and/or quantify hearing aid transduced signals in the brain for the purpose

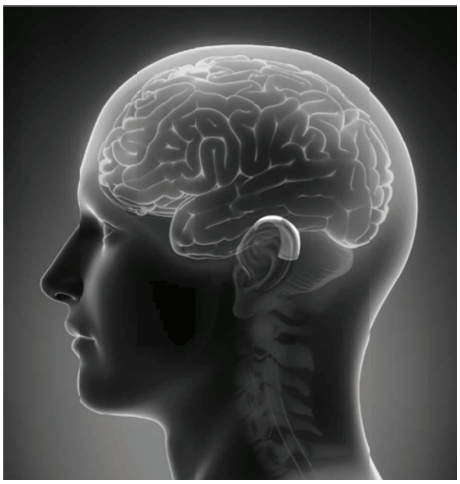

Figure 1

of guiding device fitting and/or to assess suprathreshold representation of amplified auditory signals in the brain to estimate perceptual performance and/or the related cognitive resources involved. This expanded use of objective measures is relevant to patients of all ages but is particularly germane to the pediatric population where the use of behavioral tools is limited. As described by L. M. Jenstad et al. in this issue, behavioral threshold information is not usually available before age 6 months (and often later), speech testing is unavailable, and subjective questionnaires are limited to caregiver observation of behaviors. Thus, there is greater reliance on objective procedures to measure the effects of amplification beyond the tympanic membrane in infants and young children. One such measure is the use of auditory evoked potentials (AEPs).

The use of AEPs to aid in clinical assessment is not new. Click-evoked auditory brainstem responses were tried long ago to estimate unaided and aided thresholds in infants and young children. However they proved to be unsuccessful because the short duration signal (click, tone-pip) interacted with the hearing aid circuitry in a way that introduced ringing and other artifacts [1]. In this special issue, S. Anderson and N. Kraus reintroduce the concept of using complex speech evoked ABRs (also called the frequency following response (FFR)) and provide a case study to show that it is possible to record FFRs while a person is wearing a hearing aid. But FFR and speech evoked ABR research is still in its infancy. It will become necessary to define the interactions that take place between the instrument and brainstem responses, especially in people with hearing loss and who wear hearing aid devices to determine if some of the obstacles encountered when recording cortical evoked potentials (CAEPs) also apply to the FFR.

There is much literature exploring the role of CAEPs in assessing people with hearing loss, but the inclusion of people with hearing loss who wear hearing aids is still quite sparse. In this special issue, investigators from different laboratories describe some of caveats and concerns when measuring evoked CAEPs in combination with hearing aid amplification. L. M. Jenstad et al. showed that CAEPs (P1-N1-P2) do not reliably reflect hearing aid gain, even when different types of hearing aids (analog and digital) and their parameters (e.g., gain, frequency response) are manipulated. As shown in Figure 2, N1 latency (and N1-P2 amplitudes) is not significantly affected even when the gain of the input signal is increased by 20 dB or 40 dB. This is true even when different hearing aid processing types are involved. These results reinforce those previously published by Billings et al. [2, 3] over the past few years, as well as those presented in this special issue. Billings et al. demonstrate continuing evidence that the signal-to-noise ratio (SNR) produced by the hearing aid and entering the brain influences aided cortical potentials (P1-N1-P2) in a way that obscures the biological representation of hearing aid gain. These results, and those described earlier by Billings et al., provide converging evidence that time-locked cortical activity does not always reflect the expected (and electroacoustically verified) gain of hearing aids, and that different hearing aids with similar electroacoustic parameters (i.e., gain) may result in substantially different evoked results. The use of CAEPs as an accurate proxy of hearing aid gain therefore remains problematic. At issue, in addition to simple gain, is the influence of hearing aid signal processing features such as channel-specific compression time constants, noise reduction algorithms, and adaptive directionality on cortical and subcortical evoked potentials.

L. M. Jenstad et al. describe how these issues impact the clinician. It would be possible to appropriately fit a hearing aid (for degree and configuration of the hearing loss) in a client but show no evoked brain response or no improvement from the unaided cortical response, even though, behaviorally, the client shows clear perceptual improvements. In another manuscript, D. Glista et al. also show instances when cortical P1-N1-P2 responses are absent in normal hearing children. Even when cortical evoked potentials are present, their peak morphology may or may not reflect defining characteristics of the brain response. Instead they could reflect aspects of signal processing influenced by the hearing aid.

We know very little about this last point, and the research published in this issue attests to this. L. M. Jenstad et al. published examples where differences in the type/brand of hearing aid resulted in substantial differences in cortical activity, even though standard electroacoustic measures indicated no major differences among the hearing aids. In contrast, D. Glista et al. presented a few pilot cases where altering frequency compression hearing aid parameters not only improved audibility of a 4 kHz tone burst but also improved detection of cortical evoked responses. So, it appears that the presence of P1-N1-P2 in response to an amplified signal indicates the neural detection of a sound at the level of the cortex, but the morphology of the response (latency and amplitude) may say more about the signal processed acoustics of the amplified sound than the status of the neural mechanisms generating the response. A dilemma ensues when the CAEP is absent. The absence of a brainstem or cortical response, or the absence of change in the evoked brain activity, may result from a number of variables not yet fully understood and therefore complicating the use of AEPs

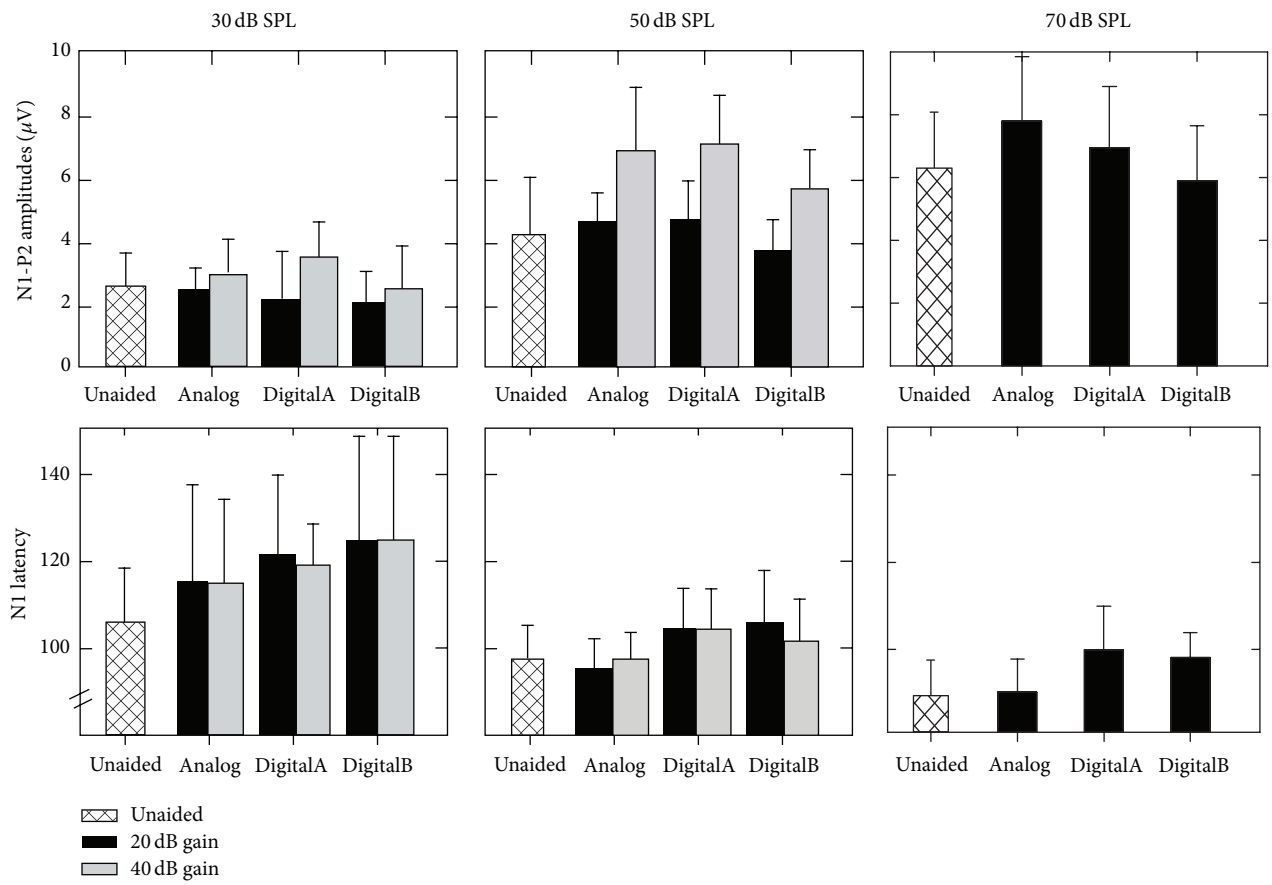

Figure 2: Mean and SD amplitude and latency data for unaided and aided conditions (Analog, DigitalA, and DigitalB) with two gain settings (20 and 40 dB) at three input levels (30, 50, and 70 dB SPL). Reprinted from L. M. Jenstad et al.

as a sensitive and specific metric of hearing loss and abnormal brain processing. To summarize then, when examining the effect of hearing aid amplification on brain maturation or plasticity, the presence or absence of change over time might simply reflect changes in signal alterations introduced by the hearing aid within a single recording or between sessions. In all cases, changes to the hearing aid prescription can be presumed to contribute to changes in the evoked neural responses. The same can likely be said for cochlear implants too. For any scientist or clinician to overlook the contribution of the hearing aid-transduced signal to stimulation-related patterns in the brain is to overlook a core essential variable.

While the previously expressed concerns highlight the cautionary aspects of combining brain measures and hearing aid amplification, the contribution of device-related changes to evoked brain activity could also be viewed as an opportunity to study the utility of specific hearing aid algorithms in a way that could be used as an outcome measure. This opportunity assumes two things: (1) it is possible to reliably characterize signal processing schemes specific to each device in a way that contributes information to the hearing aid fitting process in an efficient and informative way, and (2) that clinicians (who are reluctant to use existing technology such as probe microphone technology) would be accepting of adding an additional measure to their protocol.

However, as this direction of ear-brain testing continues, future research is needed to explore appropriate test stimuli and presentation paradigms when hearing aids are included in research. As described by D. Glista et al, we need a better understanding of the testing conditions and stimuli that yield the most valid measures electrophysiologically in relationship to behavioral measures. Traditional presentations of speech stimuli, with tones or speech syllables being presented in isolation interleaved with silent periods so the necessary average brain responses which can be obtained, might not be a comparable auditory experience to the running speech that is usually delivered to the brain by the hearing aid. For example, in this issue, V. Easwar et al. evaluated the output levels of 10 different hearing aids for phonemes in isolation and in running speech to determine the effects of processing strategies on the output level of each. Their results show remarkable differences in sound level and hearing aid activation, depending on the method of stimulus presentation. This means that different conclusions could be drawn for the same person, depending on the way in which the stimuli interact with the hearing aid. A more optimistic spin on this finding could be that it might be possible to use cortical activity to assess the effects of hearing aid fine-tuning (e.g., changes to hearing aid gain and frequency shaping), or other aspects of hearing aid signal processing, such as frequency

lowering, as illustrated by some participants reported by D. Glista et al. However, validation of aided findings in larger groups of participants with hearing loss including infants and in young children would be needed to establish valid clinical interpretations both for present and absent aided CAEPs. Particularly for assessing the outcome of an infant's hearing aid fitting, appropriate interpretation of an absent CAEP is critical, so that appropriate communication of results to caregivers can take place.

Another important point is that there might be optimal montages for recording brainstem and cortical activity that have not yet been realized. Even though most published data to date show that it is possible to record brainstem and cortical responses using only a small number of electrodes, this need not mean that the routine recording montage typically used in clinical settings is the most sensitive and specific recording approach for research purposes. A recent example of what information can be lost when using a standard vertex recording of the P1-N1-P2 complex in adults can be found in Tremblay et al. [5]. They showed how a significant amount of information about sound processing and relevance to auditory rehabilitation is lost when electrodes over the temporal lobes are not analyzed. This point is especially important when testing the pediatric population. When testing infants and children, recording montages will be influenced by the changing distribution of evoked activity generated by the maturing brain.

Another potential area of research is to examine the interaction between onset and change responses evoked by the auditory cortex. This information provides an objective quantification of the relationship between the onset of the processed signal versus changes within a between processed sounds. The P1-N1-P2 cortical response is appropriate for this purpose because it is sensitive to the onset of sound as well as to acoustic changes within an ongoing sound [4, 5]. When recorded in response to acoustic changes, the same P1-N1-P2 complex is described as an acoustic change complex (ACC). These differences in the CAEPs may have implications for hearing aid-brain interactions. To date little is known about the relationship between hearing aid signal processing in response to the onset of sound and the dynamic changes in hearing aid circuitry that are triggered by the dynamic nature of the speech signal. Rapid and slow dynamic changes in the incoming acoustic signal activate the hearing aid circuitry in different ways that likely influence evoked brain activity measured from the scalp. For this reason, the literature base that is developing around the P1-N1-P2 response may not apply to the acoustic influences contributing to the ACC. Taking this point even further, EEG responses that are not time locked and outside the traditional practice of audiology may prove to be of value to the selection, fitting, and design of hearing aids. The P1-N1-P2 and ACC are being studied because they are already familiar to the audiology community and could be implemented into clinical practice. But there are many ways in which EEG activity can be measured that are not discussed here.

For example, brain measures could guide manufacturer designs. Knowing how the auditory system works and how it responds to gain, noise reduction, and/or compression circuitry could influence future generations of biologically motivated changes in hearing aid design. Biological codes have been harnessed to drive the motion of an artificial limb/prosthesis, it might therefore be possible one day to design a hearing prosthesis that includes neuromachine interface systems driven by a person's listening effort or attention. At this stage, however, finding ways to quantify a person's listening effort is still being defined. For example, demonstrating the benefits of digital noise reduction algorithms, as implemented in hearing aids, on improved speech recognition in noise has been elusive. Anecdotal reports suggest that such algorithms may provide nonspeech perception benefits such as improved ease of listening. Therefore physiologic measures of stress and effort are being explored to determine if they could provide an objective measure of nonspeech benefits. One such example is pupillometry. In this special issue, T. Koelewijn and colleagues were able to demonstrate a difference in the amount of pupil dilation among normal hearing participants listening to speech masked by fluctuating noise versus a single talker. The authors posit that the degree of pupil dilation may reflect the additional cognitive load resulting from the more difficult listening task and suggest that pupillometry may offer a viable objective measure of the benefits associated with specific hearing aid signal processing features such as digital noise reduction.

However, as illustrated by K. L. Tremblay and C. W. Miller, the successful use of hearing aids to improve human communication will ultimately depend on more than just brain measures. Many factors can contribute to aided speech understanding in noisy environments, including device centered (e.g., directional microphones, signal processing, and gain settings) and patient centered variables (e.g., age, attention, motivation, biology, personality, and lifestyle). Research aimed at exploring one variable in isolation (e.g., neural mechanisms underlying sound detection and/or discrimination) is likely to fall short when trying to optimize the many interactive stages involved in human communication.

To summarize and conclude, the collection of articles that appear in this special issue of IJO show that (1) it is possible to use brain measures to determine a person's neural detection of amplified sound and (2) brain measures can provide information about the use of this amplified signal sound. The use of brain measures to quantify and model neural mechanism associated with the perception of amplified sound is complex and sometimes counter-intuitive. For this reason, it is important to remind clinicians and neuroscientists that the interpretation of aided evoked neural activity cannot simply be based on prior published data acquired from normal hearing or unaided participants.

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Review Article

How Neuroscience Relates to Hearing Aid Amplification

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Hearing aids are used to improve sound audibility for people with hearing loss, but the ability to make use of the amplified signal, especially in the presence of competing noise, can vary across people. Here we review how neuroscientists, clinicians, and engineers are using various types of physiological information to improve the design and use of hearing aids.

1. Introduction

Despite advances in hearing aid signal processing over the last few decades and careful verification using recommended clinical practices, successful use of amplification continues to vary widely. This is particularly true in background noise, where approximately 60% of hearing aid users are satisfied with their performance in noisy environments [1]. Dissatisfaction can lead to undesirable consequences, such as discontinued hearing aid use, cognitive decline, and poor quality of life [2, 3].

Many factors can contribute to aided speech understanding in noisy environments, including device centered (e.g., directional microphones, signal processing, and gain settings) and patient centered variables (e.g., age, attention, motivation, and biology). Although many contributors to hearing aid outcomes are known (e.g., audibility, age, duration of hearing loss, etc.), a large portion of the variance in outcomes remains unexplained. Even less is known about the influence interacting variables can have on performance. To help advance the field and spawn new scientific perspectives, Souza and Tremblay [4] put forth a simple framework for thinking about the possible sources in hearing aid performance variability. Their review included descriptions of emerging technology that could be used to quantify the acoustic content of the amplified signal and its relation to perception. For example, new technological advances (e.g., probe microphone recordings using real speech) were making it possible to explore the relationship between amplified speech signals, at the level of an individual's ear, and the perception of those same signals. Electrophysiological recordings of amplified signals were also being introduced as a potential tool for assessing the neural detection of amplified sound. The emphasis of the framework was on signal audibility and the ear-to-brain upstream processes associated with speech understanding. Since that time, many new directions of research have emerged, as has an appreciation of the cognitive resources involved when listening to amplified sounds. We therefore revisit this framework when highlighting some of the advances that have taken place since the original Souza and Tremblay [4] article (e.g., SNR, listening effort, and the importance of outcome measures) and emphasize the growing contribution of neuroscience (Figure 1).

2. Upstream, Downstream, and Integrated Stages

A typical example highlighting the interaction between upstream and downstream contributions to performance outcomes is that involving the cocktail party. The cocktail party effect is the phenomenon of a listener being able to attend to a particular stimulus while filtering out a variety of competing stimuli, similar to partygoer focusing on a single conversation in a noisy room [5, 6]. The ability of a particular individual to "tune into" a single voice and "tune out" all that is coming out of their hearing aid is also an example of how variables specific to the individual can also contribute to performance outcomes.

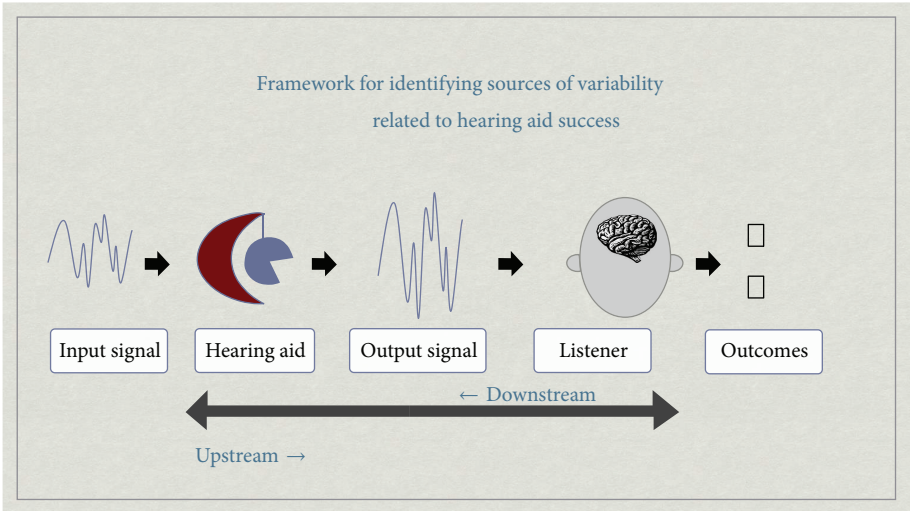


Figure 1: Framework for identifying sources of variability related to hearing aid success.

When described as a series of upstream events that could take place in someone's everyday life, the *input signal* refers to the acoustic properties of the incoming signal and/or the context in which the signal is presented. It could consist of a single or multiple talkers; it could be an auditory announcement projected overhead from a loudspeaker at the airport, or it could be a teacher giving homework instructions to children in a classroom. It has long been known that the ability to understand speech can vary in different types of listening environments because the signal-to-noise ratio (SNR) can vary from −2 dB, when in the presence of background noise outside the home, to +9 dB SNR, a level found inside urban homes [7]. Support for the idea that environmental SNR may influence a person's ability to make good use of their hearing aids comes from research showing that listeners are more dissatisfied and receive less benefit with their aids in noise than in quiet environments (e.g., [1, 8, 9]). From of a large-scale survey, two of the top three reasons for nonadoption of aids were that aids did not perform well in noise (48%) and/or that they picked up background sounds (45%; [10]). And of the people who did try aids, nearly half of them returned their aids due to lack of perceived benefit in noise or amplification of background noise. It is therefore not surprising that traditional hearing aid research has focused on hearing aid engineering in attempt to improve signal processing in challenging listening situations, so that optimal and audible signals can promote effective real-world hearing.

The next stage emphasizes the contribution ofthe *hearing aid* and how it modifies the acoustic signal (e.g., compression, gain and advanced signal processing algorithms). Examples include the study of real-world effectiveness of directional microphone and digital noise reduction features in hearing aids (e.g., [11, 12]). Amplification of background noise is one of the most significant consumer-based complaints associated with hearing aids, and directional hearing aids can improve the SNR of speech occurring in a noisy background (e.g., [13, 14]). However, these findings in the laboratory may not translate to perceived benefit in the real world. When participants were given a four-week take-home trial, omnidirectional microphones were preferred over directional microphones [15]. Over the past several decades, few advances in hearing aid technology have been shown to result in improved outcomes (e.g., [9, 16]). Thus, attempts at enhancing the quality of the signal do not guarantee improved perception. It suggests that something, in addition to signal audibility and clarity, contributes to performance variability.

What is received by the individual's auditory system is not the signal entering the hearing aid but rather a modified signal leaving the hearing aid and entering the ear canal. Therefore, quantification of the signal at the *output of the hearing aid* is an important and necessary step to understanding the biological processing of amplified sound. Although simple measures of the hearing aid output (e.g., gain for a given input level) in a coupler (i.e., simulated ear canal) have been captured for decades, current best practice guidelines highlight the importance of measuring hearing aid function in the listener's own ear canal. Individual differences in ear canal volume and resonance and how the hearing aid is coupled to an individual's ear can lead to significant differences in ear canal output levels [17]. Furthermore, as hearing aid analysis systems become more sophisticated, we are able to document the hearing aid response to more complex input signals such as speech or even speech and noise [18], which provides greater ecological validity than simple pure tone sweeps. In addition, hearing aid features can alter other acoustic properties of a speech signal. For example, several researchers have evaluated the effects of compression parameters on temporal envelope or the slow fluctuations in a speech signal [19–23], spectral contrast or consonant vowel ratio [20, 24–26], bandwidth [24], effective compression ratio [23, 24, 27], dynamic range [27], and audibility [24, 27, 28]. For example, as the number of compression channels increases, spectral differences between vowel formants decrease [26], the level of consonants compared to the level of vowels increases [29], and dynamic range decreases [27]. Similarly,

as compression time constants get shorter, the temporal envelope will reduce/smear [20, 21, 23] and the effective compression ratio will increase [27]. A stronger compression ratio has been linked to greater temporal envelope changes [21, 23]. Linear amplification may also create acoustic changes, such as changes in spectral contrast if the high frequencies have much more gain than the low frequencies (e.g., [24]). The acoustic changes caused by compression processing have been linked to perceptual changes in many cases [19–22, 24, 26, 30]. In general, altering compression settings (e.g., time constants or compression ratio) modifies the acoustics of the signal and the perceptual effects can be detrimental. For this reason, an emerging area of interest is to examine how frequency compression hearing aid technology affects the neural representation and perception of sound [31].

Characteristics of the *listener* (e.g., biology) can also contribute to a person's listening experience. Starting with bottom-up processing, one approach in neuroscience has been to model the auditory-nerve discharge patterns in normal and damaged ears in response to speech sounds so that this information can be translated into new hearing aid signal processing [32, 33]. The impact of cochlear dead regions on the fitting of hearing aids is another example of how biological information can influence hearing aid fitting [34]. Further upstream, Willott [49] established how aging and peripheral hearing loss affects sound transmission, including temporal processing, at higher levels in the brain. For this reason, brainstem and cortical evoked potentials are currently being used to quantify the neural representation of sound onset, offset and even speech envelope, in children and adults wearing hearing aids, to assist clinicians with hearing aid fitting [36–39]. When evoked by different speech sounds at suprathreshold levels, patterns of cortical activity (e.g., P1-N1-P2—also called acoustic change responses (ACC)) are highly repeatable in individuals and can be used to distinguish some sounds that are different from one another [4, 37]. Despite this ability, we and others have since shown that P1-N1-P2 evoked responses do not reliably reflect hearing aid gain, even when different types of hearing aids (analog and digital) and their parameters (e.g., gain and frequency response) are manipulated [40–45]. What is more, the signal levels of phones when repeatedly presented in isolation to evoke cortical evoked potentials are not the same as hearing aid output levels when phonemes are presented in running speech context [46]. These examples are provided because they reinforce the importance of examining the output of the hearing aid. Neural activity is modulated by both endogenous and exogenous factors and, in this example, the P1-N1-P2 complex was driven by the signal-to-noise ratio (SNR) of the amplified signal. Figure 2 shows the significant effect hearing aid amplification had on SNR when Billings et al. [43] presented a 1000 Hz tone through a hearing aid. Hearing aids are not designed to process steady-state tones, but results are similar even when naturally produced speech syllables were used [37]. Acoustic waveforms, recorded in-the-canal, are shown (unaided = left; aided=right). The output of the hearing aid, as measured at the 1000 Hz centered 1/3 octave band, was approximately equivalent at 73 and 74 dB SPL for unaided and aided conditions. Noise levels in that same 1/3 octave band, however, approximated 26 dB in the unaided condition and 54 dB SPL in the aided condition. Thus SNRs in the unaided and aided conditions, measured at the output of the hearing aid, were very different, and time-locked evoked brain activity shown in Figure 3 was influenced more by SNR than absolute signal level. Most of these SNR studies have been conducted in normal hearing listeners and thus the noise was audible, something unlikely to occur at some frequencies if a person has a hearing loss. Nevertheless, noise is always present in an amplified signal and contributors may range from amplified ambient noise to circuit noise generated by the hearing aid. It is therefore important to consider the effects of noise, among the many other modifications introduced by hearing aid processing (e.g., compression) on evoked brain activity. This is especially important because commercially available evoked potential systems are being used to estimate aided hearing sensitivity in young children [47].

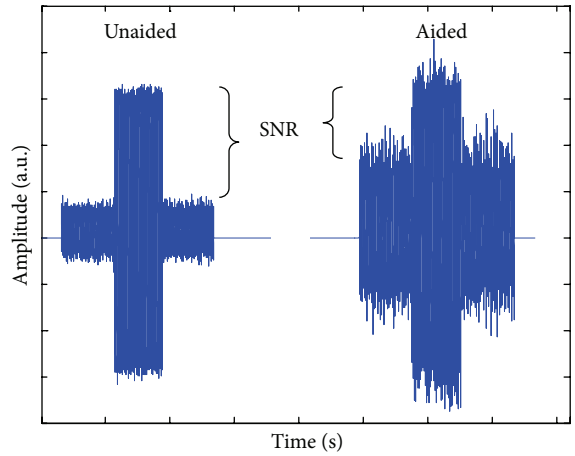
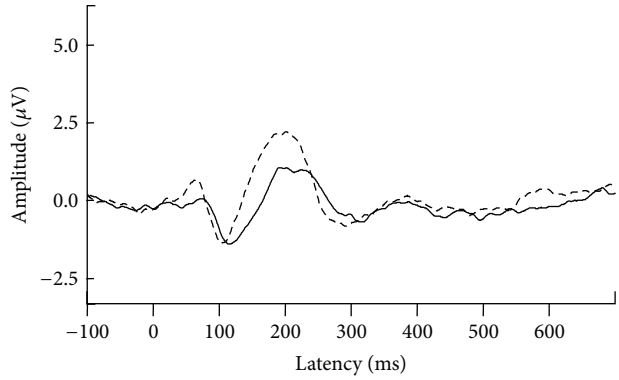


Figure 2: Time waveforms of in-the-canal acoustic recordings for one individual. The unaided (left) and aided (right) conditions are shown together. Signal output as measured at the 1000 Hz centered 1/3 octave band was approximately equivalent at 73 and 74 dB SPL for the unaided and aided conditions. However, noise levels in the same 1/3 octave band were approximately 26 and 54 dB SPL, demonstrating the significant change in SNR.

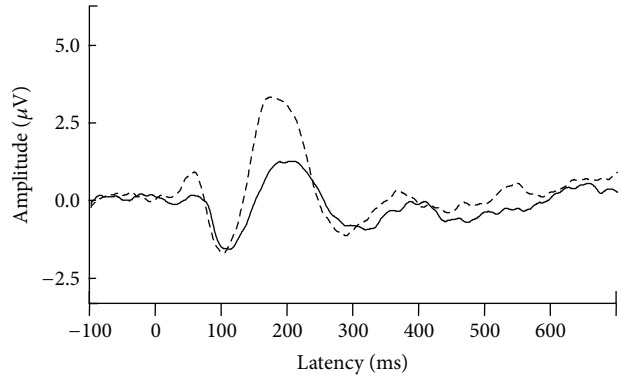
What remains unclear is how neural networks process different SNRs, facilitate the suppression of unwanted competing signals (e.g., noise), and process simultaneous streams of information when people with hearing loss wear hearing aids. Individual listening abilities have been attributed to variability involving motivation, selective attention, stream segregation, and multimodal interactions, as well as many other cognitive contributions [48]. It can be mediated by the biological consequences of aging and duration of hearing loss, as well as the peripheral and central effects of peripheral pathology (for reviews see [44, 49]). Despite the obvious importance of this stage and the plethora of papers published each year on the topics of selective attention, auditory streaming, object formation, and spatial hearing, the inclusion of people with hearing loss and who wear hearing aids remains relatively slim.



	Input signal	Gain	Output signal	Output noise	SNR
--- Unaided	40	—	44.0	26.4	17.6
—— Aided	40	0	43.5	31.9	11.6

*Input and output values are in dB SPL

(a)



	Input signal	Gain	Output signal	Output noise	SNR
--- Unaided	60	—	64.4	26.6	37.8
—— Aided	60	20	66.7	47.0	19.7

*Input and output values are in dB SPL

(b)

Figure 3: Two examples showing grand mean CAEPs recorded with similar mean output signal levels. Panels: (a) 40 dB input signals and (b) 60 dB input signals show unaided and aided grand mean waveforms evoked with corresponding in-the-canal acoustic measures. Despite similar input and output signal levels, unaided and aided brain responses are quite different. Aided responses are smaller than unaided responses, perhaps because the SNRs are poorer in the aided condition.

Over a decade ago, a working group that included scientists from academia and industry gathered and discussed the need to include central factors when considering hearing aid use [50] and since then there has been increased awareness about including measures of cognition, listening effort, and other top-down functions when discussing rehabilitation involving hearing aid fitting [51]. However, finding universally agreed upon definitions and methods to quantify cognitive function remains a challenge. Several self-report questionnaires and other subjective measures have evolved to measure listening effort, for example, but there are also concerns that self-report measures do not always correlate with objective measures [52, 53]. For this reason, new explorations involving objective measures are underway.

There have been tremendous advances in technology that permit noninvasive objective assessments of sensory and cognitive function. With this information it might become possible to harness cognitive resources in ways that have been previously unexplored. For example, it might become possible to use brain measures to guide manufacturer designs. Knowing how the auditory system responds to gain, noise reduction, and/or compression circuitry could influence future generations of biologically motivated changes in hearing aid design. The influence of brain responses is especially important with current advances in hearing aid design featuring binaural processing, which involve algorithms making decisions based on cues received from both hearing aids. Returning to the example of listening effort, pupillometry [54], an objective measure of pupil dilation, and even skin conductance (EMG activity; [55]) are being explored as an objective method for quantifying listening effort and cognitive load. Other approaches include the use of EEG and other neuropsychological correlates of auditive processing for the purpose of setting a hearing device by detecting listening effort [56]. In fact, there already exist a number of existing patents for this purpose by hearing aid manufacturers such as Siemens, Widex, and Oticon, to name a few. These new advances in neuroscience make it clear that multidisciplinary efforts that combine neuroscience and engineering and are verified using clinical trials are innovative directions in hearing aid science. Taking this point one step further, biological codes have been used to innervate motion of artificial limbs/prostheses, and it might someday be possible to design a hearing prosthesis that includes neuromachine interface systems driven by a person's listening effort or attention [57–59]. Over the last decade, engineers and neuroscientists have worked together to translate brain-computer-interface systems from the laboratory for widespread clinical use, including hearing loss [60]. Most recently, eye gaze is being used as a means of steering directional amplification. The visually guided hearing aid (VGHA) combines an eye tracker and an acoustic beam-forming microphone array that work together to tune in the sounds your eyes are directed to while minimizing others [61]. The VGHA is a lab-based prototype whose components connect via computers and other equipment, but a goal is to turn it into a wearable device. But, once again, the successful application of future BCI/VGHA devices will likely require interdisciplinary efforts, described within our framework, given that successful use of amplification involves more than signal processing and engineering.

If a goal of hearing aid research is to enhance and empower a person's listening experience while using hearing

aids, then a critical metric within this framework is the outcome measure. Quantifying a person's listening experience using a hearing aid as being positive [√] or negative [✗] might seem straight forward, but decades of research on the topic of outcome measures show this is not the case. Research aimed at modeling and predicting hearing aid outcome [9, 62] shows that there are multiple variables that influence various hearing aid outcomes. A person's age, their expectations, and the point in time in which they are queried can all influence the outcome measure. The type of outcome measure, self-report or otherwise, can also affect results. It is for this reason that a combination of measures (e.g., objective measures of speech-understanding performance; self-report measures of hearing aid usage; and self-report measures of hearing aid benefit and satisfaction) is used to characterize communication-related hearing aid outcome. Expanding our knowledge about the biological influences on speech understanding in noise can inspire the development of new outcome measures that are more sensitive to a listener's perception and to clinical interventions. For example, measuring participation in communication may assess a listener's use of their auditory reception on a deeper level than current outcomes asking how well speech is understood in various environments [63], which could be a promising new development in aided self-report outcomes.

3. Putting It All Together

Many factors can contribute to aided speech understanding in noisy environments, including device centered (e.g., directional microphones, signal processing, and gain settings) and patient centered variables (e.g., age, attention, motivation, and biology). The framework (Figure 1) proposed by Souza and Tremblay [4] provides a context for discussing the multiple stages involved in the perception of amplified sounds. What is more, it illustrates how research aimed at exploring one variable in isolation (e.g., neural mechanisms underlying auditory streaming) falls short of understanding the many interactive stages that are involved in auditory streaming in a person who wears a hearing aid. It can be argued that it is necessary to first understand how normal hearing ear-brain systems stream, but it can also be argued that interventions based on normal hearing studies are limited in their generalizability to hearing aid users.

A person's self-report or aided performance on an outcome measure can be attributed to many different variables illustrated in Figure 1. Each variable (e.g., input signal) could vary in different ways. One listener might describe themselves as performing well [√] when the input *signal* is a single speaker in moderate noise conditions, provided they are paying *attention* to the speaker while using a hearing aid that makes use of a *directional microphone*. This same listener might struggle [✗] if this single speaker is a lecturer in the front of a large classroom who paces back and forth across the stage and intermittently speaks into a microphone. In this example, changes in the quality and direction of a single source of input may be enough to negatively affect a person's use of sound upstream because of a reduced neural capacity to follow sounds when they change in location and in space. This framework and these examples are overly simplistic, but they are used to emphasize the complexity and multiple interactions that contribute to overall performance variability. We also argue that it is overly simplistic for clinicians and scientists to assume that explanations of performance variability rest solely one stage/variable. For this reason, interdisciplinary research that considers the contribution of neuroscience as an important stage along the continuum is encouraged.

The experiments highlighted here serve as examples to show how far, and multidisciplinary, hearing aid research has come. Since the original publication of Souza and Tremblay [4], advances have been made on the clinical front as shown through the many studies aimed at using neural detection measures to assist with hearing aid fitting. And it is through neuroengineering that that next generation of hearing prostheses will likely come.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Review Article

The Potential Role of the cABR in Assessment and Management of Hearing Impairment

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Hearing aid technology has improved dramatically in the last decade, especially in the ability to adaptively respond to dynamic aspects of background noise. Despite these advancements, however, hearing aid users continue to report difficulty hearing in background noise and having trouble adjusting to amplified sound quality. These difficulties may arise in part from current approaches to hearing aid fittings, which largely focus on increased audibility and management of environmental noise. These approaches do not take into account the fact that sound is processed all along the auditory system from the cochlea to the auditory cortex. Older adults represent the largest group of hearing aid wearers; yet older adults are known to have deficits in temporal resolution in the central auditory system. Here we review evidence that supports the use of the auditory brainstem response to complex sounds (cABR) in the assessment of hearing-in-noise difficulties and auditory training efficacy in older adults.

1. Introduction

In recent years, scientists and clinicians have become increasingly aware of the role of cognition in successful management of hearing loss, particularly in older adults. While it is often said that "we hear with our brain, not just with our ears," the focus of the typical hearing aid fitting continues to be one of providing audibility. Despite evidence of age-related deficits in temporal processing [1–6], abilities beyond the cochlea are seldom measured. Moreover, when auditory processing is assessed, behavioral measures may be affected by reduced cognitive abilities in the domains of attention and memory [7, 8]; for example, an individual with poor memory will struggle to repeat back long sentences in noise. The assessment and management of hearing loss in older adults would be enhanced by an objective measure of speech processing. The auditory brainstem response (ABR) provides such an objective measure of auditory function; its uses have included evaluation of hearing thresholds in infants, children, and individuals who are difficult to test, assessment of auditory neuropathy, and screening for retrocochlear function [9]. Traditionally, the ABR has used short, simple stimuli, such as pure tones and tone bursts, but the ABR has also been recorded to complex tones, speech, and music for more than three decades, with the ABR's frequency following response (FFR) reflecting the temporal discharge of auditory neurons in the upper midbrain [10, 11]. Here, we review the role of the ABR to complex sounds (cABR) in assessment and documentation of treatment outcomes, and we suggest a potential role of the cABR in hearing aid fitting.

2. The cABR Approach

The cABR provides an objective measure of subcortical speech processing [12, 13]. It arises largely from the inferior colliculus of the upper midbrain [14], functioning as

part of a circuit that interacts with cognitive, top-down influences. Unlike the click-evoked response, which bears no resemblance to the click waveform, the cABR waveform is remarkably similar to its complex stimulus waveform, whether a speech syllable or a musical chord, allowing for fine-grained evaluations of timing, pitch, and timbre representation. The click is short, nearly instantaneous, or approximately 0.1 ms, but the cABR may be elicited by complex stimuli that can persist for several seconds. The cABR's response waveform can be analyzed to determine how robustly it represents different segments of the speech stimulus. For example, in response to the syllable /da/, the onset of the cABR occurs at approximately 9 ms after stimulus onset, which would be expected when taking into account neural conduction time. The cABR onset is analogous to wave V of the brainstem's response to a click stimulus, but the cABR has potentially greater diagnostic sensitivity for certain clinical populations. For example, in a comparison between children with learning impairments versus children who are typically developing, significant differences were found for the cABR but not for responses to click stimuli [15]. The FFR comprises two regions: the transition region corresponding to the consonant-vowel (CV) formant transition and the steady-state region corresponding to the relatively unchanging vowel. The CV transition is perceptually vulnerable [16], particularly in noise, and the transition may be more degraded in noise than the steady state, especially in individuals with poorer speech-in-noise (SIN) perception [17].

The cABR is recorded to alternating polarities, and the average response to these polarities is added to minimize the cochlear microphonic and stimulus artifact [18, 19]. Phase locking to the stimulus envelope, which is noninverting, enhances representation of the envelope and biases the response towards the low frequency components of the response. On the other hand, phase locking to the spectral energy in the stimulus follows the inverting phase of the stimulus; therefore, adding responses to alternating polarities cancels out much of the spectral energy [13, 20]. Subtracting responses to alternating polarities, however, enhances the representation of spectral energy while minimizing the response to the envelope. One might choose to use added or subtracted polarities, or both, depending on the hypothetical question. For example, differences between good and poor readers are most prominent in the spectral region corresponding to the first formant of speech and are therefore more evident in subtracted polarities [21]. In contrast, the neural signature of good speech-in-noise perception is in the low frequency component of the response, which is most evident with added polarities [22]. The average response waveform of 17 normal hearing older adults (ages 60 to 67) and its evoking stimulus and stimulus and response spectra (to added and subtracted polarities) are displayed in Figure 1.

The cABR is acoustically similar to the stimulus. That is, after the cABR waveform has been converted to a .wav file, untrained listeners are able to recognize monosyllabic words from brainstem responses evoked by those words [23]. The fidelity of the response to the stimulus permits evaluation of the strength of subcortical encoding of multiple

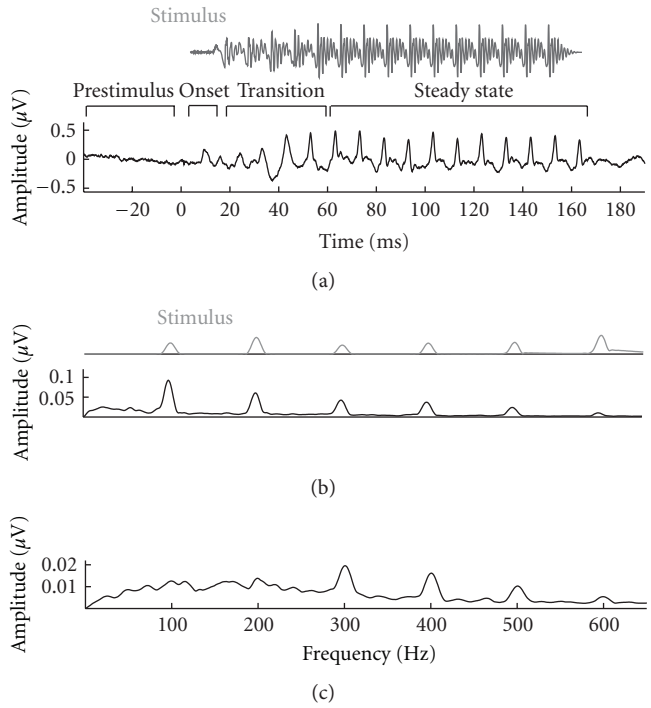

Figure 1: The stimulus /da/ (gray) is displayed with its response (black) in time and frequency domains. (a) Time domain. The response represents an average of 17 older adults (ages 60 to 67) all of whom have audiometrically normal hearing. The periodicity of the stimulus is reflected in the response with peaks repeating every ~10 ms (the F_0 of the vowel /a/). (b) and (c) Frequency domain. Fast Fourier transforms were calculated over the steady-state region of the response, showing frequency energy at the F_0 (100 Hz) and its integer harmonics for responses obtained by adding (b) and subtracting (c) responses to alternating polarities.

acoustic aspects of complex sounds, including timing (onsets, offsets), pitch (the fundamental frequency, F_0), and timbre (the integer harmonics of the F_0) [13]. Analyses of the cABR include measurement of latency and amplitude in the time domain and magnitude of the F_0 and individual harmonics in the frequency domain. Because of the cABR's remarkable stimulus fidelity, cross-correlation between the stimulus and the response also provides a meaningful measure [24]. In addition, responses between two conditions can be cross-correlated to determine the effects of a specific condition such as noise on a response [25].

Latency analysis has traditionally relied on picking individual peaks, a subjective task that is prone to error. Phase analysis provides an objective method for assessing temporal precision. Because the brainstem represents stimulus frequency differences occurring above 2000 Hz (the upper limits of brainstem phase locking) through timing [26] and phase representation [27, 28], the phase difference between two waveforms (in radians) can be converted to timing differences and represented in a "phaseogram." This analysis provides an objective measure of the response timing on a frequency-specific basis. For example, the brainstem's ability to encode phase differences in the formant trajectories between syllables

such as /ba/ and /ga/ can be assessed and compared to a normal standard or between groups in a way that would not be feasible if the analysis was limited to peak picking (Figure 2). Although the response peaks corresponding to the F_0 are discernible, the peaks in the higher frequency formant transition region such as in Figure 2 would be difficult to identify, even for the trained eye.

In natural speech, frequency components change rapidly, and a pitch tracking analysis can be used to evaluate the ability of the brainstem to encode the changing fundamental frequency over time. From this analysis, a measure of pitch strength can be computed using short-term autocorrelation, a method which determines signal periodicity as the signal is compared to a time-shifted copy of itself. Pitch-tracking error is determined by comparing the stimulus F_0 with the response F_0 for successive periods of the response [29, 30]. These and other measures produced by the pitch-tracking analysis reveal that the FFR is malleable and experience dependent, with better pitch tracking in individuals who have heard changing vowel contours or frequency sweeps in meaningful contexts, such as in tonal languages or music [24, 31].

Other automated analyses which could potentially be incorporated into a clinical protocol include the assessment of response consistency and phase locking. Response consistency provides a way of evaluating trial-to-trial within-subject variability, perhaps representing the degree of temporal jitter or asynchronous neural firing that might be seen in an impaired or aging auditory system [6]. Auditory neuropathy spectrum disorder would be an extreme example of dyssynchronous neural firing, affecting even the response to the click [32–34]. A mild form of dyssynchrony, however, may not be evident in the results of the typical audiologic or ABR protocol but might be observed in a cABR with poor response consistency. The phase-locking factor is another measure of response consistency, providing a measure of trial-to-trial phase coherence [35, 36]. Phase locking refers to the repetitive neural response to periodic sounds. While response consistency is determined largely by the stimulus envelope, the phase-locking factor is a measure of consistency of the stimulus-evoked oscillatory activity [37].

3. The cABR and Assessment of Hearing Loss and the Ability to Hear in Noise

The cABR may potentially play an important role in assessment of hearing loss and hearing in noise. It has good test-retest reliability [39, 40], a necessity for clinical comparisons and for documentation of treatment outcomes. Just as latency differences of 0.2 ms for brainstem responses to click stimuli can be considered clinically significant when screening for vestibular schwannomas [9], similar differences on the order of fractions of milliseconds in the cABR have been found to reliably separate clinical populations [41, 42]. Banai et al. [41] found that the onset and other peaks in the cABR are delayed 0.2 to 0.3 ms in children who are good readers compared to poor readers. In older adults, the offset latency is a strong predictor of self-assessed SIN perception in older adults, with latencies ranging from 47 to 51 ms in responses to a 40 ms /da/ (formant transition only) [43]. Temporal processing deficits

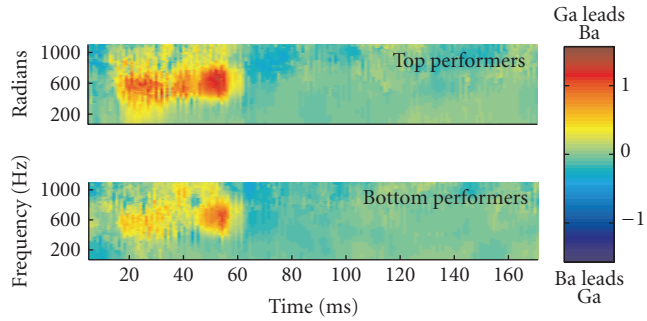

Figure 2: A phaseogram displaying differences in phase (radians, colorbar) in responses to /ba/ and /ga/ syllables, which have been synthesized so that they differ only in the second formant of the consonant-to-vowel transition. The top and bottom groups are children (ages 8 to 12) who differ on a speech-in-noise perception measure, the Hearing in Noise Test (HINT). The red color indicates greater phase difference, with /ga/ preceding /ba/, as expected given cochlear tonotopicity. Note that phase differences are only present in the transition, not in the steady state, during which the syllables are identical. Modified from [27].

are also seen in children with specific language impairment, who have decreased ability to track frequency changes in tonal sweeps, especially at faster rates [44].

Because of the influence of central and cognitive factors on speech-in-noise perception, the pure-tone audiogram, a largely peripheral measure, does not adequately predict the ability to hear in background noise, especially in older adults [45–47]. Due to the convergence of afferent and efferent transmission in the inferior colliculus (IC) [48, 49], we propose that the cABR is an effective method for assessing the effects of sensory processing and higher auditory function on the IC. While the cABR does not directly assess cognitive function, it is influenced by higher-level processing (e.g., selective attention, auditory training). The cABR is elicited passively without the patient's input or cooperation beyond maintaining a relaxed state, yet it provides in essence a snapshot in time of auditory processing that reflects both cognitive (auditory memory and attention) and sensory influences.

In a study of hearing-, age-, and sex-matched older adults (ages 60–73) with clinically normal hearing, the older adults with good speech-in-noise perception had more robust subcortical stimulus representation, with higher root-mean-square (RMS) and F_0 amplitudes compared to older adults with poor speech-in-noise perception (Figure 3) [38]. Perception of the F_0 is important for object identification and stream segregation, allowing us to attend to a single voice from a background of voices [50]; therefore, greater representation of the F_0 in subcortical responses may enhance one's ability to hear in noise. When we added noise (six-talker babble) to the presentation of the syllable, we found that the responses of individuals in the top speech-in-noise group were less degraded than in the bottom speech-in-noise group (Figure 3). These results are consistent with research from more than two decades documenting suprathreshold deficits

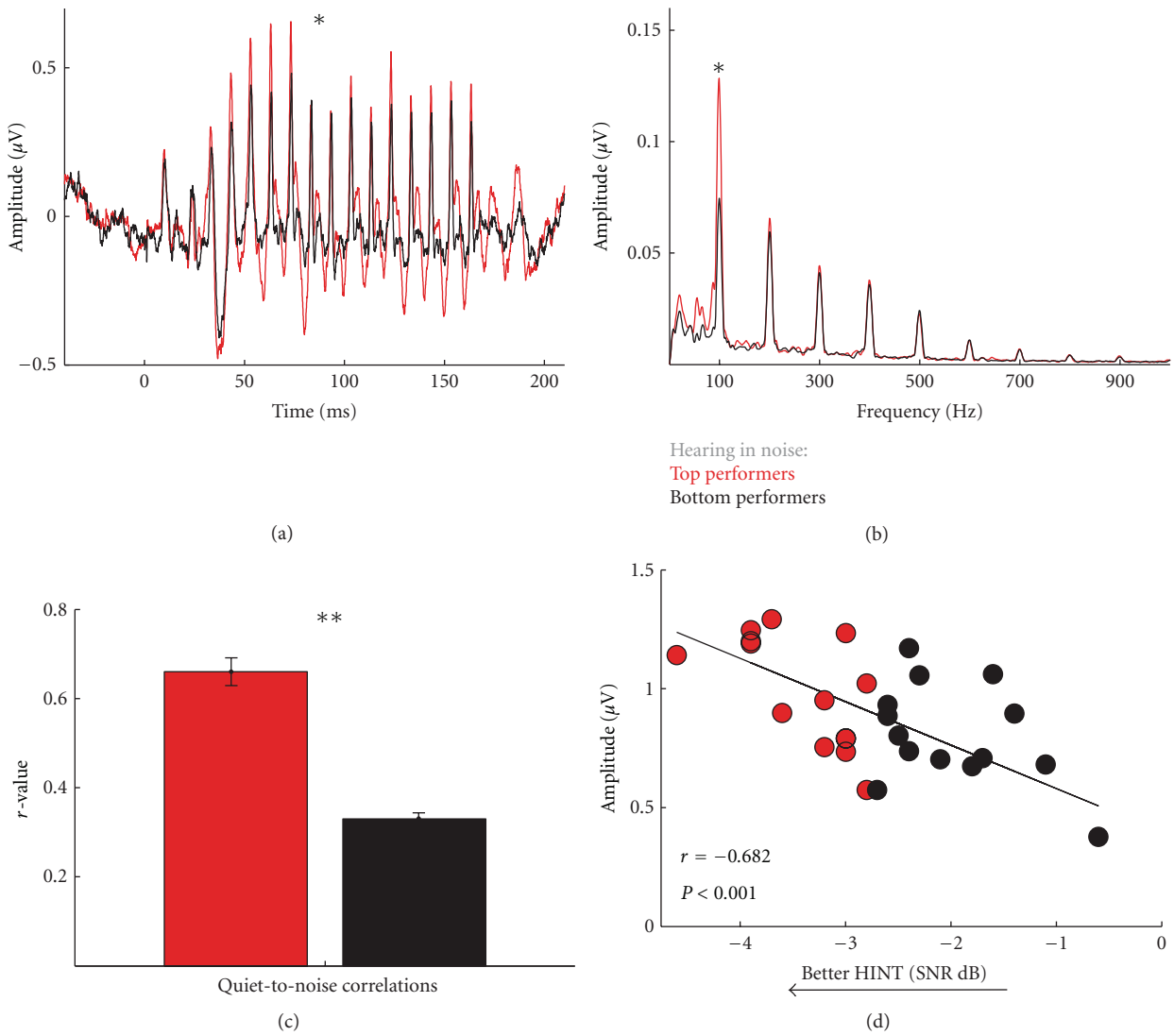

Figure 3: Responses to the syllable /da/ are more robust in older adults with good speech-in-noise perception compared to those with poor speech-in-noise perception, demonstrated by greater RMS amplitude (a) and amplitude of the F_0 in the good speech-in-noise group (b). The responses in the poor speech-in-noise group were more susceptible to the degrading effects of noise, as shown by greater differences in responses to the /da/ in quiet and noise (cross-correlations) (c). Relationship between speech-in-noise perception and the quiet-noise correlation (d). $^{*}P < 0.05$, $^{**}P < 0.01$. Modified from [38].

that cannot be identified by threshold testing [46, 47, 51–58]. Even in normal-hearing young adults, better speech-in-noise perception is related to more robust encoding of the F_0 in the cABR [53]. Furthermore, in a study with young adult participants, Ruggles et al. [51] found that spatial selective auditory attention performance correlates with the phase locking of the FFR to the speech syllable /da/. Furthermore, they found that selective attention correlates with the ability to detect frequency modulation but is not related to age, reading span, or hearing threshold.

The cABR provides evidence of age-related declines in temporal and spectral precision, providing a neural basis for speech-in-noise perception difficulties. In older adults, delayed neural timing is found in the region corresponding to the CV formant transition [59, 60], but timing in the steady-state region remains unchanged. Importantly, age-related differences are seen in middle-aged adults as young as 45, indicating that declines in temporal resolution are not limited to the elderly population. Robustness of frequency representation also decreases with age, with the amplitude of the fundamental frequency declining in middle- and in older-aged adults. These results provide neural evidence for the finding of adults having trouble hearing in noise as soon as the middle-aged years [61].

What is the role of the cABR in clinical practice? The cABR can be collected in as little as 20 minutes, including electrode application. Nevertheless, even an additional twenty minutes would be hard to add to a busy practice. To be efficacious, the additional required time must yield information not currently provided by the existing protocol.

One of the purposes of an audiological evaluation is to determine the factors that contribute to the patient's self-perception of hearing ability. To evaluate the effectiveness of possible factors, we used multiple linear regression modeling to predict scores on the speech subtest of the Speech, Spatial, and Qualities Hearing Scale [62]. Pure-tone thresholds, speech-in-noise perception, age, and timing measures of the cABR served as meaningful predictors. Behavioral assessments predicted 15% of the variance in the SSQ score, but adding brainstem variables (specifically the onset slope, offset latency, and overall morphology) predicted an additional 16% of the variance in the SSQ (Figure 4). Therefore, the cABR can provide the clinician with unique information about biological processing of speech [43].

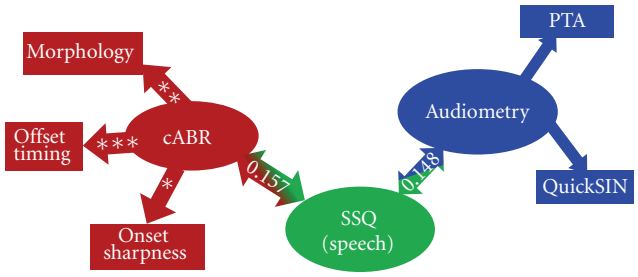

Figure 4: Self-perception of speech, assessed by the Speech Spatial Qualities Hearing scale (SSQ), is predicted by audiologic and cABR measures. The audiometric variables predict 15% of the variance in SSQ; the cABR variables predict an additional 16%. In the multiple linear regression model, only the contributions of the cABR onset time and morphology variables are significant. $^{*}P < 0.05$, $^{***}P < 0.01$.

4. The cABR is Experience Dependent

As the site of intersecting afferent and efferent pathways, the inferior colliculus plays a key role in auditory learning. Indeed, animals models have demonstrated that the cortico-collicular pathway is essential for auditory learning [63, 64]. Therefore, it is reasonable to expect that the cABR reflects evidence of auditory training; in fact, the cABR shows influences of both life-long and short-term training. For example, native speakers of tonal languages have better brainstem pitch tracking to changing vowel contours than speakers of non-tonal languages [24]. Bilingualism provides another example of the auditory advantages conferred by language expertise. Bilingualism is associated with enhanced cognitive skills, such as language processing and executive function, and it also promotes experience-dependent plasticity in subcortical processing [65]. Bilingual adolescents, who reported high English and Spanish proficiency, had more robust subcortical encoding of the F_0 to a target sound presented in a noisy background than their age-, sex-, and IQ-matched monolingual peers. Within the bilingual group, a measure of sustained attention was related to the strength of the F_0; this relation between attention and the F_0 was not seen in the monolingual group. Krizman et al. [65] proposed that diverse language experience heightens directed attention toward linguistic inputs; in turn, this attention becomes increasingly focused on features important for speaker identification and stream segregation in noise, such as the F_0.

Musicianship, another form of auditory expertise, also extends to benefits of speech processing; musicians who are nontonal language speakers have enhanced pitch tracking to linguistically relevant vowel contours, similar to that of tonal language speakers [31]. Ample evidence now exits for the effects of musical training on the cABR [28, 60, 67–73]. The OPERA (Overlap, Precision, Emotion, Repetition, and Attention) hypothesis has been proposed as the mechanism by which music engenders auditory system plasticity [74]. For example, there is overlap in the auditory pathways for speech and music, explaining in part the musician's superior abilities for neural speech-in-noise processing. The focused attention required for musical practice and performance results in strengthened sound-to-meaning connections, enhancing top-down cognitive (e.g., auditory attention and memory) influences on subcortical processing [75].

Musicians' responses to the cABR are more resistant to the degradative effects of noise compared to nonmusicians [68, 73]. Background noise delays and reduces the amplitude of the cABR [76]; however, musicianship mitigates the effects of six-talker babble noise on cABR responses in young adults, with earlier peak timing of the onset and the transition in musicians compared to nonmusicians. Bidelman and Krishnan [73] evaluated the effects of reverberation on the FFR and found that reverberation had no effect on the neural encoding of pitch but significantly degraded the representation of the harmonics. In addition, they found that young musicians had more robust responses in quiet and in most reverberation conditions. Benefits of musicianship have also been seen in older adults; when comparing effects of aging in musicians and nonmusicians, the musicians did not have the expected age-related neural timing delays in the CV transition indicating that musical experience offsets the effects of aging [60]. These neural benefits in older musicians are accompanied by better SIN perception, temporal resolution, and auditory memory [77].

But, what about the rest of us who are not able to devote ourselves full time to music practice—can musical training improve our auditory processing as well? Years of musical training in childhood are associated with more robust responses in adults [67], in that young adults with zero years of musical training had responses closer to the noise floor compared to groups of adults with one to five or six to eleven years of training who had progressively larger signal-to-noise ratios. In a structural equation model of the factors predicting speech-in-noise perception in older adults, two subsets were compared—a group who had no history of musical training and another group who had at least one year of musical training (range 1 year to 45 years). Cognitive factors (memory and attention) played a bigger role in speech-in-noise perception in the group with musical training, but life experience factors (physical activity and socioeconomic status) played a bigger role in the group with no experience. Subcortical processing (pitch encoding, harmonic encoding, and cross-correlations between responses in quiet and noise)

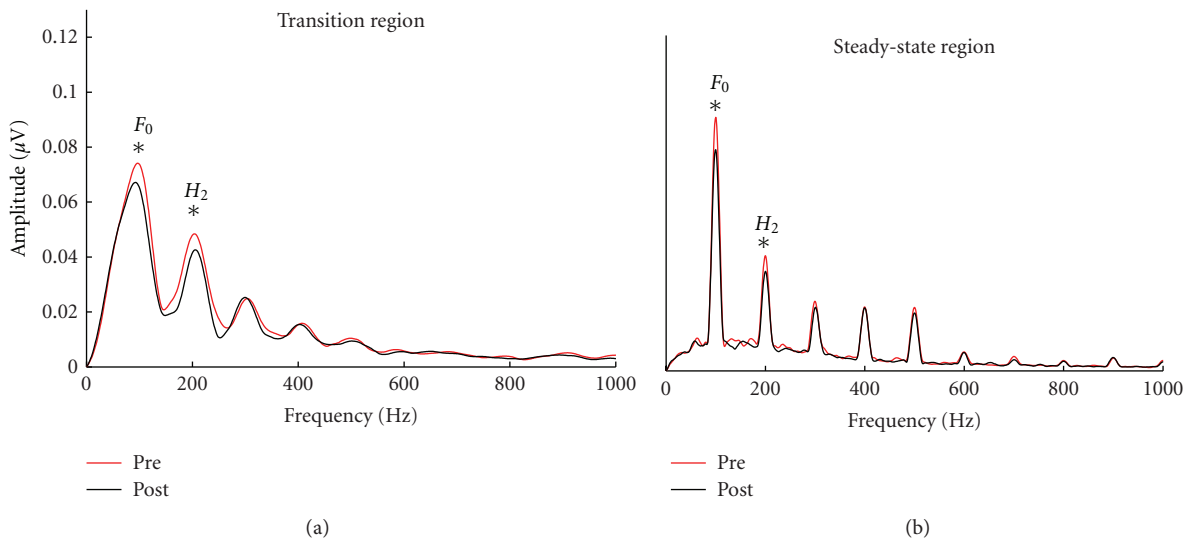

Figure 5: Young adults with normal hearing have greater representation of the F_0 in subcortical responses to /da/ presented in noise after undergoing LACE auditory training. The F_0 and the second harmonic have greater amplitudes in the postcondition when calculated over the transition (20–60 ms) (b) and the steady state (60–170 ms) (a). Modified from [66].

accounted for a substantial amount of the variance in both groups [78].

Short-term training can also engender subcortical plasticity. Carcagno and Plack [79] found changes in the FFR after ten sessions of pitch discrimination training that took place over the course of approximately four weeks. Four groups participated in the experiment: three experimental groups (static tone, rising tone, and falling tone) and one control group. Perceptual learning occurred for the three experimental groups, with effects somewhat specific to the stimulus used in training. These behavioral improvements were accompanied by changes in the FFR, with stronger phase locking to the F_0 of the stimulus, and changes in phase locking were related to changes in behavioral thresholds.

Just as long-term exposure to tonal language leads to better pitch tracking to changing vowel contours, just eight days of vocabulary training on words with linguistically relevant contours resulted in stronger encoding of the F_0 and decreases in the number of pitch-tracking errors [29]. The participants in this study were young adults with no prior exposure to a tonal language. Although the English language uses rising and falling pitch to signal intonation, the use of dipping tone would be unfamiliar to a native English speaker, and, interestingly, the cABR to the dipping tone showed the greatest reduction in pitch-tracking errors.

Training that targets speech-in-noise perception has also shown benefits at the level of the brainstem [80]. Young adults were trained to discriminate between CV syllables embedded in a continuous broad-band noise at a +10 dB signal-to-noise ratio. Activation of the medial olivocochlear bundle (MOCB) was monitored during the five days of training through the use of contralateral suppression of evoked otoacoustic emissions. Training improved performance on the CV discrimination task, with the greatest improvement occurring over the first three training days. A significant increase in MOCB activation was found, but only in the participants who showed robust improvement (learners). The learners showed much weaker suppression than the nonlearners on the first day; in fact, the level of MOCB activation was predictive of learning. This last finding would be particularly important for clinical purposes—a measure predicting benefit would be useful for determining treatment candidacy.

There is renewed clinical interest in auditory training for the management of adults with hearing loss. Historically, attempts at auditory training had somewhat limited success, partly due to constraints on the clinician's ability to produce perceptually salient training stimuli. With the advent of computer technology and consumer-friendly software, auditory training has been revisited. Computer technology permits adaptive expansion and contraction of difficult-to-perceive contrasts and/or unfavorable signal-to-noise ratios. The Listening and Communication Enhancement program (LACE, Neurotone, Inc., Redwood City, CA) is an example of an adaptive auditory training program that employs top-down and bottom-up strategies to improve hearing in noise. Older adults with hearing loss who underwent LACE training scored better on the Quick Speech in Noise test (QuickSIN) [81] and the hearing-in-noise test (HINT) [82]; they also reported better hearing on self-assessment measures—the Hearing Handicap Inventory for the Elderly/Adults [83] and the Client Oriented Scale of Improvement [84, 85]. The control group did not show improvement on these measures.

The benefits on the HINT and QuickSIN were replicated in young adults by Song et al. [66]. After completing 20 hours of LACE training over a period of four weeks, the participants improved not only on speech-in-noise performance but also had more robust speech-in-noise representation in the cABR (Figure 5). They had training-related increases in the subcortical representation of the F_0 in response to speech sounds presented in noise but not in quiet. Importantly, the

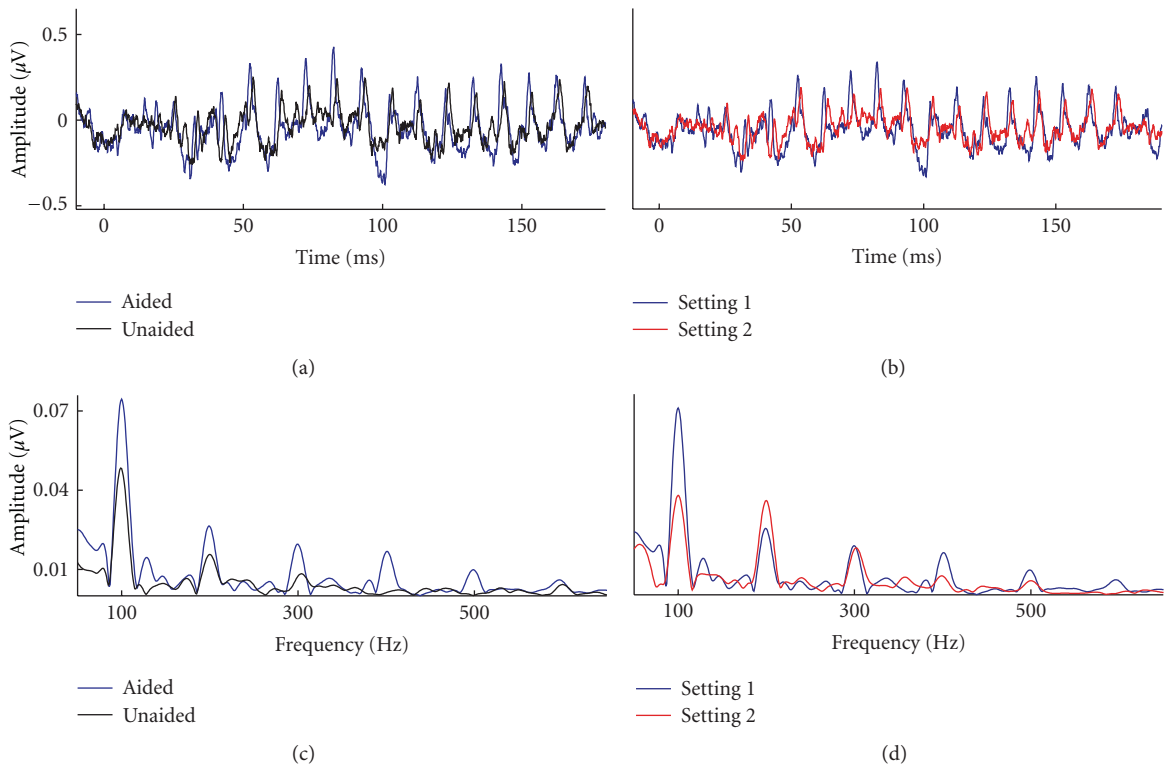

Figure 6: Responses were obtained to the stimulus /da/ presented at 80 dB SPL in sound field in aided (blue) versus unaided (black) conditions ((a) and (c)) and different settings in the same hearing aid ((b) and (d)). Responses show greater RMS and F_0 amplitudes in aided versus unaided conditions and for setting 1 versus setting 2.

amplitude of the F_0 at pretest predicted training-induced change in speech-in-noise perception. The advantages of computer-based auditory training for improved speech-in-noise perception and neural processing have also been observed in older adults [86]. Based on this evidence, the cABR may be efficacious for documenting treatment outcomes, an important component of evidence-based service.

5. The cABR and Hearing Aid Fitting

Any clinician who has experience with fitting hearing aids has encountered the patient who continues to report hearing difficulties, no matter which particular hearing aid or algorithm is tried. Although we have not yet obtained empirical evidence on the role of the cABR in the hearing aid fitting, we suggest that implementation of the cABR may enhance hearing aid fittings, especially in these difficult-to-fit cases. The clinician might be guided in the selection of hearing aid algorithms through knowledge of how well the brainstem encodes temporal and spectral information. For example, an individual who has impaired subcortical timing may benefit from slowly changing compression parameters in response to environmental changes.

We envision incorporating the cABR into verification of hearing aid performance. Cortical-evoked potentials have been used for verifying auditory system development after hearing aid or cochlear implant fitting in children [87–89]. In adults, however, no difference is noted in the cortical response between unaided and aided conditions, indicating that the cortical response may reflect signal-to-noise ratio rather than increased gain from amplification [90]. Therefore, cortical potentials may have limited utility for making direct comparisons between unaided and aided conditions in adults. We recently recorded the cABR in sound field and compared aided and unaided conditions and different algorithms in the aided condition. There is a marked difference in the amplitude of the waveform in response to an aided compared to an unaided condition. By performing stimulus-to-response correlations, it is possible to demonstrate that certain hearing aid algorithms resulted in a better representation of the stimulus than others (Figure 6). These preliminary data demonstrate the feasibility and possibility of using this approach. Importantly, these data also demonstrate meaningful differences easily observed in an individual.

6. Conclusions

With improvements in digital hearing aid technology, we are able to have greater expectations for hearing aid performance than ever before, even in noisy situations [91]. These improvements, however, do not address the problems we continue to encounter in challenging hearing aid fittings that leave us at a loss for solutions. The cABR provides an opportunity to evaluate and manage an often neglected part of hearing—the central auditory system—as well as the

biological processing of key elements of sound. We envision future uses of the cABR to include assessment of central auditory function, prediction of treatment or hearing aid benefit, monitoring treatment or hearing aid outcomes, and assisting in hearing aid fitting. Because the cABR reflects both sensory and cognitive processes, we can begin to move beyond treating the ear to treating the person with a hearing loss.

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Research Article

Electroacoustic Comparison of Hearing Aid Output of Phonemes in Running Speech versus Isolation: Implications for Aided Cortical Auditory Evoked Potentials Testing

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Background. Functioning of nonlinear hearing aids varies with characteristics of input stimuli. In the past decade, aided speech evoked cortical auditory evoked potentials (CAEPs) have been proposed for validation of hearing aid fittings. However, unlike in running speech, phonemes presented as stimuli during CAEP testing are preceded by silent intervals of over one second. Hence, the present study aimed to compare if hearing aids process phonemes similarly in running speech and in CAEP testing contexts. *Method*. A sample of ten hearing aids was used. Overall phoneme level and phoneme onset level of eight phonemes in both contexts were compared at three input levels representing conversational speech levels. *Results*. Differences of over 3 dB between the two contexts were noted in one-fourth of the observations measuring overall phoneme levels and in one-third of the observations measuring phoneme onset level. In a majority of these differences, output levels of phonemes were higher in the running speech context. These differences varied across hearing aids. *Conclusion*. Lower output levels in the isolation context may have implications for calibration and estimation of audibility based on CAEPs. The variability across hearing aids observed could make it challenging to predict differences on an individual basis.

1. Introduction

Hearing aid validation using aided speech evoked auditory evoked potentials is of research and clinical interest. Such measurements involve elicitation of an evoked potential using a speech stimulus that has been processed through a hearing aid. Hearing aids, being mostly nonlinear, may have implications for the nature of speech stimulus used as input. The present study focuses on the effect of nonlinear hearing aid processing on speech stimuli used for measurement of cortical auditory evoked potentials (CAEPs).

Nonlinear hearing aids are sensitive to the characteristics of input stimuli. Factors such as input level, duration, crest factor (ratio of peak to root mean square (RMS) amplitude), modulation depth, and modulation frequency of the input signal may affect the gain applied by the hearing aid, in ways that would not occur with a linear system [1–4]. These effects have been attributed to the level-dependent signal processing architecture, which in many hearing aids includes frequency specific compression threshold, compression ratio, compression time constants, number of channels, gain in each channel, expansion threshold, and expansion time constants [1, 5–12]. In addition, hearing aid processing may also consider the frequency characteristics of the input stimulus (e.g., [13, 14]). Hence the output of a hearing aid to a specific input is the product of complex interactions between input stimuli and hearing aid features that may or may not be known to or may not be adjustable by the end user.

Nonlinear hearing aids, being sensitive to features of the input signal, process speech or speech-like stimuli differently

from nonspeech stimuli [3, 7, 10, 15]. Since the main goal of hearing aid validation procedures is to assess benefit of hearing aid use while listening to speech, it is preferable that such procedures use speech stimuli in the most natural or frequently encountered form as possible. Behavioural validation procedures (tests that require active participation of the hearing aid user) such as speech tests, mostly use speech in various natural forms. Examples include the use of sentence materials, such as the Bamford-Kowal-Bench sentence test [16], or materials with less grammatical context such as isolated words or nonsense syllables (e.g., The Nonsense Syllable test [17]). But the speech stimuli may need to be modified for use in alternative validation methods such as aided auditory evoked potentials [18–23].

Aided auditory evoked potentials are objective and electrophysiological (they record neural responses to sound) but historically have not used speech stimuli. Of these, one of the reasons CAEPs have been of interest in the validation of hearing aid fittings is because natural speech sounds can be used as stimuli [19, 23–27]. Often phonemes or syllables excised from running speech or from standard speech tests have been used to record reliable CAEPs (e.g., [27–29]). Although natural speech can be used as stimuli, CAEP testing involves presentation of these stimuli with interstimulus intervals (ISI). These ISIs usually range on the order of 1-2 seconds (e.g., [23, 29, 30]) optimized for the latency of CAEPs and refractory periods of the cortical pyramidal neurons [30–32]. These stimuli are repeated 100–200 times, with constant or slightly variable ISIs and CAEPs elicited to each of the presentations are averaged. Presence of a CAEP elicited by a specific stimulus is interpreted as the stimulus being relayed to the source of CAEPs, the auditory cortex [21, 24]. Evidence suggests that CAEP thresholds (i.e., the lowest stimulus level at which a CAEP is detected) are closely related to behavioral thresholds (i.e., the lowest stimulus level at which the participant detects the stimulus) [33, 34]. Therefore, presence of a CAEP is likely to suggest audibility of the eliciting stimulus. On these premises, recent aided CAEP protocols for hearing aid validation have used brief segments of speech in the form of phonemes or syllables (e.g., [21–25]). Depending on their length, these brief segments may differ in their representation of certain features cues such as formant transitions, compared to longer segments of these same phonemes embedded in running speech. Commercial equipment such as the HEARLab uses phonemes, sampled across the speech frequency range presented at their naturally occurring levels within running speech, and presented in isolation to permit averaging of CAEP across several sweeps [35].

Phonemes presented in isolation for CAEP protocols may differ in several important ways from phonemes presented within running speech. In CAEP protocols, the target phoneme is preceded by an ISI (a silence period) whereas the same phoneme in running speech is likely to be preceded by other phonemes. Since nonlinear hearing aids continuously and rapidly adjust band-specific gains based on the acoustic input, there is a possibility that the hearing aids may react differently to the same phoneme when presented during aided CAEP testing as compared to when they occur in running speech. With 1-2 seconds of ISI preceding every repetition of the stimulus, nonlinear hearing aids may demonstrate an overshoot at the onset of the stimulus consistent with compression circuitry [36]. Also, hearing aids of different models and different manufacturers may vary in how quickly they respond to changes in the acoustic input. Therefore, verifying that hearing aid output is comparable for phonemes presented in these two contexts (preceding silent periods/ISI versus embedded in running speech) may be an important step in evaluating the validity of using CAEP protocols in hearing aid validation. Previous reports on non-CAEP related measures suggest that certain features of nonlinear signal processing in hearing aids may attenuate the level of speech sounds immediately preceded by silence [37, 38].

The effects of CAEP protocols on the gain achieved while processing tone bursts have been reported elsewhere in this issue [40, 41]. These studies provide evidence that hearing aid gain differs for tone bursts (short and long) presented in isolation versus pure tones that are continuous. Specifically, the gain achieved during processing of tone bursts was lower than the verified gain, when measured at 30 ms poststimulus onset and at maximum amplitude. Onset level is of interest because the first 30 to 50 ms of the stimulus primarily determines the characteristics of the elicited CAEP [42]. Stimulus level of the hearing aid processed tone bursts was positively related to the CAEP amplitude, with stimulus level at 30 ms poststimulus onset being a better predictor of CAEP amplitude compared to maximum stimulus level. These reports [40, 41] substantiate the need to verify output levels of CAEP stimuli across contexts, and to consider stimulus onsets. The present study will focus upon aided processing of phonemes across contexts and measure both overall level (level measured across the entire duration of the phoneme) and onset level of the stimuli at the output of the hearing aid.

The purpose of this study was to understand if hearing aids process CAEP phonemes presented in isolation differently to phonemes presented in running speech. The primary outcome measure of interest in this study was the output level of phonemes in both contexts. Findings from this study may provide some insights into the design of hearing aid validation protocols that employ aided CAEP measures, because large differences in hearing aid output arising due to stimulus context may influence interpretation of audibility based on aided CAEPs.

2. Method

2.1. Hearing Aids. Ten hearing aids sampled across various manufacturers were chosen. A list of the hearing aids used is provided in Table 1. Hearing aids were sampled across a representative range of major manufacturers and were behind-the-ear (BTE) in style. Of the 10 hearing aids, six were programmed and verified to meet DSL v5a adult prescription targets [43] for an N4 audiogram [39]. The N4 audiogram represents hearing loss of moderate to severe degree with thresholds of 55 dB HL at 250 Hz worsening down to 80 dB HL at 6 kHz [39]. The remaining four hearing

Table 1

Hearing aids for N4 audiogram	Hearing aids for N6 audiogram
Oticon Agil Pro P	*Oticon Chilli SP*
Phonak Nios Micro V	*Phonak Naida IX SP*
Siemens Aquaris 701	*Unitron 360+*
Widex Mind 330	*Starkey S series IQ 11*
Unitron Passport	
Siemens Motion 701p	

aids were programmed and verified to meet DSL v5a targets for an N6 audiogram. The N6 audiogram represents hearing loss of severe degree with thresholds ranging from 75 dB HL at 250 Hz worsening to 100 dB HL at 6 kHz [39]. The frequency specific thresholds of the two audiograms used are provided in Table 2. Hearing aids appropriate for different audiograms were chosen from different manufacturers to obtain a representative sample of commonly available commercial products. All hearing aids were programmed to function on a basic program with all additional features such as noise reduction, feedback cancellation, and frequency lowering disabled during verification and recording. As such, variance across devices is mainly attributable to the nonlinear characteristics of the devices, in isolation of these other aspects of hearing aid signal processing.

2.2. Stimuli. Stimuli were constructed to have both running speech and phoneme-in-isolation contexts as follows. For the running speech context, eight phonemes (/a/, /i/, /u/, /s/, /ʃ/, /m/, /t/, and /g/) were identified within a recording of the Rainbow passage. The passage was spoken by a male talker and lasted 2 minutes and 14 seconds. Aided recordings of this passage were made for each hearing aid, and the level of each phoneme was measured from within the aided passage. For the isolated context, the same phonemes and phoneme boundaries were used, but were excised from the passage for use as individual stimuli. Boundaries of these phonemes were chosen such that any transitions preceding and following these phonemes due to coarticulation were excluded. The duration of each of the phonemes are as follows: /a/—87 ms, /i/—84 ms, /u/—124 ms, /s/—133 ms, /ʃ/—116 ms, /m/—64 ms, /t/—26 ms, and /g/—19 ms. The durations of these phonemes differed naturally and were not modified in order to allow direct comparisons between the two contexts. These specific phonemes were chosen as the first six of these phonemes are a part of the commonly used Ling 5 or 6 sounds test [44, 45]. The last three have been commonly used in a series of aided CAEP studies (e.g., [26, 27, 46]) and are also a part of the stimulus choices available in the HEARLab [35]. A silent interval of 1125 ms preceding each phoneme was created using sound editing software Goldwave (v.5.58). This is to simulate a CAEP stimulus presentation protocol where the ISI usually ranges between one and two seconds.

2.3. Recording Apparatus. Recordings of hearing aid output used a click-on coupler (Brüel & Kjær (B&K) type 4946 conforming to ANSI S3.7, IEC 60126 fitted with microphone type 4192) with an earplug simulator. The hearing aid was connected via 25 mm of size 13 tubing [47]. This was set up in a B&K anechoic box (Box 4232) that also housed a reference microphone. Stimuli were presented through the speaker housed in the box. The outputs of the reference and coupler microphones were captured in SpectraPLUS (v5.0.26.0) in separate channels using a sampling rate of 44.1 kHz with 16-bit sampling precision. SpectraPLUS software was used to record the reference and coupler signals as .wav files for further signal analyses.

2.4. Recording Procedure. Running speech was presented at overall RMS levels of 55, 65, and 75 dB SPL. These levels approximate speech at casual through loud vocal effort levels [48]. Since individual phonemes naturally varied in their relative levels within the Rainbow passage, the level of each isolated phoneme was matched to the level at which it occurred in the Rainbow passage, for each presentation level. With this recording paradigm, the overall input levels of each phoneme were matched between the two contexts. During presentation of phonemes in the isolation context, approximately 10 repetitions of each phoneme (each preceded by ISI of 1125 ms) were presented during any single recording.

2.5. Output Measures. Measurements were carried out offline using SpectraPLUS. Two measurements were made per phoneme and per context: the overall level of the phoneme (dB SPL RMS recorded over the entire duration of the phoneme) and the onset level of the phoneme (dB SPL RMS recorded over the first 30 ms of the stimulus phoneme). Onset measurements could not be completed for phonemes /t/ and /g/ as the duration of these phonemes was shorter than 30 ms. For these phonemes, we therefore report only overall phoneme levels. In the isolation context, measurements were completed after the first few repetitions of the phoneme. The first few repetitions were discarded as, in our preliminary recordings using a few hearing aids, interrepetition variability was observed to be high in the first few repetitions. This is likely related to nonlinear signal processing in the hearing aids but these effects were not formally evaluated in this study. Figures 1(a) and 1(b) illustrate examples of the variability observed in the first few repetitions.

2.6. Analyses. Repeated measures of analysis of variance (RM-ANOVA) were completed using SPSS (v. 16) with context (running speech and isolation), level (55, 65, and 75 dB SPL), and phoneme as the three independent factors. Separate analyses were carried out for overall phoneme level and onset level. Greenhouse-Geisser corrected degrees of freedom were used for interpretation of all tests. Multiple paired t-tests were completed to explore significant context interactions. For interpretation of these multiple t-tests, sequential Bonferroni type corrections that control for false discovery rates were used to determine critical P values [49, 50].

Table 2: Frequency specific thresholds of N4 and N6 standard audiograms [39]. The threshold at 750 Hz for the N6 audiogram was originally 82.5 dB HL but had to be rounded to 85 dB HL to allow input into the verification system.

Audiogram	Frequency specific thresholds (dB HL)								
	250	500	750	1 kHz	1.5 kHz	2 kHz	3 kHz	4 kHz	6 kHz
N4	55	55	55	55	60	65	70	75	80
N6	75	80	85	85	90	90	95	100	100

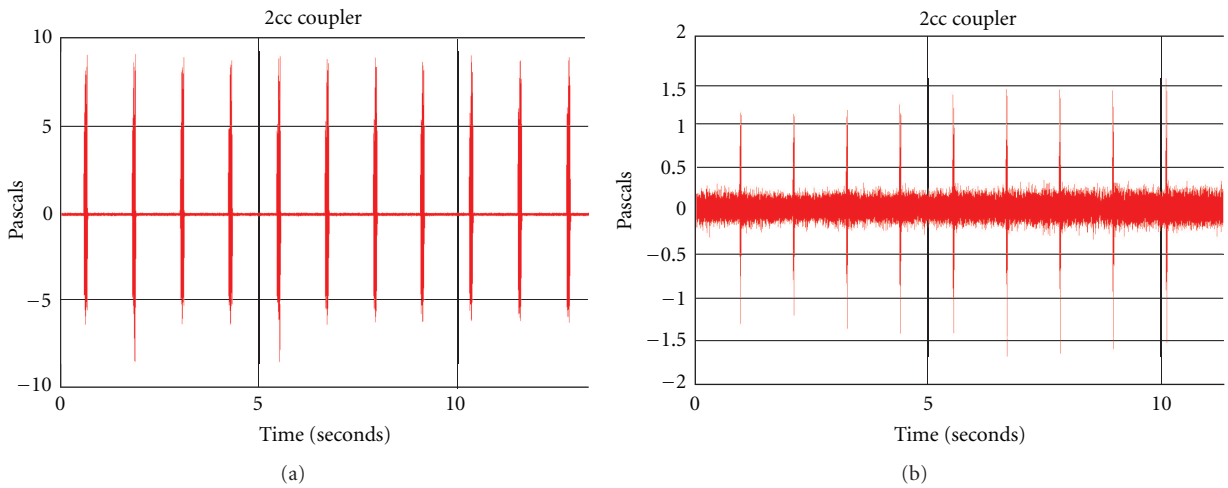

Figure 1: (a) illustrates the amplitude-time waveform of the output of one of the hearing aids when the stimulus /a/ was presented at 65 dB SPL. The hearing aid was programmed to DSL v5 targets derived for the audiogram N4. The first few repetitions are more variable than the later repetitions. (b) illustrates the amplitude-time waveform of the output of one of the hearing aids when the stimulus /g/ was presented at 55 dB SPL. The hearing aid was programmed to DSL v5 targets derived for the audiogram N4. The first few repetitions are lower in level compared to the later repetitions.

3. Results

Phonemes embedded in running speech were measurable for nearly all hearing aids in this study. For one of the hearing aids, the output level of /g/ in isolation at 55 dB SPL input level could not be measured as it was embedded within the hearing aid noise floor. Across the sample, the average overall phoneme level measured in the running speech context was 94.07 dB SPL (standard error (SE) = 1.79 dB) and in the isolation context was 92.43 dB SPL (SE = 1.94 dB). On average, the phoneme onset level measured in the running speech context was 94.67 dB SPL (SE = 1.79 dB) and in the isolation context was 94.44 dB SPL (SE = 1.83 dB). The outcome of statistical tests for overall phoneme level and phoneme onset level will be described below.

3.1. Difference in Overall Phoneme Level across Contexts. RM-ANOVA revealed a significant effect of context ($F = 10.114$ [1, 8], $P = 0.013$), input level ($F = 834.58$ [1.02, 8.12], $P < 0.001$), and phoneme ($F = 93.26$ [1.95, 15.62], $P < 0.001$). Interactions between input level and context ($F = 8.36$ [1.35, 10.82], $P = 0.011$), phoneme and context ($F = 3.38$ [2.63, 21.05], $P = 0.042$), and input level and phoneme ($F = 5.25$ [2.69, 21.56], $P = 0.009$) were also significant. The three-way interaction between input level, context, and phoneme was not significant ($F = 1.061$ [2.48, 29.79], $P = 0.388$). Paired contrasts comparing overall phoneme levels between contexts at each input level showed significant differences at the 55 and 65 dB SPL input levels but not at the 75 dB SPL input level. At input levels of 55 and 65 dB SPL, the levels of phonemes were significantly higher when they appeared in running speech compared to when they occurred in isolation (see Figure 2(a) and Table 3 for group means). In summary, the difference between contexts reduced as input level increased.

Paired contrasts comparing overall phoneme levels between contexts for each phoneme showed significant differences for all phonemes except /m/ (see Figure 2(b) and Table 4 for group means). All phonemes except /m/ were higher in level when they occurred in running speech compared to when they occurred in isolation.

3.2. Difference in Phoneme Onset Level across Contexts. A similar result was obtained for phoneme onset level. RM-ANOVA revealed a significant effect of context ($F = 7.41$ [1, 9], $P = 0.024$), input level (846.94 [1.05, 9.44], $P < 0.001$), and phoneme ($F = 52.84$ [1.78, 16.04], $P < 0.001$). Interactions between input level and context ($F = 17.71$ [1.20, 10.81], $P = 0.001$), and phoneme and context (3.95 [3.45, 31.09], $P = 0.013$) were significant. Interaction between input level and phoneme ($F = 1.49$ [2.06, 18.56], $P = 0.250$) and the three-way interaction between input level, context, and phoneme were not significant ($F = 0.89$ [3.25, 29.25], $P = 0.473$). Paired contrasts between phoneme

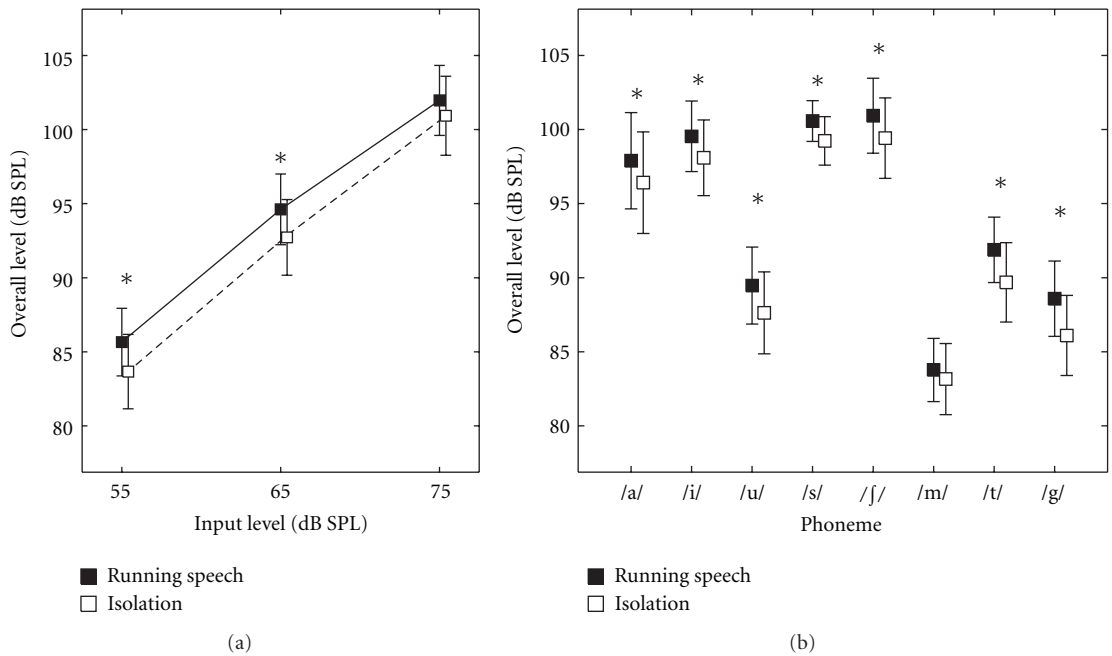

Figure 2: (a) presents variation of overall phoneme level in running speech and isolation context across input levels. (b) presents the same across phonemes. Error bars represent SE. ∗ indicates a statistically significant difference in paired contrasts. The symbols have been offset slightly to improve clarity.

Table 3: Results of post hoc tests for level context interaction.

	Input level	Running speech (mean (dB SPL), SE (dB))	Isolation (mean (dB SPL), SE (dB))	t-statistic, df	P value	Critical P value
Overall level	55	85.66, 2.28	83.66, 2.51	4.437, 9	0.002*	0.017
	65	94.61, 2.38	92.72, 2.55	3.803, 9	0.004*	0.033
	75	101.97, 2.36	100.9, 2.66	1.173, 9	0.121	0.050
Onset level	55	86.35, 2.27	85.96, 2.31	4.234, 9	0.002*	0.017
	65	95.18, 2.35	94.93, 2.38	2.739, 9	0.023*	0.033
	76	102.48, 2.41	102.43, 2.44	0.446, 9	0.653	0.050

*Indicates a statistically significant difference.

onset levels of both contexts at each input level showed significant differences between contexts at 55 and 65 dB SPL but not at the 75 dB SPL input level. At input levels of 55 and 65 dB SPL, the onset levels of phonemes were significantly higher when they appeared in running speech compared to when they occurred in isolation (see Figure 3(a) and Table 3 for group means). Similar to overall phoneme level, the difference between contexts reduced with increasing input level.

Paired contrasts comparing phoneme onset levels between contexts for each phoneme revealed no significant differences for all phonemes except /ʃ/ and /u/ (see Figure 3(b) and Table 4 for group means). Phonemes /ʃ/ and /u/ were higher in onset level when they occurred in running speech compared to when they occurred in isolation.

3.3. Individual Differences across Hearing Aids. The mean difference in overall phoneme level averaged across hearing aids, input levels, and phonemes was found to be 1.64 dB, where phonemes in running speech measured higher on average. The mean difference in phoneme onset level computed similarly was 0.23 dB, onset of phonemes in running speech measuring higher on average. Although the mean value suggests a clinically insignificant difference due to context, inspection of individual data highlights the differences observed across hearing aids and phonemes. Tables 5(a) and 5(b) provide the difference (in dB) in the output measures (overall phoneme level and phoneme onset level) in both contexts, averaged across all three input levels. These differences were obtained by subtracting the level of each phoneme in isolation from the corresponding level in running speech. Hence, a positive value indicates that the level of the phoneme is higher when it occurs in running speech, as it would in daily life, versus in isolation, as it would during CAEP measurement. Differences of greater than 3 dB are presented in bold.

The proportion of difference values greater than ±3 and ±5 dB are presented in Table 6 for both overall phoneme levels and phoneme onset levels at each input level. Pooled

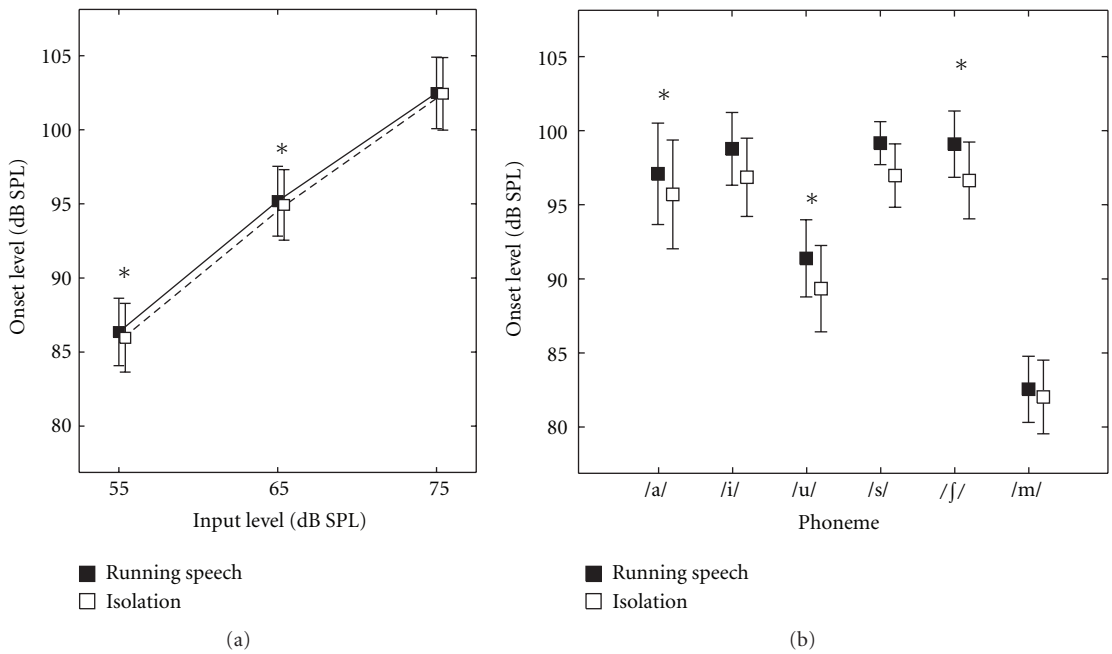

Figure 3: (a) presents variation of phoneme onset level in running speech and isolation context across input levels. (b) presents the same across phonemes. Error bars represent SE. ∗ indicates a statistically significant difference in paired contrasts. The symbols have been offset slightly to improve clarity.

Table 4: Results of post hoc tests for phoneme context interaction.

	Phoneme	Running speech (mean (dB SPL), SE (dB))	Isolation (mean (dB SPL), SE (dB))	t-statistic, df	P value	Critical P value
Overall level	/a/	97.89, 3.25	96.41, 3.43	3.197, 9	0.011*	0.025
	/i/	99.54, 2.38	98.09, 2.55	2.856, 9	0.019*	0.031
	/u/	89.46, 2.59	87.62, 2.77	4.231, 9	0.002*	0.006
	/s/	100.57, 1.37	99.23, 1.64	3.506, 9	0.007*	0.019
	/∫/	100.93, 2.52	99.42, 2.71	3.981, 9	0.003*	0.013
	/m/	83.77, 2.13	83.16, 2.39	0.954, 9	0.365	0.050
	/t/	91.88, 2.20	89.69, 2.67	2.425, 9	0.038*	0.044
	/g/	88.57, 2.54	88.1, 2.70	2.450, 9	0.037*	0.038
Onset level	/a/	97.09, 3.41	95.69, 3.67	2.376, 9	0.042	0.042
	/i/	98.77, 2.45	96.85, 2.64	2.588, 9	0.029	0.025
	/u/	91.37, 2.60	89.34, 2.91	3.143, 9	0.012*	0.017
	/s/	99.16, 1.44	96.97, 2.14	2.497, 9	0.034	0.033
	/∫/	99.08, 2.24	96.63, 2.59	4.161, 9	0.002*	0.008
	/m/	82.54, 2.23	82.07, 2.48	0.634, 9	0.542	0.050

*Indicates a statistically significant difference.

across both directions of differences and input levels, about 24% of the overall phoneme levels (total of 239 observations across three levels, 10 hearing aids and eight phonemes, 1 missing value) showed differences of greater than ±3 dB and 7% showed differences of greater than ±5 dB. In case of phoneme onset levels, about 33% of the observations (total of 180 observations across three levels, 10 hearing aids and six phonemes) showed differences of over ±3 dB and nearly 13% showed differences of over ±5 dB. In general, differences greater than 3 dB are well outside of test-retest differences in electroacoustic measurement, while differences greater than 5 dB are greater than a typical audiometric step size. The latter is likely clinically significant, while the former may have impact for interpretation of research data and calibration. We note that the majority of aided phoneme levels agreed between the two contexts within ±3 dB.

Table 5: (a) Difference (dB) in overall phoneme level averaged across input levels (positive value indicates higher overall phoneme level in running speech). (b) Difference (dB) in phoneme onset level averaged across input levels (positive value indicates higher phoneme onset level in running speech).

(a)

Hearing aid	Phoneme							
	/a/	/i/	/u/	/s/	/∫/	/m/	/t/	/g/
1	0.68	0.56	**3.88**	2.66	**3.47**	1.58	2.45	2.25
2	**3.10**	2.53	**3.40**	1.16	2.16	**3.27**	**6.54**	**6.03**
3	2.46	**3.58**	**3.45**	2.89	2.29	2.20	**5.15**	**6.23**
4	**4.56**	**4.72**	1.84	**3.22**	**3.24**	**3.44**	**5.23**	**4.69**
5	−0.31	−0.01	−0.06	0.85	0.29	−2.20	−1.28	−0.46
6	1.19	0.55	0.75	0.40	0.61	0.44	2.60	1.50
7	0.62	0.70	1.73	0.60	0.59	−1.09	−1.05	0.77
8	0.62	0.92	1.46	1.55	1.34	0.31	**3.00**	**5.22**
9	0.70	0.25	1.85	0.53	0.42	0.34	−0.12	**3.66**
10	1.15	0.65	0.13	−0.43	0.67	−2.16	−0.61	−0.39

(b)

Hearing aid	Phoneme					
	/a/	/i/	/u/	/s/	/∫/	/m/
1	−0.39	0.01	2.53	2.23	**3.64**	0.52
2	2.44	2.81	**3.69**	1.81	**3.74**	**3.95**
3	**4.11**	**6.53**	**6.81**	**7.06**	**5.69**	**3.04**
4	**4.70**	**5.37**	2.28	**7.48**	**4.45**	**4.31**
5	−0.47	−0.71	−0.43	0.71	1.37	−2.59
6	1.72	0.70	1.05	0.05	1.26	−0.38
7	0.12	0.84	0.99	0.29	0.66	−2.23
8	−0.17	1.78	2.06	1.27	2.67	0.81
9	0.55	0.51	1.25	0.34	0.82	0.38
10	1.33	1.36	0.19	0.61	0.19	−2.61

4. Discussion

Results suggest that hearing aid output level of a phoneme in isolation may either match or may differ from the output level of the same phoneme when it occurs in running speech. Agreement was observed in approximately 66% to 75% of cases, while differences exceeding 3 dB were observed in 24% to 33% of cases. Agreement occurred in more cases (75%) for measures of overall level of phoneme, and in fewer cases (66%) for measures of phoneme onset level. When differences existed, they typically manifested as the hearing aid producing a lower output for the phoneme in isolation than it did for the phoneme in running speech. Differences reduced with increases in input level and varied across phonemes and hearing aids. Similar trends were observed in overall phoneme level and phoneme onset level.

Results from the present study are similar to the findings from other reports in this issue [40, 41]. Specifically, these reports and the current study show that across measurement strategies and stimulus types, hearing aids may apply lower gain and output (at onset as well as at maximum amplitude) to brief stimuli that are immediately preceded by silence, such as those commonly used to elicit the CAEP. However, one may note that the hearing aids used in these studies [40, 41] were set to function linearly, unlike the hearing aids used in the present study. Another study has used a nonlinear hearing aid to study the effect of hearing aid processing on the tone burst onset while comparing it with the unaided condition [36]. The aided condition in this study produced a marginal increase in the level at onset due to the presence of an overshoot. In the present study, there were fewer instances of significant overshoot, but recall that the unaided condition was not assessed in this study. Therefore, the present results pertain only to the comparison of aided levels between the isolation context and running speech. Overshoot may be present in both conditions. Also, the effects of overshoot attributable to nonlinear signal processing in hearing aids may vary across devices, with the effects being idiosyncratic to specific devices or stimuli. Results similar to the majority of the observations in the present study have also been noted in non-CAEP related studies of nonlinear signal processing in hearing aids [37, 38].

Table 6: Proportion of observations (%) showing differences greater than 3 or 5 dB (positive value indicates higher output levels of phoneme in running speech).

	Input level	>3 dB	< −3 dB	>5 dB	< −5 dB
Overall level	55	9.62	—	—	2.93
	65	9.21	—	—	2.09
	75	4.60	0.84	—	2.51
Onset level	55	12.78	—	7.22	—
	65	12.22	1.11	3.33	—
	75	6.11	1.11	1.67	0.56

4.1. Effect of Input Level and Phoneme on Difference due to Context. The decrease in differences in overall and onset level of phonemes between contexts with increase in input level could indicate an effect of output limiting. As the output levels of phonemes come close to the maximum power output of the hearing aids, they are subject to compression limiting [1, 5]. Compression limiting restricts the maximum output level by using a very high or infinite compression ratio in an output controlled compression system [1]. Hence, at higher input levels, where the output levels are likely subject to output limiting in both stimulus contexts, the differences seen are smaller compared to lower input levels that are relatively less likely to be affected by output limiting.

Analyses revealed that differences across contexts varied across phonemes. We did not perform a direct comparison across phonemes because the individual phonemes occur at different levels relative to, the overall RMS level of running speech. Compression, being a level-dependent nonlinear factor in the hearing aid, may therefore vary the gain applied for each of these phonemes, especially when they are presented in isolation. In addition, compression features such as compression ratio and time constants were likely different across different frequencies due to the slightly sloping configurations of audiograms chosen and the presence of multiple channels in our hearing aid sample.

Since phonemes varied in their spectral composition and position of spectral peaks, they could have been subject to different compression features in different channels. One stimulus characteristic that could have been influential in determining overall phoneme output levels is the duration of phonemes. Table 5(a) suggests that differences larger than 3 dB occurred more often for /g/ and /t/ relative to other phonemes. Among all eight phonemes, /t/ and /g/ were the lowest in level and shortest in duration, measuring 26 ms and 19 ms, respectively. This may have made these phonemes in isolation more susceptible to the dynamic effects of hearing aid nonlinearity [1, 37, 38]. However, this study did not study systematically the effects of duration and level as they interact with context. Further study on this may be necessary to determine the effects of phoneme level and duration. Also, the preceding context within running speech may have differed in ways crucial to determination of gain/compression characteristics for the target phoneme.

4.2. Interhearing Aid Variability. Tables 5(a) and 5(b) illustrate that individual hearing aids may amplify individual phonemes differently, even though they were set to produce similar gain for long-duration signals. Hearing aids not only varied in differences due to context but also showed differences for the same phoneme in the same context. This illustrates that different manufacturers may employ different nonlinear signal processing strategies. Differences across hearing aid manufacturers were also reported by Jenstad et al. [40]. Differences in other parameters across hearing aid manufacturers have also been reported among hearing aids that were matched in gain characteristics (e.g., sound quality comparisons by Dillon et al. [51]). The finding that hearing aids show large individual variability makes it challenging to predict the nature of differences on a case-by-case basis in clinical practice.

4.3. Implications for Aided CAEP Testing. CAEPs are level dependent [26, 46, 52, 53]. Parameters such as amplitude and latency of individual peaks reflect changes in stimulus level or sensation level of the stimulus with reference to the behavioral threshold of the CAEP stimulus. A change in sensation level of the stimulus from a positive (above threshold; audible) to a negative (below threshold; inaudible) value is likely to decrease the probability of eliciting a CAEP. If output levels of phonemes in running speech are considered to be the reference condition of interest, CAEP test measures may underestimate audibility when phonemes are presented in isolation. These data indicate that underestimation is minimal (about 2 dB) on average, but was between 3 and 8 dB in over 24% of cases. There were also instances that may result in overestimation of audibility, but these are far fewer in number and magnitude.

Since the experimental conditions used in this study were limited to one duration of ISI and one naturally occurring preceding context per phoneme, generalization to other instances and variation across durations or levels of phonemes may require further investigation. Investigation of the effects of hearing aid signal processing on spectral characteristics such as formant transitions may also be possible, but these effects were not evaluated in this study. The effects of other aspects of hearing aid signal processing, such as digital noise reduction, may also be relevant and were not explored in this study. Based on this study, we conclude that significant differences in hearing aid functioning between running speech and isolated phoneme contexts occur, along with considerable interhearing aid variability. In over a fourth of aided phonemes, the magnitude of these differences was large enough to impact calibration, or interpretation of group data. This may indicate the need to perform acoustic calibration for individual hearing aids for the purpose of well-defined CAEP stimuli. In 7%–13% of phonemes, the differences exceeded that of an audiometric step size and therefore may be clinically important.

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Research Article

Slow Cortical Potentials and Amplification—Part II: Acoustic Measures

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In a previous study, we investigated slow cortical potential (SCP) N1-P2 amplitudes and N1 latencies in aided and unaided conditions, with the finding that despite being set to provide 20 or 40 dB of gain, none of the hearing aids resulted in a reliable increase in SCP response amplitude relative to the unaided (Marynewich et al., in press). The current study investigates the effects of hearing-aid processing on acoustic measures for two 1000-Hz tonal stimuli: short (60 ms) and long (757 ms), presented at three intensities (30, 50, 70 dB SPL) in aided and unaided conditions using three hearing aids (Analog, DigitalA, DigitalB) with two gain settings (20, 40 dB). Acoustic results indicate that gain achieved by the hearing aids, measured at 30 ms after stimulus onset, for both the short and long stimuli, was less than real-ear insertion gain measured with standard hearing aid test signals. Additionally, the digital hearing aids altered the rise time of the stimuli such that maximum gain was reached well past 30 ms after stimulus onset; rise times differed between the digital aids. These results indicate that aided SCP results must be cautiously interpreted and that further research is required for clinical application.

1. Introduction

Slow cortical potentials (SCPs) are being considered for their application in hearing aid fitting, particularly for infants [1–7]. SCP studies related to this purpose have produced mixed results; thus it is not known whether SCPs provide an accurate measure of the brain's response to (and, hopefully, behavioural perception of) 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 [2, 4–10], and some have reported that stimuli can be reliably differentiated in the SCP response in aided conditions [5, 6]. In contrast, there is a puzzling finding that the provision of gain via a hearing aid does not lead to the expected increase in SCP response amplitude for either tonal or speech stimuli [2, 6, 11]. To understand why SCP amplitude does not increase with hearing aid gain, it is important to quantify the acoustic effects of 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 earlier studies which found that due to hearing-aid processing, stimuli used to measure some auditory-evoked potentials (AEP) may not result in valid measurements of hearing aid gain or output [12, 13].

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 both hearing-aid processing and AEPs. For instance, complex stimuli such as speech-weighted composite noise provide better estimates of gain for speech compared with tonal stimuli [14–16], whereas tonal stimuli provide better measures of maximum power output (MPO) than do complex stimuli [15]. 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 [12, 13, 17]. 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 [12, 13].

The N1 response does not reflect stimulus changes beyond the first 20–40 ms [18–22], and rise times between 20 and 30 ms result in the largest N1 amplitudes [20, 21]. The tonal stimulus used by Billings et al. [2] was atypical for SCP stimuli in that it had a more rapid rise time (7.5 ms) than is required to elicit a large-amplitude SCP and maintain reasonable stimulus frequency specificity; their stimulus was also much longer in duration than can be reflected by the SCP. The possibility that stimulus characteristics were the reason for the inability to measure hearing aid gain via SCPs was addressed by Marynewich et al. [11], who compared N1-P2 amplitudes and N1 latencies in unaided and aided conditions in normal-hearing listeners, using a stimulus designed to elicit larger N1 amplitudes with less compromise of frequency specificity; that is, a 60-ms duration tonal stimulus with a 20-ms rise time [20, 21, 23]. The results of Marynewich et al. [11] were similar to those of Billings et al. [2] in that the SCP amplitude did not increase as expected despite the provision of 20 or 40 dB of gain.

What is not clear from many of the SCP studies is what effect hearing-aid processing had on the stimuli. Billings et al. [24] have since examined the effect of signal-to-noise ratio (SNR) and showed that the SNR may have been similar across aided and unaided conditions in their previous research [2, 6], which may explain why N1 amplitudes were not larger in the aided conditions compared with unaided.

The purpose of this study was to measure, in the ear canals of subjects, tonal stimuli used for SCP testing before and after hearing-aid processing to determine how hearing-aid processing affected the stimuli, measured under the same conditions as the SCP testing. 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 on gain, particularly within the first 30 ms after stimulus onset, even with all advanced features disabled.

2. Materials and Methods

2.1. Subjects. Five subjects participated (mean age: 23 ± 2.1 years; 4 females). Four of these subjects also participated in Marynewich et al. [11]. Subjects were briefed on the study procedures and provided informed written consent prior to participating. All subjects were screened for normal middle/outer-ear function by immittance audiometry. Normal tympanograms were defined by a single-peak static admittance between ±50 daPa in response to a 226-Hz probe tone [25]. Although hearing status would not be expected to influence the results of the acoustic recordings, all subjects had normal hearing.

2.2. Hearing Aids. The same three behind-the-ear hearing aids, coupled with Comply snap tip 9-mm foam earmolds, were used for each participant: (i) Oticon E27 ("Analog"), (ii) Phonak Savia 211 dSZ ("DigitalA"), and (iii) Siemens Acuris S ("DigitalB"). These were the same hearing aids and settings used in our previous study [11].

The digital hearing aids were set, using NOAH 3 and the NOAHLink, to have two programs: one with 20 dB and the second with 40 dB real-ear insertion gain (REIG) for a 50 dB SPL 1000 Hz pure tone, as verified with the Fonix 7000 real-ear system. 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. Other frequencies were set to minimum gain. Settings for the digital instruments were saved in the NOAH 3 software for each subject and recalled in follow-up sessions.

Gain settings for the analog hearing aid were achieved by setting the volume control to 1 (minimum) and turning the dB SPL trim-pot until the REIG was 20 dB for a 1000-Hz pure tone at a 50 dB SPL input level. To achieve the 40-dB gain setting, the volume control wheel was turned up until REIG equalled 40 dB at 1000 Hz. The volume control wheel was then marked for that setting. These gain settings were remeasured in follow-up sessions.

REIG measures at 1000 Hz for all subjects and hearing aids are given in Table 1.

2.3. Stimuli. Two 1000-Hz "SCP" stimuli were used for the acoustic measures: (i) a stimulus of 60-ms total duration (including a 20-ms rise/fall time), the same as used by Marynewich et al. [11], and (ii) a stimulus of 757-ms duration (with a 7.57-ms rise/fall time), similar to the one used by Billings et al. [2]. Stimuli were presented with offset-to-onset interstimulus intervals (ISI) of 940 ms. Stimuli generated by Neuroscan's Stim2 software were further amplified by a Wavetek Rockland 852 filter (providing 20 dB of amplification below 3000 Hz), 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).

2.4. Procedure. Subjects were asked to complete two test sessions, 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 outer- and middle-ear function at each 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 octave-band noise levels in the sound-attenuated booth at 0.5, 1, 2, and 4 kHz were 12, 10, 10, and 12 dB SPL, respectively. There were 36 test conditions (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(s). During testing, participants were asked to sit as still as

Table 1: Real-ear insertion gain (REIG) (dB) measured with a hearing aid test system for 20 and 40 dB gain settings.

Gain setting (dB)	20		40
Input level (dB SPL)	50	70	50
Subject		Analog	
1	20.40	20.70	39.90
2	19.90	20.00	39.80
3	20.00	19.70	39.40
4	20.00	20.70	39.70
14	20.50	19.30	39.90
Mean	20.16	20.08	39.74
SD	0.27	0.62	0.21
Subject		Digital A	
1	20.30	19.80	40.40
2	20.30	20.10	39.60
3	19.50	19.70	39.90
4	20.10	20.30	40.30
14	20.10	19.40	40.40
Mean	20.06	19.86	40.12
SD	0.33	0.35	0.36
Subject		Digital B	
1	20.10	20.00	40.10
2	20.10	20.60	40.10
3	19.80	19.70	40.40
4	20.00	20.10	40.10
14	20.00	20.20	39.90
Mean	20.00	20.12	40.12
SD	0.12	0.33	0.18

possible while watching a movie of their choice in closed-captioning and no audio. Subjects sat in a reclining chair set in the upright position so that each participant was seated with their head above the chair back, the same position used for the previous SCP measurements.

2.4.1. Recording. An ER7C probe-tube output (set to provide 20 dB of attenuation) was routed through a second (passive) 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) by the Neuroscan Synamps[2] system and averaged and analyzed by the Neuroscan Scan analysis system, using a 204.75-ms analysis time for the short stimulus (including a 70-ms prestimulus baseline) and a 960-ms analysis time for the long stimulus (including a 100-ms prestimulus 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, which included baseline correction across the stimulus duration and averaging of the single trials.

2.5. Data Analysis. Acoustic measures of interest were (i) gain at 30 ms after stimulus onset, (ii) maximum gain, and (iii) latency of maximum gain, or "rise time" (defined as the time at which the amplitude first reached 90% of maximum amplitude relative to an individually determined 0-ms point). Actual 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 after stimulus onset was chosen because several studies indicate this is the most effective rise time and evokes the largest N1-P2 amplitudes; increases in stimulus levels beyond 20–40 ms have little to no effect on SCP amplitudes [18–22]. Maximum gain was calculated to determine the maximum gain produced at any time during the stimulus, even if this occurred past 30 ms after stimulus onset. Rise time was measured in order to determine whether hearing-aid processing resulted in stimulus rise times longer than 30 ms. The stimulus onsets (0-ms points) were determined for each waveform by a research assistant blind to the study purpose. Using the same zoom settings to visually inspect each waveform, the research assistant identified the time point at which there was periodicity in the recording. A random subset of the waveforms were retested to determine test-retest reliability of the 0-point identification protocol; the average of the absolute values of the errors was less than 1 ms (0.24 ms).

2.6. Statistical Analysis. For the short-duration (60-ms) stimulus, two repeated-measures analyses of variance (ANOVA) were conducted for each of the dependent variables: gain measured at 30 ms, maximum gain, and rise time: (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 this study, main effects and interactions for all analyses were considered significant if $p < .10$. Huyn-Feldt correction factors were applied to the degrees of freedom and reported where appropriate (i.e., when the assumption of sphericity was not met). Significant interactions were examined by analyzing the simple main effects, then conducting paired t-tests for any significant simple main effects. Neuman-Keuls post hoc analyses were performed for significant main effects not involved in an interaction. Post hoc analyses were considered statistically significant if $p < .10$.

3. Results

The following section is divided into: (i) gain and rise time results for the short stimulus and (ii) gain and rise time results for the long stimulus. Mean data for gain measured at 30 ms and maximum amplitude, along with the rise times for maximum amplitude, are provided in Table 2. ANOVA

Table 2: Mean ($n = 5$) and standard deviation for gain measured at (i) 30 ms and (ii) maximum amplitude, and rise time (measured at 90% of maximum amplitude) for the cortical stimuli.

Nominal gain (dB)	Short stimulus (60 ms)					Long stimulus (757 ms)				
	20			40		20			40	
Input (dB SPL)	30	50	70	30	50	30	50	70	30	50
Analog										
30 ms (dB)	15.9 ± 1.6	17.0 ± 0.9	17.3 ± 1.0	34.9 ± 2.2	36.2 ± 1.8	16.0 ± 0.7	16.7 ± 0.6	17.4 ± 0.7	35.5 ± 0.5	35.5 ± 1.4
Max (dB)	15.7 ± 1.1	16.9 ± 1.0	17.4 ± 0.1	34.5 ± 1.9	36.2 ± 2.0	15.2 ± 1.3	16.6 ± 0.7	17.3 ± 0.8	34.7 ± 1.9	35.3 ± 1.5
Rise time (ms)	22.2 ± 1.9	21.9 ± 1.4	22.4 ± 1.7	22.5 ± 1.4	22.1 ± 1.9	12.5 ± 3.2	11.6 ± 3.2	12.8 ± 2.8	11.9 ± 1.9	12.3 ± 2.6
DigitalA										
30 ms (dB)	10.9 ± 4.5	15.8 ± 3.1	15.6 ± 3.1	30.1 ± 6.1	35.5 ± 3.5	13.1 ± 3.3	17.0 ± 2.0	16.8 ± 2.4	33.8 ± 3.2	36.1 ± 1.5
Max (dB)	15.4 ± 3.2	17.1 ± 2.7	16.7 ± 2.6	34.5 ± 4.3	36.3 ± 2.6	15.7 ± 2.4	17.8 ± 1.6	17.1 ± 2.6	36.9 ± 1.9	36.4 ± 1.3
Rise time (ms)	38.6 ± 3.7	27.9 ± 3.3	30.3 ± 3.1	33.8 ± 6.6	22.6 ± 2.3	39.4 ± 7.2	25.7 ± 7.4	23.4 ± 6.6	43.3 ± 14.7	18.8 ± 5.5
DigitalB										
30 ms (dB)	−2.7 ± 2.1	0.9 ± 2.9	0.2 ± 2.3	21.5 ± 3.4	23.1 ± 2.3	−2.0 ± 3.2	1.4 ± 5.3	4.2 ± 3.9	23.1 ± 2.2	26.1 ± 2.6
Max (dB)	4.5 ± 3.8	7.6 ± 2.3	7.5 ± 1.1	26.2 ± 2.2	27.8 ± 1.6	13.8 ± 4.1	15.8 ± 3.1	17.4 ± 2.2	35.4 ± 1.8	37.0 ± 1.8
Rise time (ms)	45.5 ± 2.7	44.2 ± 1.8	45.2 ± 1.5	41.6 ± 2.4	40.9 ± 2.5	141.5 ± 8.9	150.8 ± 4.5	149.6 ± 8.4	154.5 ± 3.3	145.6 ± 3.2

results are reported in Tables 3 and 4, along with the results for Simple Main Effects when an interaction was significant.

3.1. Short Stimulus. 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 1, 2, and 3, respectively. Both 20- and 40-dB gain settings are depicted where appropriate (e.g., 30 and 50 dB SPL input levels) 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 no. 2). Note that for optimum visual representation, the scale is different across stimulus waveform figures.

3.1.1. 20-dB Gain Condition: Short Stimulus—Gain at 30 ms after Stimulus Onset. Mean gain values presented in Table 2 indicate that at the 30-ms measurement point for the short stimulus, all hearing aids provided less than 20 dB gain. The analog hearing aid was 3-4 dB below the nominal gain across input levels; DigitalA was 4–9 dB below nominal gain, and DigitalB provided no measurable gain at 30 ms, and even attenuation of up to 3 dB.

Effect of Input Level: results from the ANOVA and post hoc analysis revealed that even though all three hearing aids were set to provide linear amplification, less gain was measured for the 30 dB SPL compared with the 50 dB SPL input level for all three hearing aids ($p < .1$) and the 30 dB SPL compared with the 70 dB SPL input level for all three hearing aids ($p < .1$). Gain was not significantly different between the 50 and 70 dB SPL input levels for any of the hearing aids ($p > .1$).

Effect of Hearing Aid: the DigitalB hearing aid provided significantly less gain than both the DigitalA and Analog hearing aids at every input level ($p < .01$). The DigitalA hearing aid provided significantly less gain than the Analog hearing aid for the 30 dB SPL input ($p < .05$). There was no significant difference between gain provided by Analog and DigitalA aids for the higher input levels, 50 and 70 dB SPL ($p > .1$).

3.1.2. 20-dB Gain Condition: Short Stimulus—Gain at Maximum Amplitude. Mean gain values presented in Table 2 show that, when measured at the maximum amplitude measurement point, once again, all of the hearing aids provided less than 20 dB gain. The Analog aid was again about 3-4 dB below nominal gain, similar to the levels measured at 30 ms. The DigitalA aid had more gain at maximum amplitude than at 30 ms but was still 3-4 dB less than nominal gain. The DigitalB aid, once again, provided less gain than Analog and DigitalA hearing aids, but unlike the 30-ms measurement point, there was some measurable gain, albeit 12–15 dB less than nominal gain. There was no significant interaction between Input Level and Hearing Aid, so the results reported here are the analyses of the Main Effects.

Effect of Input Level: the 30 dB SPL input level resulted in significantly less gain than the 50 and 70 dB SPL input levels.

Effect of Hearing Aid: 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.

3.1.3. 20-dB Gain Condition: Short Stimulus—Rise Time. The rise time was the time taken to reach maximum amplitude (or, more precisely, the time to reach 90% of maximum amplitude). There was a clear trend for the two digital aids to take longer to reach maximum amplitude than the Analog aid. Rise time for the Analog aid was about 22 ms, similar to the 20-ms rise time of the input stimulus. Rise times for DigitalA ranged from 28 to 39 ms, and DigitalB showed a markedly longer rise times than DigitalA, approximately 45 ms.

Effect of Input Level: for the Analog and DigitalB hearing aids, rise time did not differ by input level ($p > .1$). For DigitalA, measured rise time differed by input level. The measured rise time was shortest for the 50 dB SPL input level (50 versus 70: $p < .05$), longer for the 70 dB SPL input level, and longer still for the 30 dB SPL input level (30 versus 70: $p < .05$).

Effect of Hearing Aid: at every input level, the rise time differed across hearing aids ($p < .01$), with the rise time being longest for the DigitalB hearing aid, shorter for the DigitalA hearing aid, and shortest for the Analog hearing aid.

3.1.4. 40-dB Gain Condition: Short Stimulus—Gain at 30-ms after Stimulus Onset. Mean gain values in Table 2 indicate that at the 30 ms measurement point, all hearing aid provided less than 40 dB gain. The Analog hearing aid provided 3–5 dB less than nominal gain, which was similar to the 20-dB gain condition. DigitalA provided 5–10 dB less than nominal gain, and DigitalB provided almost 20 dB less than the nominal 40 dB gain.

Effect of Input Level: unlike the 20-dB gain condition, where less gain was measured for the 30 dB SPL input level compared to the higher input levels, for the 40-dB gain condition, only DigitalA was measured to have less gain for the 30 than 50 dB input levels ($p < .1$). The Analog and DigitalB aids were measured to have the same amount of gain for both 30 and 50 dB SPL input levels ($p > .1$).

Effect of Hearing Aid: similar to the 20-dB gain condition, in the 40-dB gain condition, the DigitalB hearing aid provided significantly less gain than the DigitalA and Analog hearing aids at both input levels ($p < .1$). There was a nonsignificant trend for the DigitalA to provide less gain than the Analog hearing aid for the 30 dB SPL input ($p = .11$). The DigitalA and Analog hearing aids provided equivalent gain for the 50 dB SPL input ($p > .1$).

3.1.5. 40-dB Gain Condition: Short Stimulus—Gain at Maximum Amplitude. Mean gain values follow much the same pattern at maximum amplitude for the short stimulus as at 30 ms. Once again, all of the hearing aids provided less than 40 dB gain: Analog was again 3–5 dB below nominal gain, DigitalA was 4-5 dB below nominal gain, and DigitalB provided much less gain than both Analog and DigitalA hearing aids at about 12–14 dB below nominal gain. There was not a

Table 3: ANOVA results for all measures of the short stimulus. Shown are p values of the ANOVA and of the simple main effects, where appropriate.

	20 dB gain			40 dB gain		
	Amp 30 ms	Amp Max	Rise time	Amp 30 ms	Amp Max	Rise time
Main effects						
Hearing aid (HA)	**$p < .001$**	**$p < .001$**	**$p < .001$**	**$p < .001$**	**$p < .001$**	**$p < .001$**
Input level (LVL)	**$p = .001$**	**$p = .054$**	**$p < .001$**	**$p = .045$**	**$p = .087$**	**$p = .004$**
HA × LVL	**$p = .03$**	$p = .11$	**$p < .001$**	**$p = .036$**	$p = .99$	**$p < .001$**
Simple main effects						
Effect of LVL						
For analog	**$p = .018$**	∗	$p = .744$	$p = .309$	∗	$p = .484$
For DigitalA	**$p = .004$**	∗	**$p = .001$**	**$p = .028$**	∗	**$p = .007$**
For DigitalB	**$p = .048$**	∗	$p = .121$	$p = .194$	∗	**$p = .045$**
Effect of HA						
At 30 dB SPL	**$p < .001$**	∗	**$p < .001$**	**$p = .001$**	∗	**$p < .001$**
At 50 dB SPL	**$p < .001$**	∗	**$p < .001$**	**$p = .001$**	∗	**$p < .001$**
At 70 dB SPL	**$p < .001$**	∗	**$p < .001$**	—	—	—

Boldface indicates significance at $p < .1$. ∗: Indicates the analysis was not necessary. —: Indicates no data collected for those conditions.

Table 4: ANOVA results for all measures of the long stimulus. Shown are p values of the ANOVA and of the simple main effects, where appropriate.

	20 dB gain			40 dB gain		
	Amp 30 ms	Amp max	Rise time	Amp 30 ms	Amp max	Rise time
Main effects						
Hearing aid (HA)	**$p < 0.001$**	$p = 0.656$	**$p < 0.001$**	**$p < 0.001$**	$p = 0.314$	**$p < 0.001$**
Input level (LVL)	**$p < 0.001$**	**$p = 0.005$**	$p = 0.424$	**$p = 0.027$**	$p = 0.298$	**$p = 0.016$**
HA × LVL	**$p = 0.031$**	**$p = 0.09$**	**$p < 0.001$**	**$p = 0.091$**	$p = 0.107$	**$p = 0.029$**
Simple main effects						
Effect of LVL						
For analog	**$p = 0.023$**	**$p = 0.009$**	$p = 0.647$	$p = 0.956$		$p = 0.429$
For digitalA	**$p = 0.006$**	**$p = 0.067$**	**$p = 0.02$**	$p = 0.134$		**$p = 0.030$**
For digitalB	**$p = 0.002$**	**$p = 0.003$**	**$p = 0.07$**	**$p = 0.015$**		**$p = 0.007$**
Effect of HA						
At 30 dB SPL	**$p < 0.001$**	$p = 0.457$	**$p < 0.001$**	**$p < 0.001$**		**$p < 0.001$**
At 50 dB SPL	**$p < 0.001$**	$p = 0.377$	**$p < 0.001$**	**$p < 0.001$**		**$p < 0.001$**
At 70 dB SPL	**$p < 0.001$**	$p = 0.97$	**$p < 0.001$**			

Boldface indicates significance at $p < .1$. ∗: Indicates the analysis was not necessary. —: indicates no data collected for those conditions.

significant interaction between Input Level and Hearing Aid, so the results reported here are the analyses of the significant Main Effects.

Effect of Input Level: the 30 dB SPL input level resulted in significantly less gain than the 50 dB SPL input level.

Effect of Hearing Aid: DigitalB hearing aid provided significantly less gain than either the Analog or DigitalA hearing aids ($p < .01$), and there was no significant difference between the gain provided by Analog and DigitalA ($p > .1$).

3.1.6. 40-dB Gain Condition: Short Stimulus—Rise Time. Again, there was a clear trend for the two digital aids to take longer to reach maximum amplitude than the analog aid, and the DigitalB aid took longer than DigitalA. The analog aid again mimicked the rise time of the input signal, with a measured rise time of about 22 ms. The DigitalA had a longer rise time, ranging from 22 to 34 ms. Again, DigitalB had a markedly longer rise time, taking about 41 ms to reach maximum amplitude.

Effect of Input Level: for the Analog aid, rise time did not differ between the two input levels ($p > .1$). For both the DigitalA and DigitalB hearing aids, measured rise time was longer for 30 dB SPL than the 50 dB SPL input level ($p < .1$).

Effect of Hearing Aid: at the 30 SPL input level, DigitalB had a longer rise time than both the DigitalA ($p < .05$) and Analog ($p < .001$) hearing aids. DigitalA had a longer rise time than Analog ($p < .05$). At the 50 dB SPL input level, DigitalB still had a longer rise time than both DigitalA ($p < .001$) and Analog ($p < .001$), but DigitalA and Analog had equivalent rise times ($p > .1$).

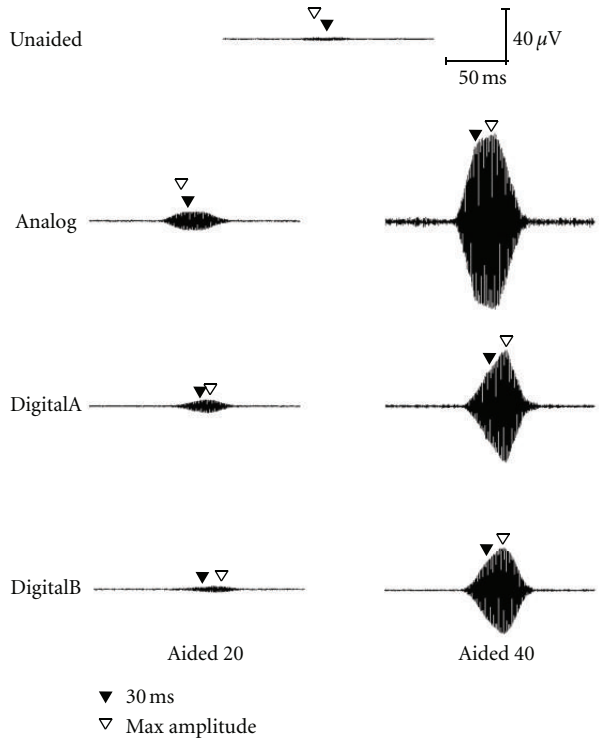


Figure 1: Waveform of the short stimulus presented at 30 dB SPL as measured in the ear canal of a single representative subject (subject 2), where "Aided 20" indicates the 20-dB gain condition and "Aided 40" indicates the 40-dB gain condition. The closed triangle indicates the point 30 ms after stimulus onset and the open triangle indicates the point at which maximum amplitude was reached.

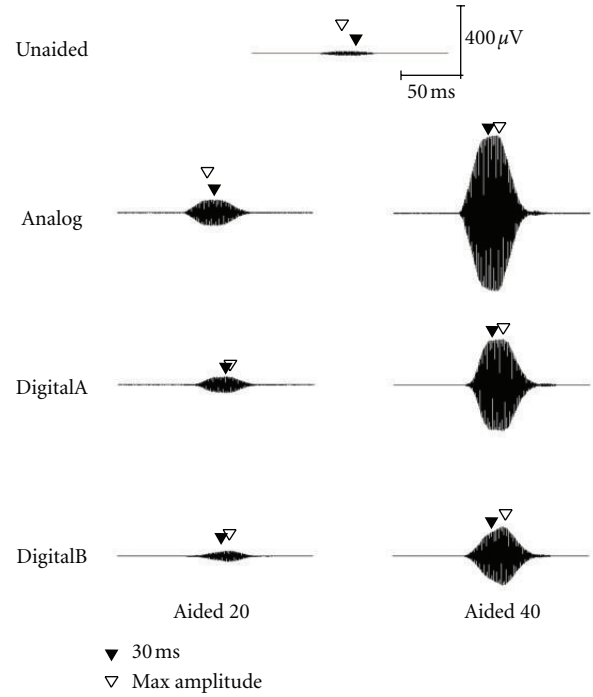


Figure 2: Waveform of the short stimulus presented at 50 dB SPL as measured in the ear canal of a single representative subject (subject 2).

3.2. Long Stimulus. 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 4, 5, and 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. Once again, all figures in the following section illustrate the acoustic measures for a single representative subject (subject no. 2).

3.2.1. 20-dB Gain Condition: Long Stimulus—Gain at 30 ms after Stimulus Onset. Mean gain values in Table 2 indicate that at the 30-ms measurement point for the long stimulus, once again all of the hearing aids provided less than 20 dB gain, in a pattern similar to that found for the short stimulus. The analog aid was 3-4 dB below nominal gain across input levels; DigitalA was 3–7 dB below nominal gain, and DigitalB ranged from 4 dB gain down to 2 dB of attenuation.

Effect of Input Level: again, although all three hearing aids were set to provide linear amplification, in general the measured gain increased slightly as input level increased. For the Analog aid, equivalent gain was measured for the 30 and 50 dB SPL inputs ($p > .1$), and the gain measured for the 70 dB SPL input was greater than both the lower input levels ($p < .1$). For the DigitalA hearing aid, less gain was measured for the 30 compared with the 50 and 70 dB SPL input levels ($p < .05$) and equivalent gain for 50 and 70 dB SPL inputs ($p > .1$). Finally, the DigitalB hearing aid provided less gain for the 30 compared with the 50 ($p < .05$) and 70 dB SPL inputs ($p < .01$) and less gain for the 50 compared with the 70 dB SPL input ($p < .1$).

Effect of Hearing Aid: the DigitalB hearing aid provided significantly less gain than both the DigitalA and Analog hearing aids at every input level ($p < .01$). The DigitalA and Analog hearing aids provided equivalent gain for all input levels ($p > .1$).

3.2.2. 20-dB Gain Condition: Long Stimulus—Gain at Maximum Amplitude. Mean gain values presented in Table 2 show that, when measured at the maximum amplitude for the long stimulus, once again all of the hearing aids provided less than 20 dB gain. The gain values were similar to the maximum gain values obtained with the short stimulus, with the notable exception of DigitalB aid, which now measured only 3–7 dB below nominal gain.

Effect of Input Level: for all three hearing aids, increasing amounts of gain were measured with increases in input level ($p < .1$).

Effect of Hearing Aid: at each input level, there was no difference among hearing aids in the amount of gain measured ($p > .1$).

3.2.3. 20-dB Gain Condition: Long Stimulus—Rise Time. Again, the figures and Table 2 show a clear trend for the two

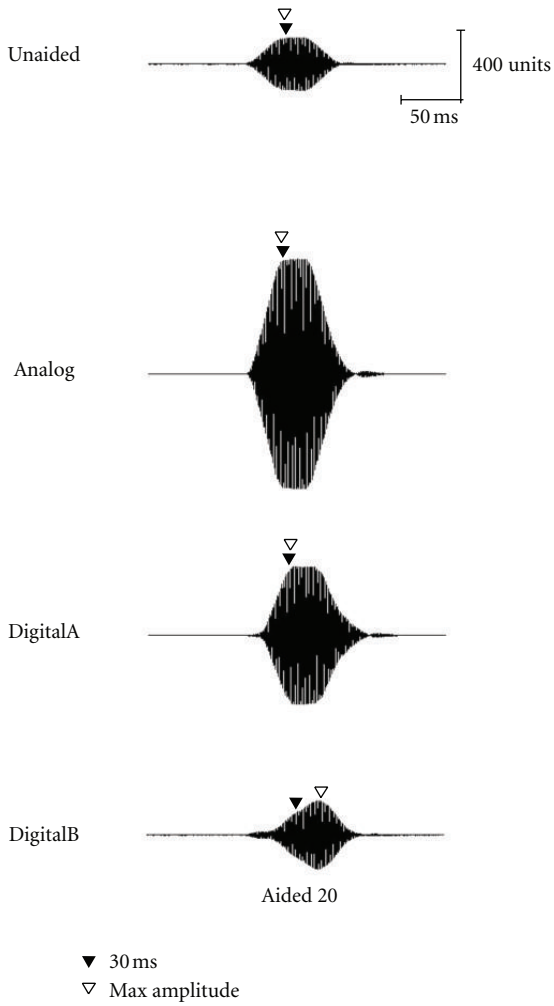


Figure 3: Waveform of the short stimulus presented at 70 dB SPL as measured in the ear canal of a single representative subject (subject 2).

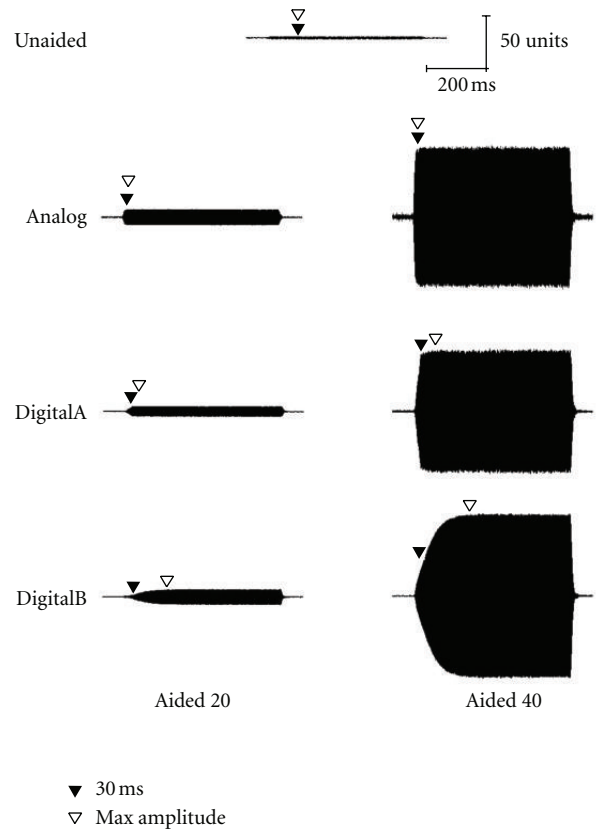


Figure 4: Waveform of the long stimulus presented at 30 dB SPL as measured in the ear canal of a single representative subject (subject 2).

digital aids to take longer to reach maximum amplitude than the analog aid, with the DigitalB aid showing a markedly longer rise time than DigitalA. Rise time for the Analog aid was about 12 ms, only slightly longer than the 7.5-ms rise time of the stimulus. Rise time for DigitalA ranged from 23 to 40 ms, and DigitalB had a much longer rise time of about 140–150 ms.

Effect of Input Level: for DigitalB, the rise time measured for 30 dB was slightly shorter than the rise time measured for the 50 dB SPL input level ($p < .05$). For DigitalA, the rise time measured for 30 was longer than that measured for 50 and 70 dB SPL input levels ($p < .1$). For Analog, rise time was equivalent across input levels.

Effect of Hearing Aid: at every input level, the rise time differed across hearing aids ($p < .01$), with the rise time being longest for the DigitalB hearing aid, shorter for the DigitalA hearing aid, and shortest for the Analog hearing aid.

3.2.4. 40-dB Gain Condition: Long Stimulus—Gain at 30-ms after Stimulus Onset. Mean gain values presented in Table 2 indicate that, similar to the 20-dB gain setting at 30 ms, all of the hearing aids provided less than 40 dB gain. DigitalB provided less gain than either the Analog or DigitalA hearing aids. The Analog aid provided about 5 dB less than nominal gain, similar to the 20-dB gain condition. DigitalA provided about 4–6 dB less than nominal gain, similar to the 20-dB gain condition, and DigitalB provided about 14–16 dB less than the nominal 40-dB gain.

Effect of Input Level: the DigitalB hearing aid provided less gain for the 30 than 50 dB input levels ($p < .1$). Both the DigitalA and Analog hearing aids provided equivalent gain across the two input levels ($p > .1$).

Effect of Hearing Aid: once again the DigitalB hearing aid provided significantly less gain than both other hearing aids at both input levels ($p < .01$), and there was no significant difference in gain provided by Analog and DigitalA at either input level ($p > .1$).

3.2.5. 40-dB Gain Condition: Long Stimulus—Gain at Maximum Amplitude. Mean gain data indicate that, at maximum amplitude for the long stimulus, all three hearing aids provided gain that was within 3–5 dB of the nominal 40 dB gain for all input levels. There were no significant effects of Hearing Aid or Input Level in the ANOVA.

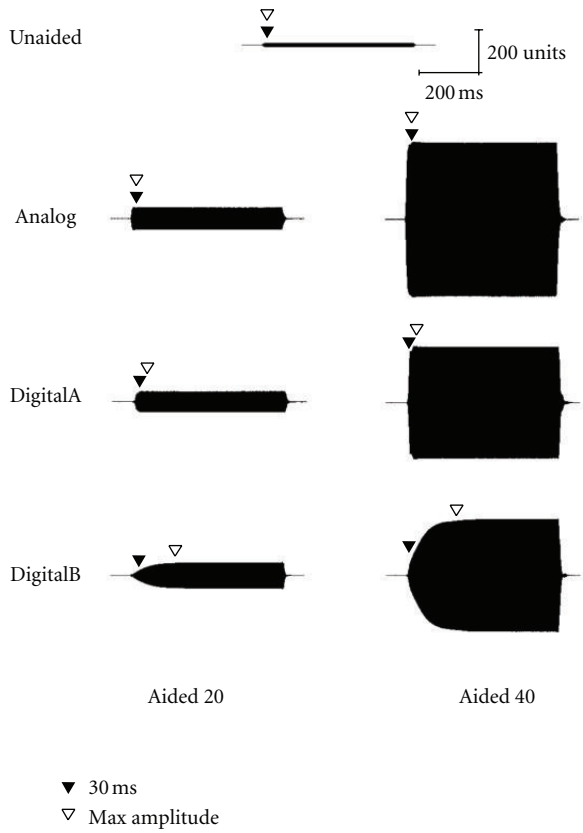


Figure 5: Waveform of the long stimulus presented at 50 dB SPL as measured in the ear canal of a single representative subject (subject 2).

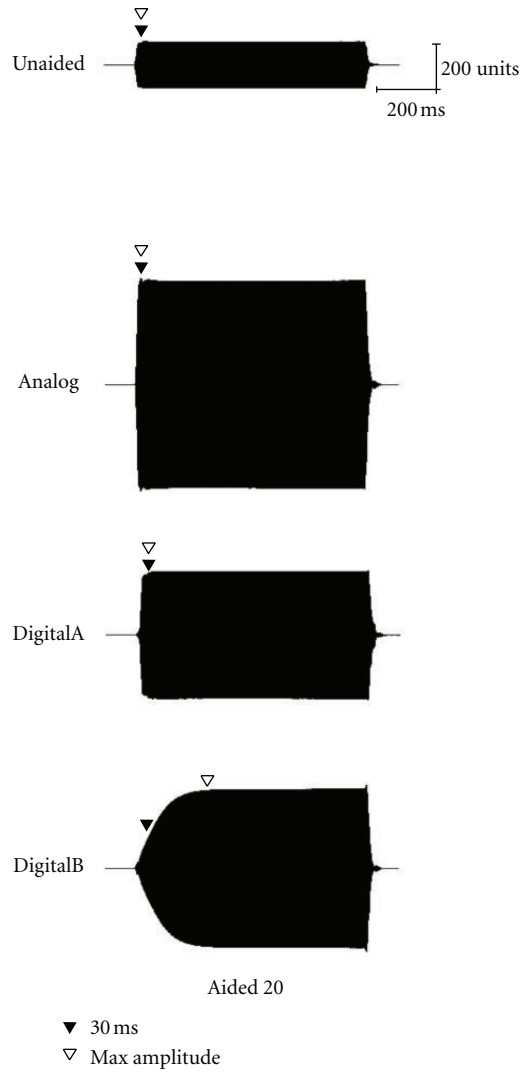


Figure 6: Waveform of the long stimulus presented at 70 dB SPL as measured in the ear canal of a single representative subject (subject 2).

3.2.6. 40-dB Gain Condition: Long Stimulus—Rise Time. Again, there was a clear trend for the two digital aids to take longer to reach maximum amplitude than the analog aid, and the DigitalB aid took longer than DigitalA. The Analog aid was measured to have a rise time of about 12 ms, DigitalA to have a rise time of 19–43 ms, and DigitalB to have the longest rise time at 145–155 ms.

Effect of Input Level: both DigitalA and DigitalB had longer measured rise times for the 30 than 50 dB SPL input level ($p < .1$). Analog had equivalent rise times at the two input levels ($p > .1$).

Effect of Hearing Aid: for both input levels, DigitalB had a longer rise time than both DigitalA and Analog, and DigitalA had a longer rise time than Analog ($p < .05$).

4. Discussion

All three hearing aids provided 20 and 40 dB of insertion gain at mid- and high-level inputs for all subjects when measured with a conventional hearing aid test system (Fonix 7000). When measuring the hearing aids with the stimuli used for the SCP measures, however, all of the hearing aids were measured to have less gain, particularly the two digital aids. The amount of gain reduction differed between the two digital hearing aids, with DigitalB showing much less gain than DigitalA in almost every condition. The two digital hearing aids, DigitalB in particular, reached their maximum gain well past 30 ms after stimulus onset. When the hearing aids were measured with a long stimulus at their maximum gain, there were no longer any differences among hearing aids in the amount of gain measured, but all three hearing aids demonstrated about 3–5 dB less gain than the gain measured with conventional REIG procedures.

The first main finding was that the maximum gain of all of the hearing aids was about 3–5 dB less than nominal gain even when a long stimulus was used. Although the conventional and SCP acoustic measures were made with different sets of equipment, different signals, and in different rooms, these effects cannot account for reductions in measured gain. Both acoustic measures (i.e., standard clinical measures and measures of the SCP stimuli) were insertion gain; thus, gain was always calculated as the aided response minus the unaided response. Additionally, the stimulus for both measures was a 1000-Hz pure tone, at the same input levels. Insertion gain is robust to differences in probe-tube insertion depth, measurement bandwidth, and small

changes in room acoustics, particularly if the hearing aid has linear processing [26–28], which was the case for all three hearing aids in this study. Care was taken to ensure that head movement did not lead to substantial changes in the soundfield during SCP testing, which was calibrated with a substitution method rather than the on-line corrections of the commercial hearing aid test system. Because this difference was found for both the analog and digital hearing aids, it is not due to the type of processing (analog versus digital) or programming.

It was a consistent finding that less gain was provided by digital hearing aids for the 30 dB SPL input level; this is likely due to the low-level expansion in both hearing aids, which could not be changed or disabled in the programming. Gain for a 30 dB SPL input was not measured with the standard hearing aid test system, as signals that low are not provided. However, reduced gain for low-level inputs cannot explain why nominal gain was not achieved even at higher input levels.

The second main finding was that both digital hearing aids altered the rise times of the stimuli such that there was a significant delay for both hearing aids to reach their maximum gain, and the amount of delay differed significantly between the two digital aids. This might be expected because of the commonly reported delays associated with digital processing [1, 29–35]. However, processing delays cannot account for the altered rise times measured for the SCP stimuli in this study for two reasons: first, processing delay was removed from the calculation of rise time by determining the 0-ms point as the time at which periodicity was first noted in the recording, rather than the time of signal presentation; second, even if this method did not fully remove the effects of processing delay, the results are inconsistent with the electroacoustic measures of delay. Electroacoustic measures of delay conducted on the Fonix 7000 system indicated that both digital hearing aids had longer delays (6.8 ms and 2.3 ms for DigitalA and DigitalB, resp.) compared with the Analog hearing aid (0.4 ms). Recall that the stimulus rise time was 20 ms for the short stimulus, so maximum amplitude would not be expected until 20 ms. Any processing delays could cause the maximum amplitude to be reached later than 20 ms. DigitalA did reach its maximum amplitude by 28 ms in some conditions, which is close to what would be expected if the electroacoustic measure of delay (6.8 ms) was added to the 20 ms stimulus rise time. In some conditions, however, DigitalA did not reach its maximum amplitude until 39 ms, beyond what could be explained by processing delay. Perhaps a stronger argument against the conventional measure of processing delay as an explanation for these results is that DigitalB, measured with the Fonix system to have only 2.3 ms processing delay, had the longest measured rise times, of 40 ms for the short stimulus and 150 ms for the long stimulus. Note the difference in measured rise times between short and long stimuli is due to the characteristics of the input signal; at 40 ms, the short stimulus was beyond its plateau and beginning to decrease in amplitude.

These changes in altered rise time are also unlikely to be due to hearing aid processing parameters. All of the hearing aids were set (and subsequently verified) to linear processing. Any compression processing, had it remained on, would be expected to have the opposite effect as found here; that is, compression would be associated with faster rise times than measured here due to the overshoot that results from compression attack time [36, 37]. All other features were disabled, but again, features such as noise reduction or feedback reduction would demonstrate the opposite effect to the one measured in this study; that is, those features would be expected to show a gradual decrease in gain for the nonspeech pure tone [38, 39]. Thus, it is not immediately apparent what could account for the two main acoustic findings of this study. Because of the unknown and somewhat random differences between the two digital hearing aids, it is clear that the stimulus used for testing aided SCP responses must be carefully evaluated with acoustic measures across a range of hearing aid types and ultimately with typical processing features enabled.

It is worth noting that the issues identified in this study are unlikely to be problematic only when using tonal stimuli, even though tonal stimuli have proven to be troublesome for measuring digital hearing aid processing [16, 39, 40]. Tremblay et al. [6] used speech stimuli to examine the effects of amplification on SCP responses and found that even providing 12–26 dB of gain had no effect on N1-P2 amplitude. Detailed acoustic analysis of their stimuli was not provided, but the lack of an amplification effect for speech stimuli suggests that the hearing aid processing altered their speech stimuli in such a way that affected the SCP measurements.

4.1. How Well Did These Acoustic Measures Predict the SCP Responses? In our previous study [11], we demonstrated that the SCP responses measured for hearing-aid processed signals did not have the expected increases in amplitude and decreases in latency that would be predicted from 20 or 40 dB of gain added by the hearing aids. In the acoustic measures of the current study, we demonstrated that (a) the hearing aids failed to achieve 20 or 40 dB of gain when measured with the SCP stimuli and (b) the digital hearing aids, in particular, reached their maximum gain much later than 30 ms after stimulus onset. The acoustic waveforms show that shape varies across hearing aids for both short- and long-duration stimuli, particularly the onset, which is reflected in the measured rise times. The maximum gain is reached more gradually in DigitalB.

To determine whether any of the measured acoustic parameters could predict the SCP response, we conducted an analysis of the group mean data from both studies. Although a thorough answer to this question would require a larger-scale parametric investigation of the relationship between acoustic variables and the SCP responses, some initial exploration of the findings can be instructive. Specifically, we used the group-mean SCP amplitude from Marynewich et al. [11] and developed a model of the relationship between acoustic measures and N1-P2 amplitude using the unaided responses. That is, we calculated the linear regression between N1-P2 amplitude and each of the three acoustic parameters: stimulus level at 30 ms, maximum amplitude, and slope of onset. See Figure 7 for stimulus level at 30 ms (left panel),

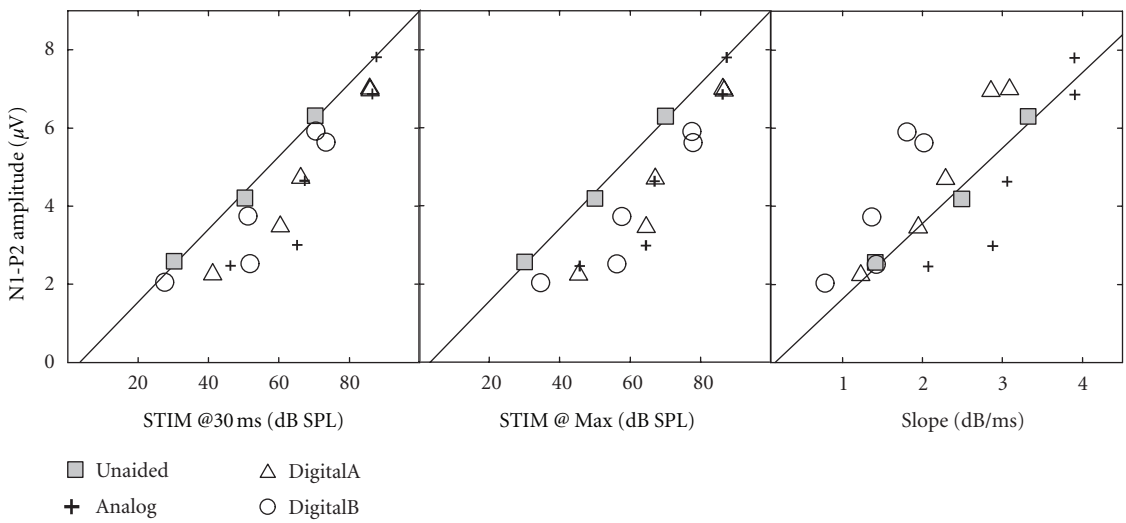


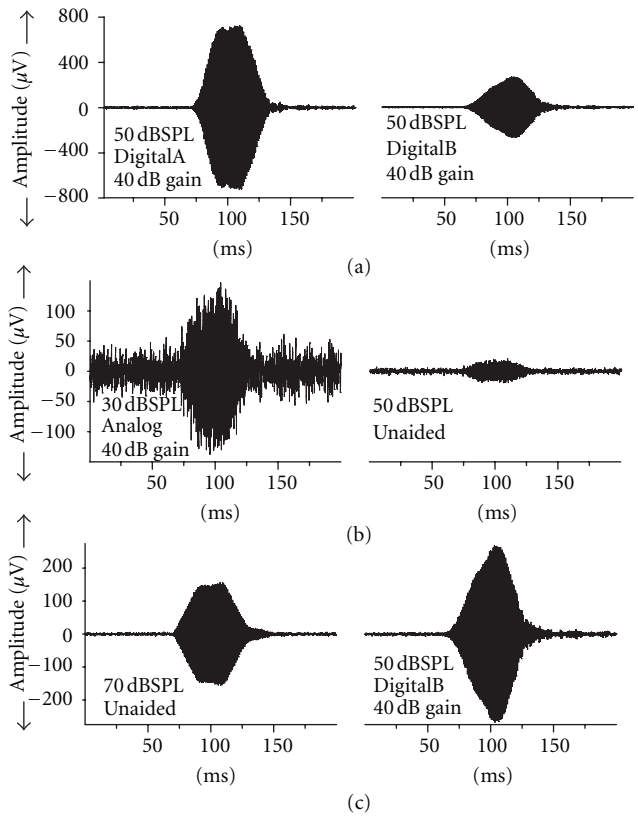
Figure 7: Group mean SCP amplitude from Marynewich et al. [11] as a function of three acoustic parameters: stimulus level at 30 ms (left panel), maximum stimulus level (middle panel), and onset slope (right panel) for unaided and 3 hearing-aid conditions. The fit line is the linear regression for the unaided condition.

maximum stimulus level (middle panel), and onset slope (right panel), where the data for the unaided condition are shown as shaded squares and the fit line is the linear regression for the unaided condition. The data for the aided conditions are also plotted on each panel. To determine how well each acoustic parameter predicted the N1-P2 amplitude in the aided conditions, we calculated the absolute value of the error between the actual mean aided SCP amplitudes and the SCP amplitudes predicted from the unaided data. As a rough estimate of overall how well each acoustic parameter could explain the data observed, we averaged the error across all input, gain, and hearing aid conditions.

The results of this analysis showed that both stimulus amplitude at 30 ms (average error: 1.08 μV) and slope of onset (average error: 1.08 μV) predicted the SCP amplitude equally well. The greatest amount of error was seen for maximum stimulus amplitude (average error: 1.35 μV). The analysis is limited because it was performed on group mean data for two different groups of subjects. However, it is likely that the group analysis is representative of individual analysis for several reasons: four participants were in both studies; the hearing aids were set for individual ears; all participants had normal hearing; the acoustic measures generally had low variability. We can cautiously interpret this analysis to mean that the effect of hearing-aid processing on the onset characteristics of the stimulus had a greater influence on SCP amplitude than did the effect of hearing-aid processing on maximum stimulus amplitude. Recall that gain at 30 ms was generally much lower than the nominal gain, and particularly for DigitalB often measured close to 0 dB gain for the 20-dB gain condition. Thus, if the SCP was responding to the first 30 ms after stimulus onset, it is not surprising that primarily there was often little to no difference between aided and unaided acoustic measures, especially for the digital aids. These results are consistent with the view that approximately the first 30 ms of stimulus onset largely determines SCP N1 presence and amplitude [4, 18–21]. However, this interpretation cannot explain why even the Analog hearing aid only showed significant increases in SCP amplitude for the higher input levels.

4.2. Stimulus Level versus SNR. Because Billings et al. [2, 3, 24] hypothesized that their SCP results were due to the SNR in the aided condition, we conducted a brief analysis of SNR for one of our participants. We chose to do the analysis for one participant who had participated in both the current study and in our previous study [11]. For this ad hoc analysis, we measured signal and noise levels on single-trial recordings rather than averages. Noise was the RMS level in a 1/3rd-octave band centred at 1000 Hz measured over a 70-ms period prior to stimulus onset. For each stimulus condition, measures of the noise levels were made for three separate samples and averaged. Stimulus level was the amplitude at 30 ms, as reported in Section 3. From these calculations, we found that SNR might explain some of the results in the SCP response, but not all of them; generally there was very little noise in the acoustic recordings, even in the aided condition. Three examples have been chosen to illustrate the relationship between SNR, stimulus level, and N1-P2 amplitude; these are shown in Figure 8.

Panel (a) of Figure 8 shows an example where SCP N1-P2 amplitude seems best related to the stimulus amplitude at 30 ms. The first column of Panel (a) shows a single-trial waveform for DigitalA aid at a 50 dB SPL input level with 40 dB gain. The second column shows a single-trial waveform for DigitalB at 50 dB SPL with 40 dB gain. The table to the right provides the relevant data points for input level, stimulus level at 30 ms, noise level, SNR, and SCP N1-P2 amplitude. In this example, although the SNRs are similar for the two conditions, SCP amplitude differs in the same way as the stimulus level changes.



Aid (dB gain)	Input (dB SPL)	Stim amp (μV)	Noise amp (μV)	SNR (dB)	N1-P2 amp (μV)
Dig-A (40)	50	297	2.4	42	7.8
Dig-B (40)	50	84	0.8	40.9	3.5
Anlg (40)	30	68	9.1	17.5	3.6
Unaid	50	9	1.1	18.3	3.8
Unaid	70	83	0.6	42.8	5.3
Dig-B (40)	50	84	0.8	41.1	3.5

Figure 8: Three examples measured for one participant to show the relationship between stimulus level, SNR, and SCP amplitude. (a): stimulus level predicts SCP amplitude; (b): SNR predicts SCP amplitude; (c): neither stimulus level nor SNR predict SCP amplitude.

Panel (b) of Figure 8 is an example that is in agreement with Billings et al. [2, 3, 24], where the SCP response seems to be related to SNR. Comparing Unaided for a 50 dB SPL input level to Analog 30 dB SPL input plus 40 dB gain, the SCP N1-P2 amplitudes are almost identical. In this case, the stimulus levels are very different, but the SNRs are similar and both could be considered good. In this case, the good SNR might be the predictor of the SCP response rather than the stimulus level.

Finally, in Panel (c), there is an example where neither stimulus amplitude at 30 ms nor SNR seems to be good predictors of the SCP response. In this example, comparing 70 dB SPL unaided to DigitalB 50 dB SPL input +40 dB gain, these measures have similar stimulus levels and similar (good) SNRs, yet the SCP response for DigitalB is much lower than unaided. In this case, neither stimulus level nor SNR can explain the different SCP responses observed.

Altogether, this set of examples shows that SCP amplitudes in our data set may be accounted for by changes to rise time in some conditions, SNR in some conditions, and some other, as yet unidentified, acoustic parameter in other conditions.

5. Conclusions

In the present study, we attempted to determine why several recent studies, including our own study [11], have been unsuccessful in demonstrating a significant amplification effect on SCP measures. We reduced sources of unknown variability as much as possible by using hearing aids with linear processing and all features disabled. The acoustic measures of the amplified SCP stimuli showed that (a) the hearing aids all provided less than expected gain for these stimuli and (b) the digital hearing aids took longer to reach their maximum gain than the analog hearing aid. The acoustic measures of stimulus level at 30 ms and onset slope were predictive of SCP response amplitude, but it is likely that additional acoustic characteristics, not measured in this study, contributed to SCP response.

In light of findings from the present study, it is likely that prior studies using speech or tonal stimuli [2, 4–9, 41] were measuring SCPs to stimuli that were substantially 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 [12, 13], 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 [1]; 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, despite the hearing aids being set to provide linear gain with all advanced processing features disabled.

The less-than-expected measureable gain resulting from hearing-aid processing for SCP stimuli suggests that SCP stimuli do not provide appropriate measures of hearing aid gain. Our acoustic analysis shows that changes to rise time, particularly in ways that affect stimulus amplitude at 30 ms, may explain our previous findings [11]. However, we cannot rule out SNR, or even another acoustic parameter, as a potential contributor to the SCP measures of Marynewich et al. [11]. As a result of these unknown factors, more research concerning aided-SCP testing is needed for clinical application of this technique, and any results must be interpreted very cautiously if used within the hearing-aid fitting process. As has been noted by others, the lack of an SCP response does not ensure that the stimulus is inaudible [42–44]; similarly, a "present" aided SCP does not ensure that the stimuli are sufficiently audible [10].

Although our studies included only participants with normal hearing, the acoustic alterations of the stimuli that we measured are independent of hearing status. The concerns raised by these studies indicate that much is unknown about the application of SCP measures in hearing aid fitting. Future research might involve (i) additional hearing aid measures to determine the source of alteration to rise time and (b) parametric study of the relationship between the stimulus acoustic measures and the SCP responses. Future research might also explore different/more-appropriate SCP stimuli or presentation paradigms for hearing aid measures to determine under what conditions aided SCP measures would be valid.

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Research Article

A Pilot Study on Cortical Auditory Evoked Potentials in Children: Aided CAEPs Reflect Improved High-Frequency Audibility with Frequency Compression Hearing Aid Technology

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Background. This study investigated whether cortical auditory evoked potentials (CAEPs) could reliably be recorded and interpreted using clinical testing equipment, to assess the effects of hearing aid technology on the CAEP. *Methods*. Fifteen normal hearing (NH) and five hearing impaired (HI) children were included in the study. NH children were tested unaided; HI children were tested while wearing hearing aids. CAEPs were evoked with tone bursts presented at a suprathreshold level. Presence/absence of CAEPs was established based on agreement between two independent raters. *Results*. Present waveforms were interpreted for most NH listeners and all HI listeners, when stimuli were measured to be at an audible level. The younger NH children were found to have significantly different waveform morphology, compared to the older children, with grand averaged waveforms differing in the later part of the time window (the N2 response). Results suggest that in some children, frequency compression hearing aid processing improved audibility of specific frequencies, leading to increased rates of detectable cortical responses in HI children. *Conclusions*. These findings provide support for the use of CAEPs in measuring hearing aid benefit. Further research is needed to validate aided results across a larger group of HI participants and with speech-based stimuli.

1. Introduction

A growing body of literature exists on the use of cortical auditory evoked potentials (CAEPs) in assessing neural activity in NH and HI listeners. Such measures reflect the sum of synchronous, time-locked neural activity detected at the level of the central auditory system, related to the strength (amplitude) and timing (latency) of a response [1, 2]. For these reasons, CAEPs have been suggested for clinical use in monitoring changes in neural activity associated with auditory rehabilitation (e.g., hearing aids). The P1-N1-P2 complex is one type of evoked potential, comprised of slow components ranging from 50 to 300 msec in latency [2]. The peaks of the complex are thought to reflect neural activation of the central auditory system in response to the spectral and temporal properties of a given stimulus [3, 4]. Studies including normal hearing (NH) and hearing impaired (HI) participants conclude that CAEPs characterized the P1-N1-P2 complex, elicited by tonal stimuli and various speech tokens, can be reliably recorded to produce distinct neural patterns in both aided and unaided conditions [3, 5–7]. Such literature is mainly comprised of studies including research-grade equipment using multiple testing channels (greater than two). Few studies have looked at using CAEPs with commercially available clinical testing equipment (e.g., Hear lab). Commercially available systems are commonly comprised of a single-channel recording system [8, 9]; such equipment has been proposed for use in aided assessment

to evaluate infant hearing aid fitting using CAEPs [10]. Literature suggests that aided CAEPs may be able to provide information related to neural detection and audibility of aided and unaided sound [7, 8]. Further research is needed to help quantify the relationship between CAEPs and hearing aid benefit.

When sound is processed through a hearing aid, it is necessary to understand what the hearing aid is doing to the signal. For this reason, recent research in this area has focused on amplification-related modifications to the CAEP stimulus, related to factors such as poor signal-to-noise ratio (SNR) and/or rise time [11–13]; both of which may interact with the input stimulus level used in the fitting process. Billings and colleagues (2011) have investigated whether hearing aids modify stimulus characteristics such as SNR, suggesting that these changes can affect the CAEP in a way that does not reliably reflect hearing aid gain [11, 14]. Low stimulus levels were used to avoid loud hearing aid output levels with NH listeners [11]. Such stimulus levels are atypical in studies of hearing aid validation where supra-threshold levels representing conversational levels of speech (e.g., 55–75 dB SPL) are traditionally used [15]. Because the stimulus level can interact with nonlinear signal processing in hearing aids, a lower level stimulus will receive more gain than higher level stimuli. For this reason, consideration of stimulus level is important if the goal of CAEP measurement is to characterize the aided response to sound. The effect of SNR has been investigated in NH listeners only and hence bound to vary with HI listeners as the effect of SNR is largely dependent on the relative level of hearing aid internal noise and hearing thresholds. In a recent study by Easwar et al. [16], the stimulus onset altering effect of a hearing aid on CAEPs elicited with tone burst stimuli of varying rise times was investigated. Findings suggest that alterations in the tone burst stimuli caused by hearing aid processing are not large enough to significantly influence CAEP responses in NH listeners with hearing aid processed stimuli [16]. Further research is needed to validate findings with HI listeners, fitted with varying types of clinically common hearing aids. In general, these studies alert clinicians and researchers of the potential for hearing aid signal processing to interact with CAEP stimuli, in ways that may or may not affect the response. More research is needed in this area to understand fully whether CAEP methodologies are sensitive to other aspects of hearing aid signal processing, for several reasons. One reason may be to investigate whether signal processing acts as a confound: does it alter the CAEP stimulus in an unexpected manner, as has been investigated elsewhere [12, 13, 17]. Another reason may be to pursue whether CAEP changes correlate with behavioral changes, specifically for hearing aid signal processing that causes performance improvement or decrement. This paper will mainly consider the latter type of question.

One example of a change in hearing aid signal processing is the application of frequency lowering. Studies suggest that frequency lowering hearing aid technology may benefit adults and children with high-frequency hearing loss (refer to [17] for a summary). For the purpose of this paper, change in audibility due to frequency lowering in the form of nonlinear frequency compression (NLFC) was investigated using CAEPs. In general, NLFC splits the incoming hearing aid signal into two channels. The high-frequency channel is compressed into a narrower bandwidth. This results in sound being lowered in frequency within the high-frequency channel only [18]. Research suggests that NLFC hearing aid processing can provide speech sound detection perception benefit for some adults and children with high-frequency hearing loss and that benefit varies across individuals [18–23]. Specifically, listeners with severe-to-profound high-frequency hearing loss are most likely to derive large benefits, because the technology changes audibility of high-frequency cues. Therefore, the aided CAEP may be sensitive to the increased audibility produced by NLFC hearing aids. However, no studies have investigated whether aided CAEPs might change in response to NLFC activation.

The present study investigated the use of CAEP measures to evaluate audibility of tone bursts with and without NLFC frequency lowering, using hearing aid-processed stimuli with HI listeners, in comparison to a group of NH listeners. Since the end goal of using aided CAEPs is to evaluate the benefit of hearing aid fittings clinically, this study uses commonly available clinical equipment for recording CAEPs. Research grade equipment with multiple channels including a channel to detect eye blink, although may provide higher quality data with additional information such as scalp topography, may be difficult to use clinically due to time, cost, and feasibility concerns. Modifications to clinically available testing equipment, the Bio-logic Navigator Pro System, allowed tone burst stimuli to be presented directly to the direct audio input (DAI) of a hearing aid with an audioshoe connector in aided testing conditions. The purpose of this study was to investigate the following: (1) whether CAEPs could be reliably recorded and interpreted using Bio-logic Navigator Pro testing equipment in NH and HI children, and (2) whether CAEPs elicited by high-frequency tone burst stimuli reflected the change in high frequency audibility due to use of NLFC hearing aid technology.

2. Materials and Methods

2.1. Participants. Participants included 15 NH children (mean = 14.4 years, range = 11–18 years) and 5 HI children with high-frequency hearing loss (one 11-year old, one 14-year old, and three 18-year olds). Children in this age range were chosen for this pilot study as they are likely to understand and follow instructions during testing (e.g., keep awake during recording) and their CAEPs may be considered representative of a maturing system [24]. NH children were included to provide a reference of typically maturing CAEPs without the influence of hearing impairment. The HI children were recruited from a study of acclimatization to frequency lowering hearing aids [25], in which they were provided with study hearing aid for approximately four months in duration. In the present study, data from one HI child were excluded because of unexplained acoustic artifacts in the hearing aid output measured with study-specific stimuli. Pure-tone air conduction testing was carried out in a

double-walled sound booth using Etymotic Research ER-3A insert earphones across octave and interoctave audiometric frequencies between 0.25 and 8 kHz (GSI-61 Audiometer; [26]). Routine otoscopic examination ruled out any contraindications such as active discharge, occluding wax, or foreign body in the ear canal. Participants reported no significant history of neurological or otological problems. The study protocol was approved by the University of Western Ontario Health Sciences Research Ethics Board.

The eligibility criteria for the NH participants included passing a hearing screen at 15 dB HL. For the HI participants, hearing thresholds were measured by coupling the insert earphones to each participant's personal earmolds. Eligibility criteria included bilateral sensorineural hearing impairment, sloping to at least a moderately severe high-frequency pure-tone average (HF-PTA) hearing level averaged across 2, 3, and 4 kHz. Hearing threshold data is displayed for each case in the Results section. Participants were required to be full-time users of digital behind-the-ear (BTE) hearing aids prior to entering the study and to maintain full time use of the study hearing aids prior to CAEP measurement. Four of the participants were binaural hearing aid wearers with symmetrical high-frequency hearing loss within 10 dB, based on HF-PTA. One participant with an asymmetrical hearing loss was monaurally aided in the better ear (Case 5); a hearing aid trial on the ear with a greater level of impairment (i.e., a profound hearing loss) was not successful. Hearing thresholds were measured at the beginning and end of the study. All participants demonstrated hearing levels within 10 dB of baseline over the course of the study. The presence of cochlear dead regions was assessed using the TEN test in dB HL [27].

2.2. Stimuli. Tone bursts at 2 and 4 kHz were generated from the Bio-logic Navigator Pro (v7.0.0). Using stimulus parameter options available within the Bio-logic software, cycle numbers were held constant across stimuli with a rise/fall Blackman ramp of 20 cycles each and a plateau of 80 cycles. This produced a 40 msec plateau and 10 msec rise/fall time for the 2 kHz tone burst and a 20 msec plateau and 5 msec rise/fall time for the 4 kHz tone burst.

2.3. Testing Conditions. Testing was carried out in sessions no longer than two hours. One session was required for NH participants and two were required for HI participants. Breaks were given when requested. Participants were seated watching a muted movie of their choice with active subtitles and were instructed to ignore the stimuli played [28, 29]. The sequence of stimulus presentation in a session was randomized. For the NH participants, test ear was alternated across participant order numbers. Unaided monaural testing was completed by routing the stimuli directly from the Navigator Pro to Bio-logic broadband insert earphones coupled to foam tips.

For the HI participants, aided monaural testing was carried out for the better listening ear, according to pure-tone average (PTA). Stimuli were routed directly from the Navigator Pro (and through an audio isolation transformer) to the DAI of the hearing aid with an audioshoe connector coupled to a study worn hearing aid, using a custom cable [30]. This DAI strategy was chosen to remove the effects of room acoustics, head azimuth/movement, and listener distance from a loudspeaker that could have affected the accuracy and reliability of sound-field presentation of stimuli. The stimulus presentation level strategy described below was chosen to present the unaided stimuli to the NH participants at a dial level of 70 and to provide individualized amplification to each HI participant for the same input level using the DAI routing. The DAI routing was verified as acoustically transparent (within 2 dB) by routing the test signals through a study hearing aid programmed to have no amplification, through a 2cc coupler to a Type I sound level meter (Larson-Davis 824). The sections below outline the procedures used to derive individual hearing aid fittings and details of stimuli and calibration.

2.3.1. Individualized Hearing Aid Fittings. Device fitting followed protocols from the Desired Sensation Level (DSL) method version 5.0 [31, 32] as implemented within the Audioscan Verifit VF-1 (a clinical test system used for hearing aid analysis). Each participant was fitted with Phonak Naida IX SP BTE hearing aids. Hearing devices were worn for the entire duration of the study with the gain/advanced features held constant throughout. Volume control, digital noise reduction, and automatic program selector features were disabled. Prescriptive targets were matched using simulated real ear measures incorporating individual real ear to coupler difference values. We selected a coupler-based verification strategy to reduce room noise/reverberation effects and concerns with feedback during verification; this promoted test environment consistency and replicable measures across the repeated fitting appointments. Aided test box measurements of speech at 55, 65, 70, and 75 dB SPL and for a 90 dB SPL pure-tone signal were completed during fitting appointments. Hearing aids were adjusted to provide the best possible match to targets. NLFC settings (i.e., cut-off frequency and compression ration) were individualized according to established procedures [30, 33] using manufacturer specific fitting software (Case specific settings are provided in Section 3 (Figures 3 to 7)).

Aided CAEP testing was completed on two different testing sessions: the first with NLFC enabled (treatment condition) and the second with NLFC disabled (no-NLFC). On average, the NFLC testing session was completed after 15 weeks of acclimatization to NLFC processing (range: 14 to 17 weeks). The no-NLFC testing session was completed 4 weeks after finishing the treatment condition of the study (range: 1 to 7 weeks). Participants were naïve to all details pertaining to the study design. Such details, along with the individualized results, were disclosed to the participants upon completion of the study.

The potential for hearing aid-induced delay affecting latency values in HI testing conditions was measured using an anechoic box (B&K 4232). Stimuli were presented by playing 2 and 4 kHz tone burst stimuli from SpectraPlus software via a study hearing aid connected to the coupler.

The output of the coupler and reference channels of the recordings were compared to each other using a 75 dB SPL input level to estimate the presence of any hearing aid-induced delay. The difference in the onset of stimuli in each recording channel was calculated to be 6.7 msec, on average, across stimuli and testing conditions. Delay values ranged from 6.3 to 7 msec. Due to the insignificance of the calculated delay values compared to the latency of CAEPs, no corrections were applied for the purpose of comparing HI and NH CAEP data.

2.4. Presentation Level. A suprathreshold presentation level was determined by generating a tone burst stimulus from the Navigator Pro at a testing dial level of 70 dB in the anechoic test box. The output was measured via broadband insert earphones connected to an HA-2 (25 mm tubing) 2cc coupler and microphone (B&K 4192). SpectraPlus software was used to capture the output RMS of the coupler in dB SPL across the tone burst plateau. The chosen dial level produced presentation levels of 69 and 71 dB SPL (re: 2cc coupler) for the 2 and 4 kHz tone burst stimuli, respectively.

The same dial level was used for aided testing with HI participants. Additional electroacoustic verification measures of aided stimuli were completed using the same set-up as described previously. Electroacoustic measures allowed individualized estimation of audibility of the tone bursts for each of the hearing aid fittings and testing conditions. Navigator Pro tone bursts generated at the chosen dial level were routed through the study hearing aid set-up, programmed with individualized fittings. Overall rms of each tone burst (including rise, plateau, and fall time) was measured for individual fittings with and without NLFC for each stimulus. Since these measures were made in a 2cc coupler, audiometric thresholds were transformed to SPL in a 2cc coupler using individualized ear canal transforms [31], and the sensation level of each stimulus was computed as aided RMS level minus audiometric threshold, with all values in coupler SPL. To account for the shorter duration of the tone bursts during estimation of tone burst SL values, correction factors of 3 and 6 dB at 2 and 4 kHz, respectively, were subtracted from the SL value obtained for pure tones used during audiometry [34]. A summary of these results along with those for all corresponding CAEP measures is presented in the results section (Table 1).

2.5. Set-Up of CAEP Equipment. An ipsilateral recording (Vertex to ipsilateral mastoid with ground Fpz) was obtained using the Navigator Pro. Tone bursts were presented at the rate of 0.5 stimuli/sec. This interstimulusinterval was the same as that used to acoustically record in the aided condition. Each recorded electroencephalogram (EEG) sweep included 100 msec of prestimulus baseline (relative to tone burst onset) and 966 msec of post stimulus activity. EEG was amplified 50000 times and digitized at the rate of 480.03 Hz. Responses were bandpass filtered between 0.1 and 100 Hz online. The artifact rejection threshold was set to $\pm 100\,\mu$V. Two averages of 100 sweeps each were obtained for each stimulus condition. Due to study-specific modifications to the traditional testing parameters used in the Bio-logic software, including epoch time/stimulus rate, it was not possible to include a calculation of residual noise level as a stop criterion.

2.6. Waveform Interpretation. Data extraction from the Bio-logic Software was limited to averaged data; therefore, previously reported statistical techniques [35] could not be applied to this data set. Averaged CEAP waveforms were exported from the testing equipment and postprocessed using a MATLAB script version 2008b (Mathworks, 2008) including a second order bandpass Butterworth filter (1–15 Hz). Replicated CAEP waveforms for all participants and across testing sessions were then interpreted by two experienced raters using subjective response detection techniques. Data between 0 and 400 msec after stimulus onset were included in rater interpretations [36]; this included a time window spanning beyond that of a traditional P1-N1-P2-N2 complex [2, 37].

Research on the maturational effects associated with the CAEP response suggests that changes to the waveform morphology associated with the P1-N1-P2 complex can be observed between the ages of 6 and 18 years [2, 24]. Therefore, based on the age range assessed in this study (11–18 years), the authors chose to interpret the data according to the presence/absence of one or more peaks, rather than using peak picking according to traditional latency values associated with specific responses in the P1-N1-P2 complex. The decision regarding response presence or absence required the agreement of both raters; disagreement resulted in a rating of "absent" for the CAEP response in question. Each rater was blind to the test condition and to the other rater's judgments during interpretation. For a response to be considered present the raters had to agree that at least one peak (according to replicable data) resided within the chosen time window. Waveform interpretation results were used in group level analyses for the NH group and in single-subject analyses for the HI cases. In single-subject designs, each participant serves as his or her own control allowing for the opportunity to measure significant changes in performance at the individual level [38], in this study, changes as a consequence of enabling frequency compression. This type of analysis was of particular interest in this study given the varying hearing loss degrees/configurations present in the HI cases, which required individualized frequency compression settings.

3. Results

3.1. Interrater Agreement. An interrater reliability analysis using the Kappa statistic was performed using SPSS software to examine consistency among the two raters. Analyses were performed including waveforms from all participants in both conditions. The interrater reliability analysis for the examiners, across all stimuli and conditions, was found to be Kappa = 1 ($P < 0.001$), 95% CI (1, 1), with a standard error of zero. A Kappa value of one implies perfect agreement between the two raters.

Table 1: Summary of electroacoustic verification results for all HI cases (1 through 5). Summaries are shown across stimuli (2 kHz and 4 kHz tone bursts) and hearing aid conditions (NLFC enabled and no-NLFC). Results are compared to the corresponding cortical auditory evoked potentials (CAEP) judged as present or absent per condition.

Case	2 kHz NLFC		2 kHz no-NLFC		4 kHz NLFC		4 kHz no-NLFC	
	SL (dB)	CAEP	SL (dB)	CAEP	SL (dB)	CAEP	SL (dB)	CAEP
1	30.29	Present	30.24	Present	11.47	Present	5.36	Present
2	19.36	Present	19.86	Present	10.06	Present	0.39	Absent
3	13.1	Present	13.36	Present	16.29	Present	6.19	Absent
4	9.68	Present	9.66	Present	−7.24	Present	−18.25	Absent
5	0.02	Present	6.51	Present	−10	Present	−18.6	Absent

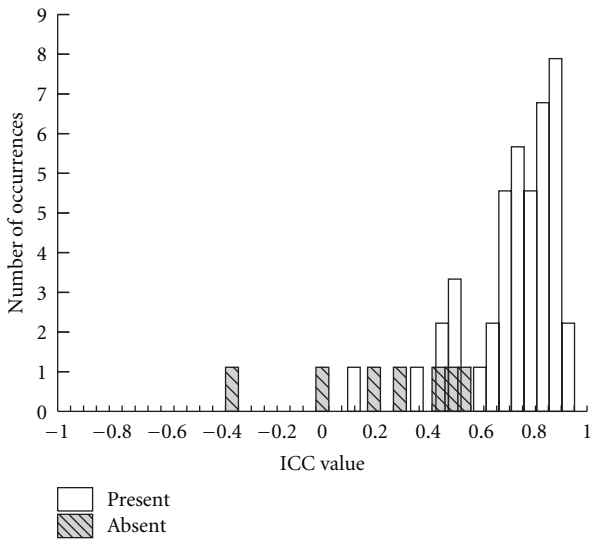

Figure 1: Histogram of ICC values for CAEPs waveforms judged to be present and absent.

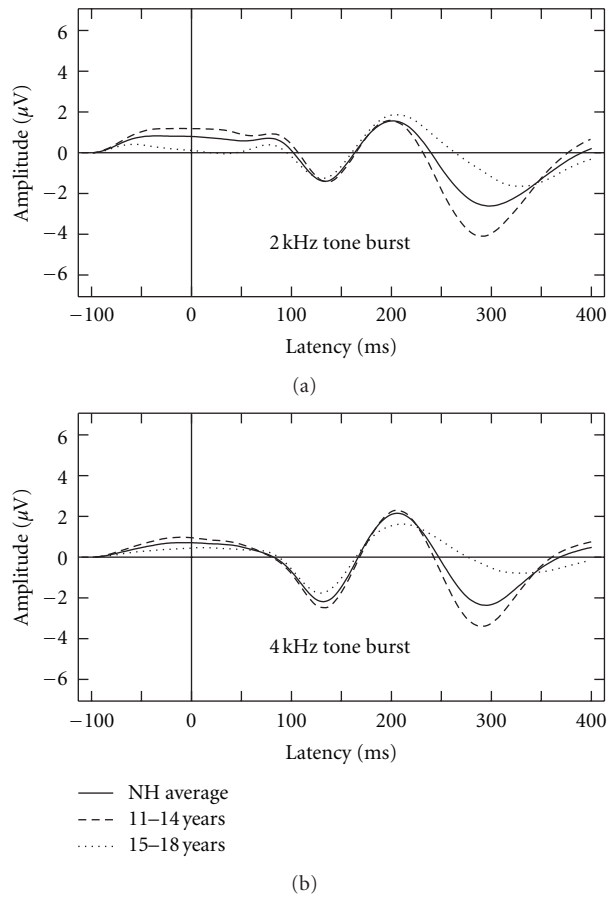

Figure 2: Averaged waveforms displayed according to stimulus for all NH groups including an average of all participants (solid line), those between the ages of 11 to 14 years (dashed line) and those between the ages of 15 to 18 years (dotted line).

3.2. Objective Index of Replicability of CAEP. The intraclass class correlation coefficient (ICC) has recently been used to quantify similarity between two waveforms [36, 39]. Visual inspection of waveform similarity and corresponding ICC values between two CAEP waveforms have been shown to agree at the group level [40]. The ICC value can provide a measure of the similarity of the waveform amplitude and shape, with higher values representing greater similarity between evoked responses [36, 40]. To investigate the use of ICC measures in waveform interpretation, the present study explored the relationship between ICC values computed for repetition 1 and repetition 2 of the recorded waveforms and final subjective interpretation of presence versus absence of the CAEP. For this purpose, one-way random ICCs were computed between two averages (repetition 1 and repetition 2) for each stimulus condition, for each hearing aid condition. The same time window used for subjective waveform interpretation was used when computing ICC values. Since the distribution of ICC values does not follow a normal distribution, median values are reported across all stimuli and conditions, and 95% confidence intervals (CI) have been limited to the range of the data in the following summary. The median ICC for CAEPs that were judged to be present subjectively was 0.816 (range: 0.11 to 0.97; CI = 0.27–0.96) and median ICC for CAEPs that were judged to be absent subjectively was much lower, 0.278 (range: −0.3 to 0.5; CI = −0.42–0.5). ICC values above .75 are indicative of good reliability [41]. A histogram of ICC values for CAEP waveforms subjectively judged as present and absent is illustrated in Figure 1.

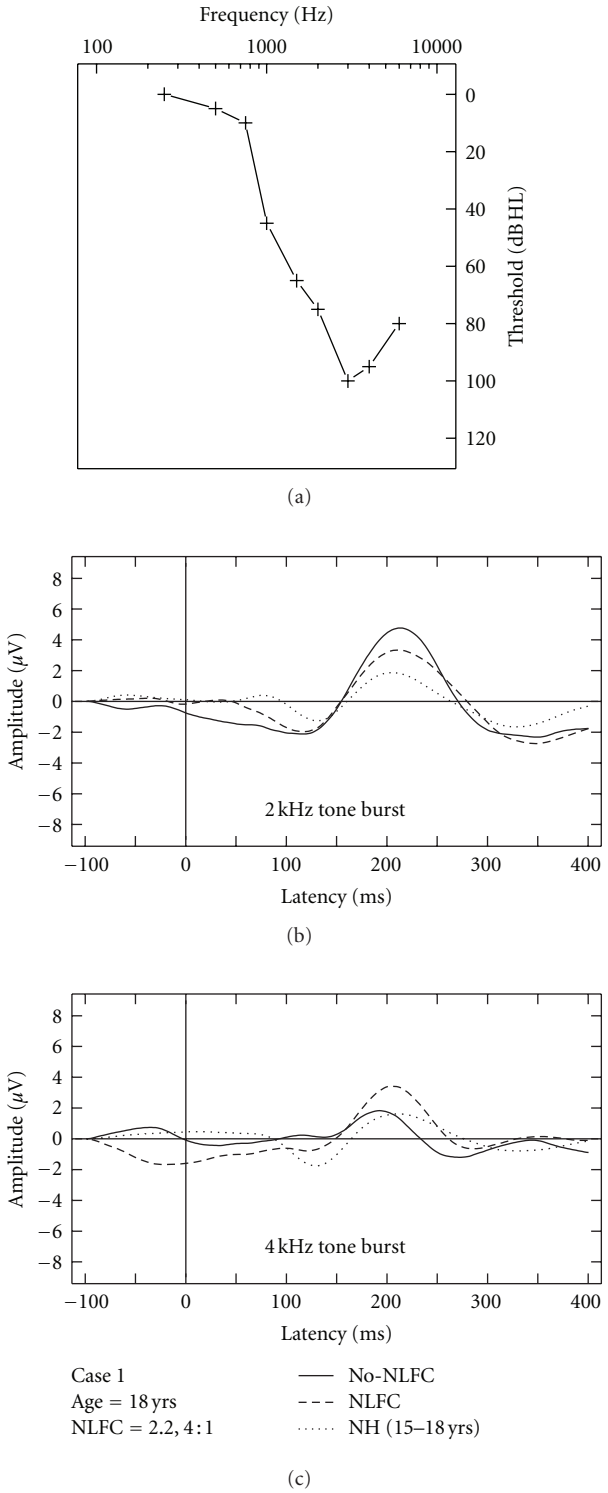

Figure 3: Case-specific grand mean CAEPs displayed across stimulus types and hearing aid conditions: no-NLFC (solid line), NLFC active (dashed line), and for each participant's age-matched NH group (dotted line). Participant age and prescribed NLFC settings (cut-off frequency, compression ratio) are indicated in the legend. The top left corner pane displays test ear hearing thresholds, suspected cochlear dead regions (DR), and responses beyond the audiometric test range.

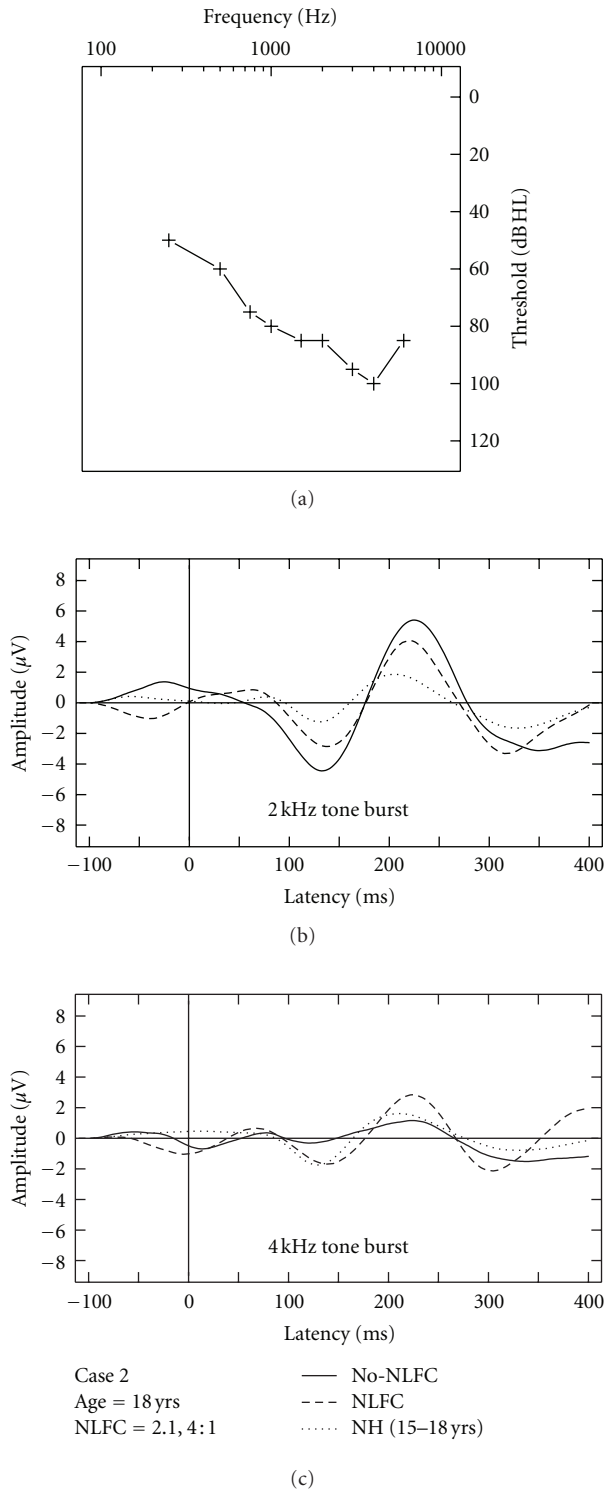

Figure 4: Case-specific grand mean CAEPs displayed across stimulus types and hearing aid conditions: no-NLFC (solid line), NLFC active (dashed line), and for each participant's age-matched NH group (dotted line). Participant age and prescribed NLFC settings (cut-off frequency, compression ratio) are indicated in the legend. The top left corner pane displays test ear hearing thresholds, suspected cochlear dead regions (DR), and responses beyond the audiometric test range.

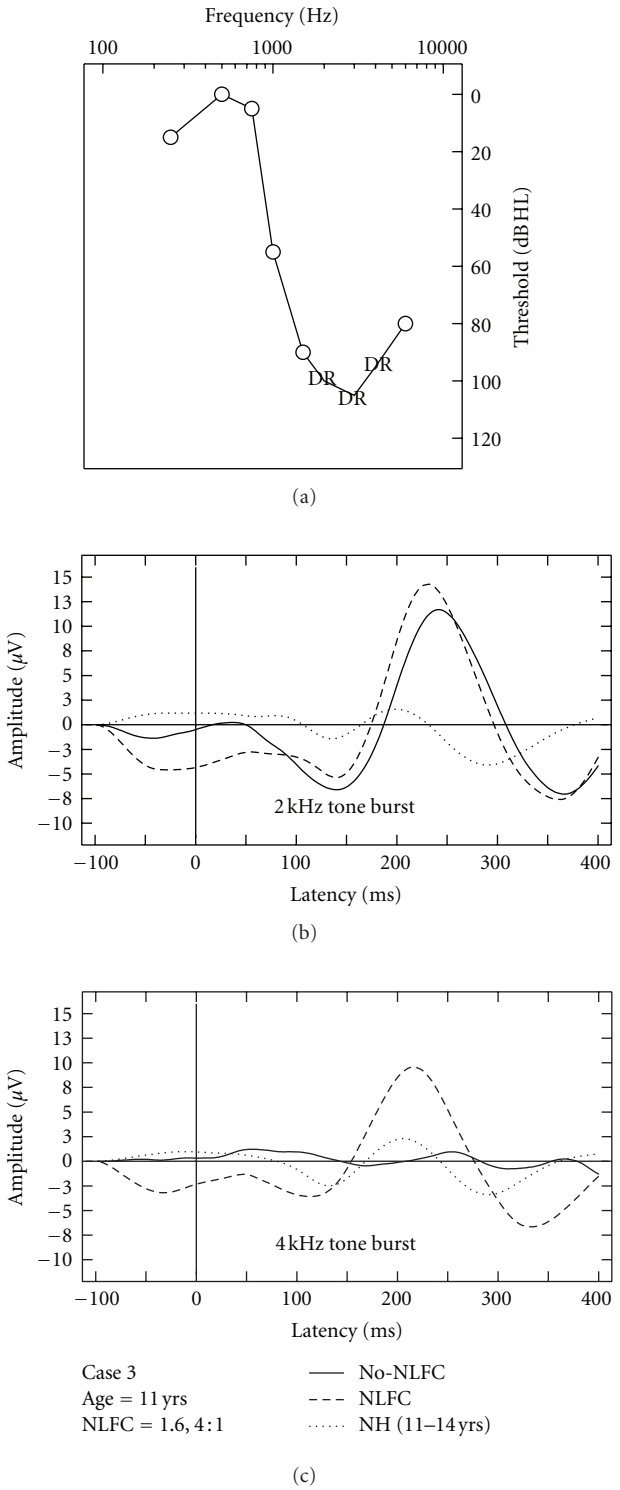

Figure 5: Case-specific grand mean CAEPs displayed across stimulus types and hearing aid conditions: no-NLFC (solid line), NLFC active (dashed line), and for each participant's age-matched NH group (dotted line). Participant age and prescribed NLFC settings (cut-off frequency, compression ratio) are indicated in the legend. The top left corner pane displays test ear hearing thresholds, suspected cochlear dead regions (DR), and responses beyond the audiometric test range.

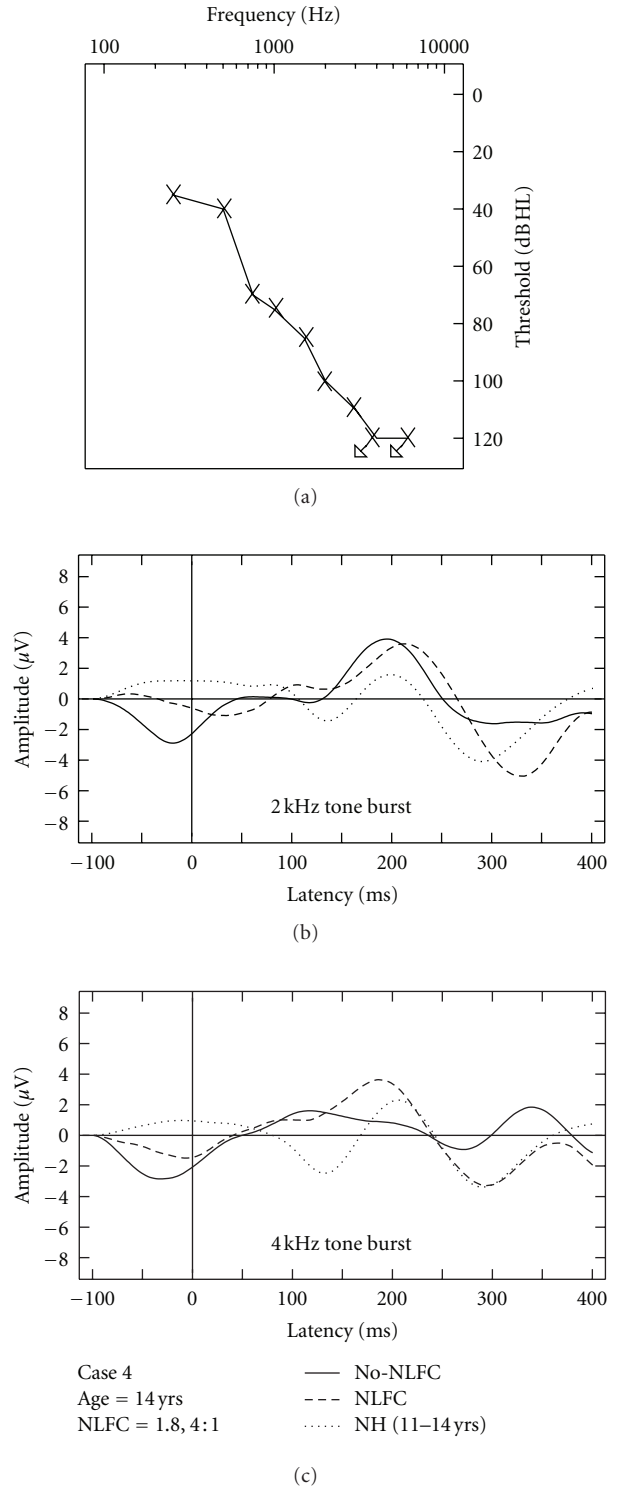

Figure 6: Case-specific grand mean CAEPs displayed across stimulus types and hearing aid conditions: no-NLFC (solid line), NLFC active (dashed line), and for each participant's age-matched NH group (dotted line). Participant age and prescribed NLFC settings (cut-off frequency, compression ratio) are indicated in the legend. The top left corner pane displays test ear hearing thresholds, suspected cochlear dead regions (DR), and responses beyond the audiometric test range.

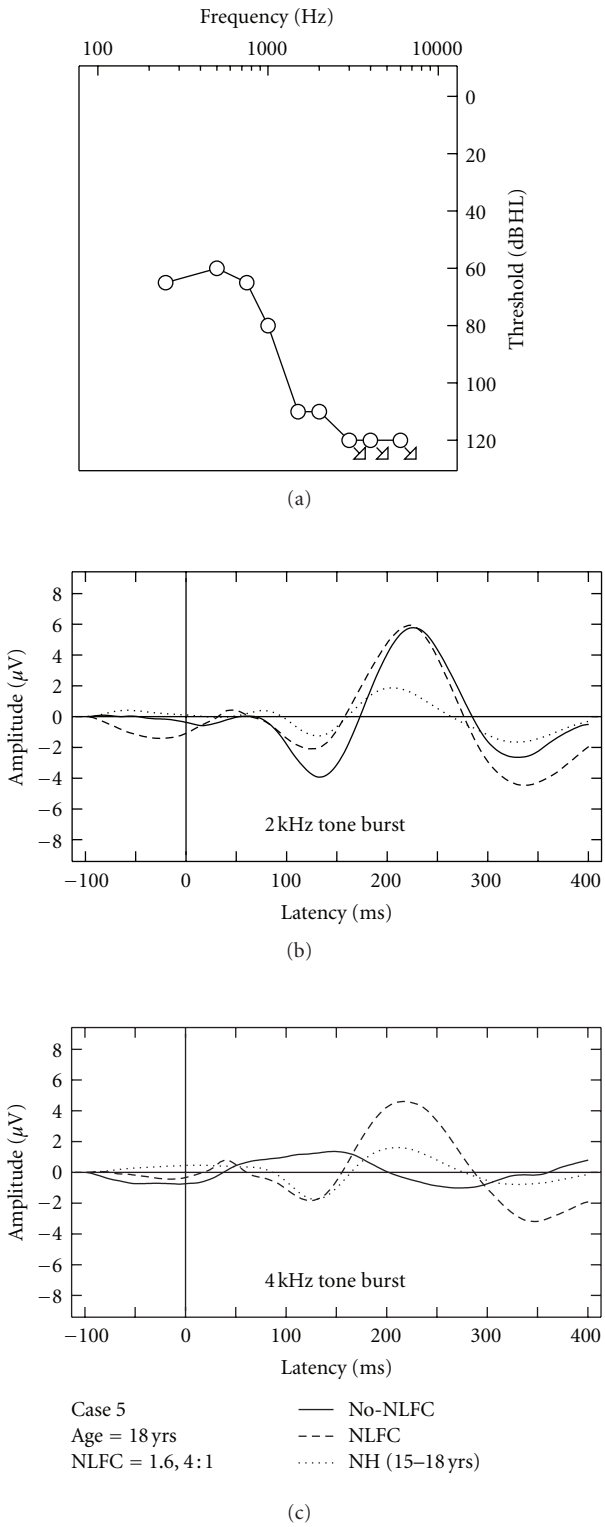

Figure 7: Case-specific grand mean CAEPs displayed across stimulus types and hearing aid conditions: no-NLFC (solid line), NLFC active (dashed line), and for each participant's age-matched NH group (dotted line). Participant age and prescribed NLFC settings (cut-off frequency, compression ratio) are indicated in the legend. The top left corner pane displays test ear hearing thresholds, suspected cochlear dead regions (DR), and responses beyond the audiometric test range.

3.3. CAEP Responses with NH Listeners. Data collected with NH listeners is displayed at the group level according to grand mean waveforms per stimulus (Figure 2), as well as in age separated groups (11–14 and 15–18 years). Of the 15 NH children tested, CAEP responses were judged to be present for most listeners. Absent waveforms were measured for one listener with the 4 kHz tone burst and for two listeners for the 2 kHz tone burst. Related analyses and figures include data from present responses only.

Developmental changes in the CAEP waveform have previously been reported over the age range that we studied [2, 24]. The effects of age subgroup on the observed CAEP for the NH children were submitted to a two-way analysis of variance (ANOVA) with age group (younger versus older children) as a between-subjects factor and the time window used in CAEP measurement (averaged sample points within each bin) as a repeated measures factor with 8 levels. Chosen bins corresponded to eight 50 msec intervals within the 400 msec window used in this study, equaling that used in previous evaluation of CAEPs [35, 42]. The amplitude of the CAEP waveforms elicited to the 2 kHz tone burst varied as little as 0.078 μV in the 4th bin (corresponding to latency region of 150−199 ms) to as large as 3.28 μV in the 6th bin (corresponding to latency region of 250–299 ms). On average, the 6th bin measured 0.38 μV in the older group compared to −2.9 μV in the younger group. The amplitude of the CAEP waveforms elicited to the 4 kHz tone burst varied as little as 0.003 μV in the 5th bin (corresponding to latency region of 200−249 ms) to as large as 3.03 μV in the 6th bin (corresponding to latency region of 250–299 ms). On average, the 6th bin measured 0.44 μV in the older group compared to −2.59 μV in the younger group. A significant interaction is reported between age group and time interval for the 2 kHz tone burst ($F = (3.07, 36.86) = 3.73$), $P < 0.05$) and for the 4 kHz tone burst ($F = (3.56, 39.33) = 4.11$), $P < 0.01$). Results suggest a need to separate mean CAEPs by age group when comparing NH listeners to their HI counterparts in the present study (Figure 2). Overall, NH waveforms can be described as displaying a negative N1 response, followed by clear P2 and N2 responses. No clear P1 response is present for either stimulus condition. The N2 response diminishes in the older age group (reduced amplitude), when compared to the younger age group.

3.4. Aided CAEP Responses and Discussion of HI Case Studies. Due to the small sample of HI listeners included in this study, aided CAEPs are displayed on a case-by-case basis using averaged waveforms across two repetitions, per stimulus and across aided conditions (Figures 3, 4, 5, 6, and 7). Grand mean NH CAEPs are displayed per stimulus (according to matched age group) for comparison purpose. Cases 1 through 5 are displayed in order of greatest to least hearing in the high frequencies, according to HF-PTA values. Case-by-case presentation enables interpretation of change in CAEP findings according to clinical or practical significance [43], considering individual factors such as hearing loss severity, configuration and the use of individualized hearing aid settings.

Table 1 summarizes waveform interpretation results for all cases, according to stimulus and aided treatment conditions. Present waveforms are reported for the 2 kHz tone burst, with and without NLFC enabled and across all participants. For the 4 kHz tone burst, present waveforms are reported with NLFC enabled across all participants and are judged to be absent for 4 out of the 5 participants, without NLFC enabled. Although waveform morphology appears to be variable across HI cases, some trends are present in the data. Waveforms are primarily dominated by large P2 responses across all aided cases. Many of the aided waveforms appear to have absent P1 responses. Responses measured with the 2 kHz tone burst appear to be larger in amplitude, when compared to those measured with the 4 kHz tone burst, across treatment conditions.

3.5. Comparison of Aided CAEP to Electroacoustic Verification Results. Electroacoustic results suggest that all tone bursts presented in the aided NLFC testing condition and 60 percent of the tone bursts presented in the aided no-NLFC testing condition were made audible by the hearing aid. The latter is a clinically common result for losses of this severity without the use of NLFC and is generally attributable to receiver limitations in modern hearing aids. A summary of these results can also be found in Table 1, along with a description of the corresponding CAEP measure (presence versus absence). For CAEPs judged as present, the estimated SLs ranged between −10 dB and 30.29 dB. For CAEPs judged as absent, the estimated SLs ranged between −18.6 dB and 6.19 dB. These ranges are overlapping but are also consistent with a pattern of higher sensation levels associated with present CAEPs.

4. Discussion and Conclusions

This study measured cortical auditory evoked potentials to tone burst stimuli using commercially available clinical testing equipment (Bio-logic Navigator Pro) in 15 NH (unaided) and 5 HI (aided) child listeners (ages 11–18 years). Firstly, this study investigated whether CAEPs could be reliably recorded and interpreted with NH and HI listeners using Bio-logic Navigator Pro testing equipment. Secondly, a case-series approach was used to examine the effects of NLFC hearing aid technology on CAEPs elicited by tone bursts. To facilitate evoked recordings, modifications to the standard testing parameters/equipment were made. In the aided testing conditions, equipment modifications allowed stimuli to be delivered directly to the hearing aid using a DAI hearing aid connector (coupled to an audioshoe). This eliminated the need to control for factors that could have affected the accuracy and reliability of sound-field presentation of stimuli. Due to software limitations, only averaged waveforms could be exported and analyzed offline.

Waveforms were subjectively interpreted by two experienced examiners using a rating protocol. Literature suggests that that statistical detection of CAEPs is consistent with those of experienced examiners [35, 42]. This study explored the relationship between an objective/statistical index of waveform replicability and subjective interpretation of presence versus absence of CAEPs. The ICC analysis provided a quantitative index of overall similarity of repeated waveforms. This preliminary analysis shows that the ICC, an index assessing test retest reliability, may have the potential to be used to aid subjective interpretation of CAEPs. However, in the absence of a validated pass/fail criterion for ICC values, the ICC data in the present paper merely serve to cross-validate the examiner ratings. Future studies could potentially pursue the development of the ICC as an objective aid to response scoring. Within the NH group, waveforms were judged to be present in most listeners. Waveforms were judged to be absent for one listener with the 4 kHz tone burst and for two listeners for the 2 kHz tone burst. It is unclear why for some NH listeners CAEPs were absent in this study. There have been a few studies that report large differences between CAEP and behavioral thresholds in some participants which imply that CAEPs are not always detected when tested at levels above the behavioral threshold in all participants. Discrepancies of >25 dB between CAEP threshold and behavioral threshold were noted in about 11% of adults tested at 0.5 kHz and 2 kHz [4] and discrepancy of >15 dB in 8.9% of adults at 2 kHz [44]. In HI children (aided and unaided), CAEPs were detected only about 77.8% at 20 dB SL or more [45]. Reasons for this discrepancy may include the specific corrections used to account for short- versus long-duration stimuli, the attention state of the participant and insufficient number of sweeps. It is possible that the results may have been different if recorded using equipment with multiple testing channels and capable of completing online calculations of noise levels. This would allow rejection of eye blinks and other potential sources of noise, as well as monitoring of the level of arousal of the participants. Although none of the participants fell asleep during this study, it is possible that some experienced a reduced level of arousal and consequently had smaller or absent responses. Taken together, these findings from the literature and those from the present student may indicate that interpretation of absent CAEP responses is complex because absent responses may occur even when stimuli are likely suprathreshold. Caution may be indicated when interpreting absent responses in clinical or research contexts.

Waveforms recorded with the NH listeners were found to differ across the age range included in this study. Displayed grand averaged waveforms, separated according to age, suggest that the main source of variability lies within the 250–350 msec range of the response window, with larger N2 amplitudes observed with the younger NH group. These findings are consistent with those reported for a group of normally hearing children listening to click trains: changes in the latency and amplitude values associated with the N2 response were observed up to approximately age 17 [24]. Results from this study differ from the above-mentioned study when considering the presence of observable peaks in the earlier part of the evoked response (specifically the P1 peak).

Considering the HI listeners, a unique pattern of results was observed across the different cases. The differences observed in waveform morphology may relate to factors

including the age range of the participants included in the study (and accompanying developmental effects on waveform morphology) and differences in hearing loss configuration across participants and/or chosen hearing aid settings. In summary, CAEPs were sensitive to changes in audibility across NLFC and no-NLFC hearing aid conditions. Responses appear to be larger, on average, for the 2 kHz tone burst, compared to the 4 kHz tone burst. This is consistent with findings of earlier studies that report an inverse relationship of CAEP amplitude and frequency of tone burst stimuli. The peak-to-peak amplitude was found to decrease as the frequency of the tone bursts increased [46, 47].

When comparing NH grand mean waveforms (separated by age group) to the individual averaged waveforms for the HI cases, aided CAEPs tend to be larger in amplitude than the unaided CAEPs measured with the NH group. However, this appears to be the only uniform pattern of results between CAEPs measured with NH and HI participants. Although the ages of the participants were relatively well matched across NH and HI children, the groups were tested with different transducers (broadband insert earphones versus direct audio input to a well-fitted hearing aid). Also, the sensation level of the stimuli was greater for the NH participants than for the HI participants. The HI children in this study had substantial hearing losses, likely accompanied by reduced dynamic ranges of hearing. If we had presented at matched sensation levels across the groups, the stimuli would have been either too loud for the HI participants or very soft for the NH participants. Instead, we presented at middynamic range for all listeners, which in turn results in high sensation levels for NH and low sensation levels for HI participants, respectively. Future studies could potentially consider presenting stimuli at equal loudness by incorporating a loudness model that accounts for the effects of sensorineural hearing loss (e.g., [48]). This may also provide insight into the differences in the relationship between presence of CAEP and SL between the two groups.

Variable CAEP waveform morphology can be observed across the different HI cases when considering gross amplitude and latency differences. For example, present waveforms in Case 3 are larger in amplitude when compared to all other cases. This is also the only case where a suspected cochlear dead region was measured. It is possible that the large response measured with the 2 kHz tone burst reflects overrepresentation of neurons in the cortical regions bordering the cochlear dead region. These findings are closely aligned with those reported in animal model research, describing the use of auditory evoked potential measures to quantify cortical over-representation (e.g., enhanced amplitude responses) corresponding to the frequency region at the cut-off slope of the audiogram in cats [49, 50]. Further research using aided CAEP measures in listeners with steeply sloping audiograms is needed to confirm this speculation.

All HI participants were long-time hearing aid users who had received a period of time to acclimatize to NLFC prior to beginning testing. The length of time allotted for acclimatization for the purpose of this study is consistent with that reported in the literature (i.e., 12–18 weeks) investigating acclimatization effects in aided speech perception [51, 52]. Presentation levels of tone bursts with higher SLs elicited CAEPs more often than those with lower sensation levels. This was true both with and without the NLFC hearing aid condition. In the cases where NLFC technology improved audibility for a given stimulus, (4 out of the 5 cases, measured with the 4 kHz tone burst), detection of the cortical response also improved. These findings were highly consistent across listeners and are not surprising, given previous work suggesting that the use of hearing aids can improve audibility and thereby increase the probability of eliciting CAEPs the ability to detect CAEPs [7, 8]. A strong relationship between the presence of repeatable aided CAEPs in children, and measures of audibility using electroacoustic verification were reported in this study. This suggests that recording CAEPs to tone burst stimuli, and at a suprathreshold level, can provide physiological evidence that these stimuli have been detected at the level of the auditory cortex with hearing aids fitted. This study did not evaluate specific latency/amplitude differences in the aided P1-N1-P2 complex; rather it generally looked at the presence/absence of a response as indicated by subjective waveform interpretation. Although this study is the first to compare aided CAEPs to electroacoustic verification results, others have compared the relationship between CAEPs and functional outcomes for aided infants [53, 54]. Functional measures of hearing aid performance may provide an important cross-check against aided CAEPs, because both measurement types offer both strengths and limitations.

In summary, this study demonstrated that CAEPs can be recorded with clinical testing equipment in NH and HI children. In the aided testing condition, tone burst stimuli were directly presented to the hearing aid via a DAI connector. Repeatable present waveforms were measured in most participants, although were missing in a small number of NH listeners. Present waveforms in the HI children may have been associated with higher sensation levels. The younger NH children were found to have significantly different responses than the older NH children, with grand averaged waveforms differing mainly between 250 and 350 msec in the response window (the N2 response). For most of the HI cases included in this study, frequency compression hearing aid technology improved audibility of the 4 kHz tone burst; this translated into improved detection of CAEP responses. These findings suggest that the CAEP may be sensitive to the effects of frequency compression signal processing and that frequency compression may have augmented audibility of high-frequency tone bursts on a case-by-case basis. This contribution to the literature provides insight into possible strengths (sensitivity to changes within the aided condition) and limitations (present responses were not always measureable in normal listeners) of CAEP. As this study evaluated *gross* changes in high-frequency audibility, the resulting effects on the CAEP response could be observed as either present or absent. However, further research is needed to assess the effects of hearing aid fine tuning (e.g., changes to hearing aid gain and frequency shaping), or other aspects of hearing aid signal processing, on CAEPs. In addition, more research is needed to validate aided findings reported in

this study across a larger group of HI participants including infants and with speech-based stimuli.

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Research Article

Slow Cortical Potentials and Amplification—Part I: N1-P2 Measures

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Slow cortical potentials (SCPs) are currently of great interest in the hearing aid fitting process for infants; however, there is conflicting evidence in the literature concerning the use of SCPs for this purpose. The current study investigated SCP amplitudes and latencies in young normal-hearing listeners 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, and DigitalB) with two gain settings (20 and 40 dB). Results showed that SCP 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 the unaided across conditions. SCP latencies in analog conditions were not significantly different from latencies in the unaided conditions; however, both digital hearing aids resulted in significantly delayed SCP latencies. The results of the current study (as well as several previous studies) indicate that the SCP may not accurately reflect the amplified stimulus expected from the prescribed hearing aids. Thus, "aided-SCP" results must be interpreted with caution, and more research is required concerning possible clinical use of this technique.

1. Introduction

With the advent of Universal Newborn Hearing Screening programs, it has become increasingly common for infants to be fitted with hearing aids by six months of age, which requires that reliable methods are in place for fitting hearing aids in this young population [1–4]. There are two general categories of test procedures used in the hearing aid fitting process: (i) behavioural measures, which require active participation by the patient (e.g., pure-tone thresholds, 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, especially of threshold). In infants, behavioural measures are much more limited: behavioural thresholds are not usually available before the age of 6 months (and often later), speech testing is unavailable, and subjective questionnaires are limited to caregiver observation of behaviours [2]. Thus, there is greater reliance on objective measures such as auditory evoked potentials (AEPs) [1, 5, 6] rather than the subjective measures possible with most adults [2, 7, 8]. AEPs have been considered for use within the hearing aid fitting process for at least two different purposes: (i) to determine whether aided AEPs can be obtained at input levels where unaided AEPs were absent and/or (ii) to determine whether changes in hearing aid settings (e.g., gain) and/or stimuli can be measured using AEPs. Several AEPs have been assessed as potential objective measures, including the auditory brainstem response (ABR), the auditory steady-state response (ASSR), and the slow cortical potential (SCP).

Previous ABR research revealed that the click and brief-tone stimuli required for ABR testing are too short to activate the compression processing and steady-state response of the hearing aid; thus, the ABR has been deemed not to be suitable for assessment of responses to hearing-aid-processed stimuli [9–11]. In light of these findings, subsequent researchers have considered the 80-Hz ASSR as a solution to the stimulus problem (e.g., [12–14]). For the purpose of hearing aid

measures, 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 [13]. However, recent research suggests 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 [15]. Thus, the 80-Hz ASSR may be subject to the same limitations as the ABR [16] and, like the ABR, shows poorer frequency specificity than indicated by the acoustics [17, 18]. 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.

Slow cortical potentials have an advantage over brainstem AEPs because they originate in the auditory cortex [19–21] and can be elicited by a variety of signals, including tonal stimuli longer than those used for the ABR as well as speech stimuli (for reviews, see: [22, 23]). As a result, the SCP is now the AEP of greatest interest for use with hearing aids [24–30]. 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 fitting process using a variety of stimuli, including tonal and complex ones [27–29, 31–33]. Findings such as these have led a group of researchers at the National Acoustic Laboratories to develop a new device currently being marketed as a hearing aid validation tool for infants and children [34, 35].

Despite the current upsurge of interest in the use of cortical SCPs with hearing aids in infants and children, there are several recent studies which indicate that concerns exist regarding the use of SCPs for this purpose. For example, some recent studies on SCPs in subjects with normal hearing using speech and tonal stimuli have found that the addition of hearing aid gain does not lead to the expected increase in N1-P2 amplitude that occurs in response to higher stimulus levels, such that there were either no significant differences between unaided and aided N1-P2 amplitudes [25, 30] or *smaller* aided N1-P2 amplitudes compared to unaided [24]. Billings and colleagues [24, 36] have examined the effect of SNR (i.e., stimulus amplitude to background noise ratio) on aided and unaided SCPs and, as a result, have concluded that the lack of an amplification effect on the SCP seen in their earlier work [25, 30] was due, at least in part, to the SNR being similar across aided and unaided conditions (i.e., in addition to amplifying the stimulus, the hearing aid also introduced higher noise levels).

The stimuli used by Billings and colleagues [24, 25, 30, 36] were atypical for SCP stimuli in that they had a more rapid rise time (7.5 ms) than is optimal for eliciting the SCP, and they were much longer in duration (756 ms) than is known to be reflected by the SCP. Research has shown that N1 shows little effect of stimulus changes beyond the first 20–40 ms [37–41]. Also, rise times between 20 and 30 ms result in the largest N1 amplitudes with either no further change, or decrease, in amplitude with longer rise times [40, 41]. Perhaps different results may have been obtained using stimuli with more typical rise/fall times and overall durations, such as 20 ms rise time and 60 ms total duration [22, 23, 42–44], although a recent study by Easwar and colleagues [45] suggests little difference in SCPs to hearing-aid-processed stimuli with 7.5 ms versus 20 ms rise times.

Due to conflicting evidence regarding the use of SCPs for hearing aid measurements, the primary aims of the current study were: (i) to determine the effects of hearing aid gain on the SCP and, specifically, whether different hearing aid gains are accurately reflected by the SCP, and (ii) to assess whether results are similar for different hearing aids set with the same gain characteristics. Of particular interest was whether there would be a difference between unaided and aided response amplitudes and latencies when hearing aids were set for 20 or 40 dB of gain and whether unaided and aided response amplitudes and latencies would be comparable for equivalent nominal output levels.

2. Methods

2.1. Subjects. Thirteen normal-hearing subjects participated in this study (mean age: 25 ± 5.5 years; 5 females). 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 and for normal middle/outer ear function by immittance audiometry. 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 [46]. Normal tympanograms were defined by a single-peak static admittance between ±50 daPa in response to a 226-Hz probe tone. 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.

2.2. Recording. 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 Synamps 2 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 correction across the total sweep duration, artifact rejection (±100 μV in any of the channels, and ±75 μV 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 (thereby ensuring any changes in subject state or noisiness were equivalent across replications). Averages were baseline-corrected using the prestimulus interval.

2.3. Hearing Aids. The same three behind-the-ear stock hearing aids, coupled with Comply snap tip 9-mm foam earmolds inserted to be flush with the ear canal entrance, were used for each participant: (i) Oticon E27 (Analog), (ii) Phonak Savia 211 dSZ (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 hearing aid was selected to account for possible discrepancies between results for digital hearing aids and those in the published literature which primarily used analog hearing aids [24, 25, 28, 30, 31, 33].

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. [25], 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 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. Maximum output was set to the highest level. 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 equalled 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 remeasured in follow-up sessions.

REIG was determined for each individual participant 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 (i.e., tonal, rather than speech-like) 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; these were the same input levels and gains used in the SCP testing. For all participants, all hearing aids matched target gain within 1 dB for all input levels used in SCP testing, except 30 dB SPL, which could not be verified as it was below the levels available in the hearing aid test system. These measures provided further verification that all three hearing aids were providing linear processing and that all measures were below the maximum output of the hearing aid.

Electroacoustic measures of processing delay were conducted on the Fonix 7000 test system. The delays were 0.4 ms for the Analog aid, 6.8 ms for DigitalA, and 2.3 ms for DigitalB.

2.4. SCP Stimuli. The stimulus used was a 1000-Hz tonal stimulus of a 60 ms total duration (including a 20 ms rise/fall time). 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 [37–41]. Stimuli were presented with offset-to-onset interstimulus intervals (ISIs) of 940 ms. Stimuli generated by Neuroscan's Stim 2 software were 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 and 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 signal to 90 dB SPL, after hearing aid gain. For the 40 dB gain condition, the maximum stimulus intensity was 50 dB SPL, again to limit the maximum signal to 90 dB SPL. 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.

2.5. Procedure. Participation in this study involved two sessions on separate days, each 2 -3 hours in length (i.e., 4–6 hours total). Subjects were screened for normal hearing and for normal outer- and middle-ear function. 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 octave band noise levels in the sound-attenuated booth at 500, 1000, 2000, and 4000 Hz were 12, 10, 10, and 12 dB SPL, respectively. There were 18 test conditions, and the presentation order for each subject was randomly assigned prior to the test dates. 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. This position was chosen because (i) calibration measures showed that it was the "flattest" spot in the soundfield, making it appropriate for the substitution method of calibration employed in these measures, and (ii) it was easy to monitor via closed-circuit television whether the participant's head position had moved substantially out of the calibrated spot in the soundfield.

2.6. 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 [28, 47–50]. 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., [28, 47–50]). Latencies were not estimated for no-response results. Due to absent responses for 5 out of 13 subjects in some of the 30 dB SPL conditions, latency results for this input level were excluded from statistical analyses.

2.6.1. 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 (Analog, 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, the main effects and interactions were considered significant if $P < 0.05$. Huyn-Feldt epsilon (ε) 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 $P < 0.05$.

3. Results

Grand mean waveforms for unaided compared with 20 and 40 dB of hearing aid gain for three hearing aids (Analog, DigitalA, and Digital B) across three input levels (30, 50, and 70 dB SPL) are shown in Figure 1. Mean (and standard deviation) amplitude and latency results for each condition are presented in Figure 2.

3.1. Unaided Compared with 20 dB Gain Condition. In Figures 1 and 2, it is apparent that any differences between unaided and aided response amplitudes in the 20 dB gain condition are quite small, with some aided results even appearing smaller in amplitude than the unaided condition. Indeed, results of the ANOVA revealed a significant interaction between hearing aid type and input level ($F(6, 72) = 2.48$, $\varepsilon = 0.78$, $P = 0.046$); 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. 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 ($F(3,36) = 6.99$, $\varepsilon = 0.95$, $P < 0.001$) and for input level ($F(2, 24) = 74.97$, $\varepsilon = 1.00$, $P < 0.001$). 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 shows that N1 latencies in both digital hearing aid conditions appear longer by as much as 10–20 ms, 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

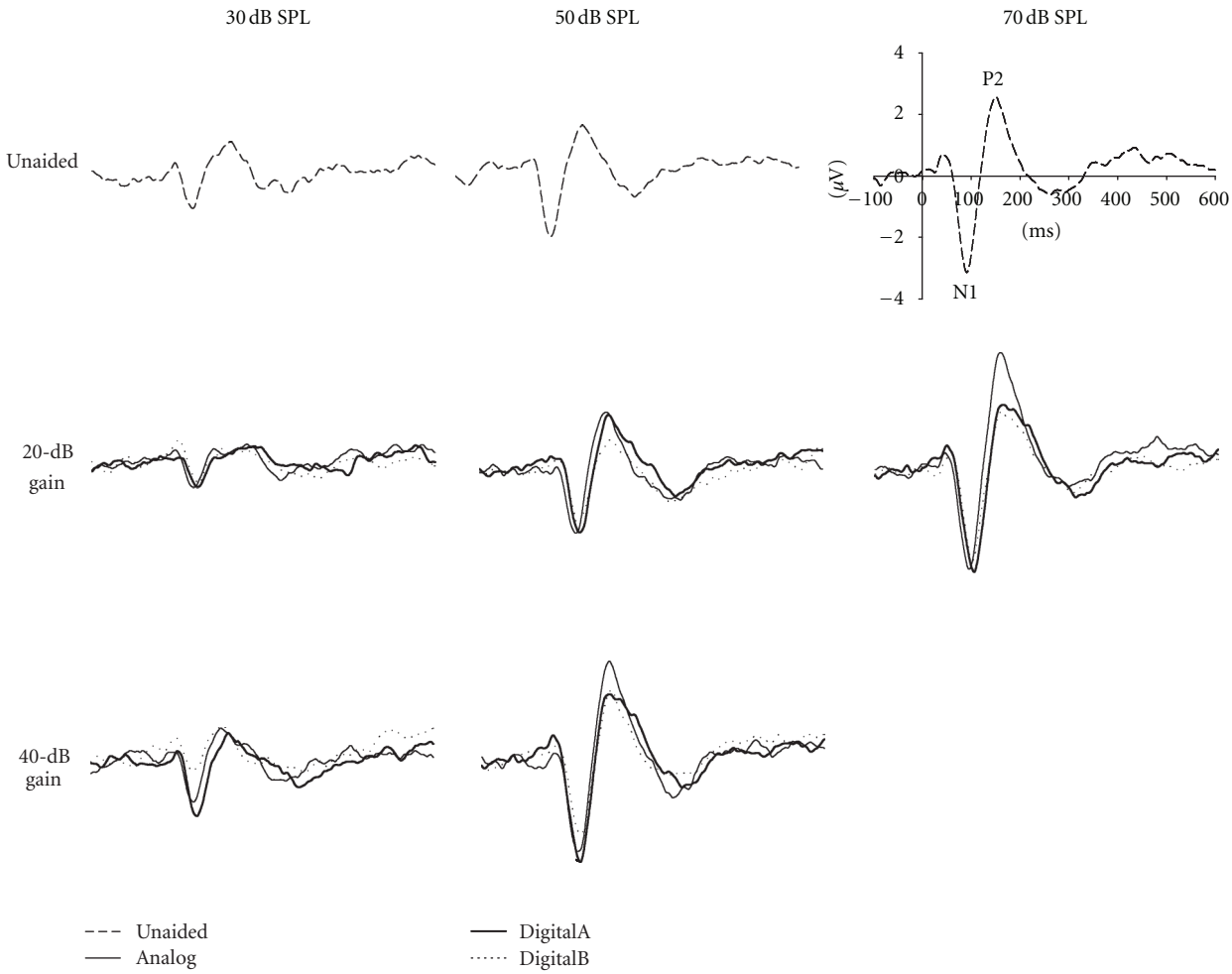

Figure 1: Grand mean ($N = 13$) waveforms from electrode Cz for unaided and aided conditions (Analog, DigitalA, and DigitalB) with two gain settings (20 and 40 dB) at three input levels (30, 50, and 70 dB SPL).

effect of input level ($F(1, 12) = 47.88$, $\varepsilon = 1.00$, $P < 0.001$), such that response latencies were longer for the 50 compared with the 70 dB SPL input level. There was a nonsignificant trend (hearing aid type X input level interaction: $F(3, 36) = 2.42$, $\varepsilon = 0.817$, $P = 0.096$), such that latencies were the same in unaided and Analog hearing aid conditions for both input levels, but latencies for both digital hearing aids were longer compared with Analog and unaided conditions.

3.2. Unaided Compared with 40 dB Gain Condition. In Figures 1 and 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 ($F(3,36) = 7.58$, $\varepsilon = 1.00$, $P < 0.001$) and input level ($F(1, 12) = 115.26$, $\varepsilon = 1.00$, $P < 0.001$), as well as a significant interaction between hearing aid type and input level ($F(3, 36) = 5.42$, $\varepsilon = 0.96$, $P = 0.004$). Post hoc analysis showed that N1-P2 amplitudes were significantly smaller in DigitalB aided conditions compared with Analog and DigitalA. Additionally, post hoc analysis of the 40 dB gain results 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 30 dB SPL 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 again, by as much as 10–15 ms, 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 were not included in the statistical analysis due to missing data. The ANOVA results revealed a statistically significant difference between hearing aid conditions ($F(3, 36) = 5.57$, $\varepsilon = 1.00$, $P = 0.003$]. As is evident in Figure 2, there was a trend for both digital hearing aid conditions to be delayed compared with unaided and analog hearing aid conditions;

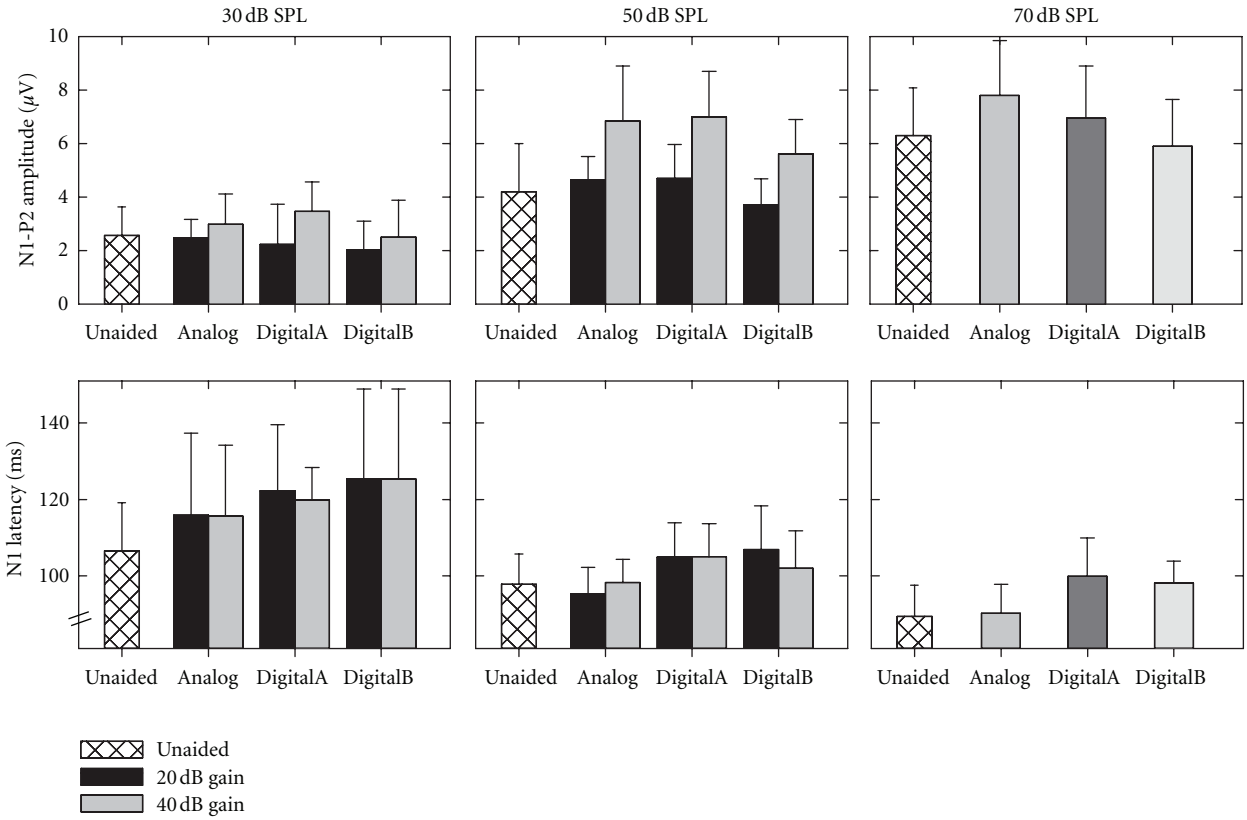

Figure 2: Mean and SD amplitude and latency data for unaided and aided conditions (Analog, DigitalA, and DigitalB) with two gain settings (20 and 40 dB) at three input levels (30, 50 and 70 dB SPL). Sample size is $N = 13$ for all conditions, except for the following latency results: 30 dB SPL/DigitalA/20 dB gain ($N = 10$); 30 dB SPL/DigitalB/20 dB gain ($N = 11$); 30 dB SPL/DigitalB/40 dB gain ($N = 12$).

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.

3.3. 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 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 ($F(1, 12) = 73.18$, $\varepsilon = 1.00$, $P < 0.001$) and for input level ($F(1, 12) = 118.13$, $\varepsilon = 1.00$, $P < 0.001$), as well as a significant interaction between gain setting and input level ($F(1, 12) = 7.32$, $\varepsilon = 1.00$, $P = 0.019$). 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 ($F(2, 24) = 6.01$, $\varepsilon = 1.00$, $P = 0.008$), 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 ($F(2, 24) = 1.02$, $\varepsilon = 1.00$, $P = 0.37$), hearing aid type and input level ($F(2, 24) = 1.73$, $\varepsilon = 1.00$, $P = 0.20$), or hearing aid type by gain setting by input level ($F(2, 24) = 0.41$, $\varepsilon = 1.00$, $P = 0.67$). These results are consistent with findings reported previously 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 show that, once again, there was an obvious difference between N1 latencies across digital and analog aided conditions, such that N1 latencies were longer in the digital hearing aid conditions (particularly for the 20 dB gain setting); however, there did 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 ($F(2, 24) = 13.66$, $\varepsilon = 1.00$, $P < 0.001$), 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 ($F(1, 12) = 0.46$, $\varepsilon = 1.00$, $P = 0.51$), indicating that latencies for the 40 dB gain setting were not significantly shorter than latencies for the 20 dB gain setting. There was a nonsignificant 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 ($F(2, 24) = 2.92$, $\varepsilon = 1.00$, $P = 0.07$).

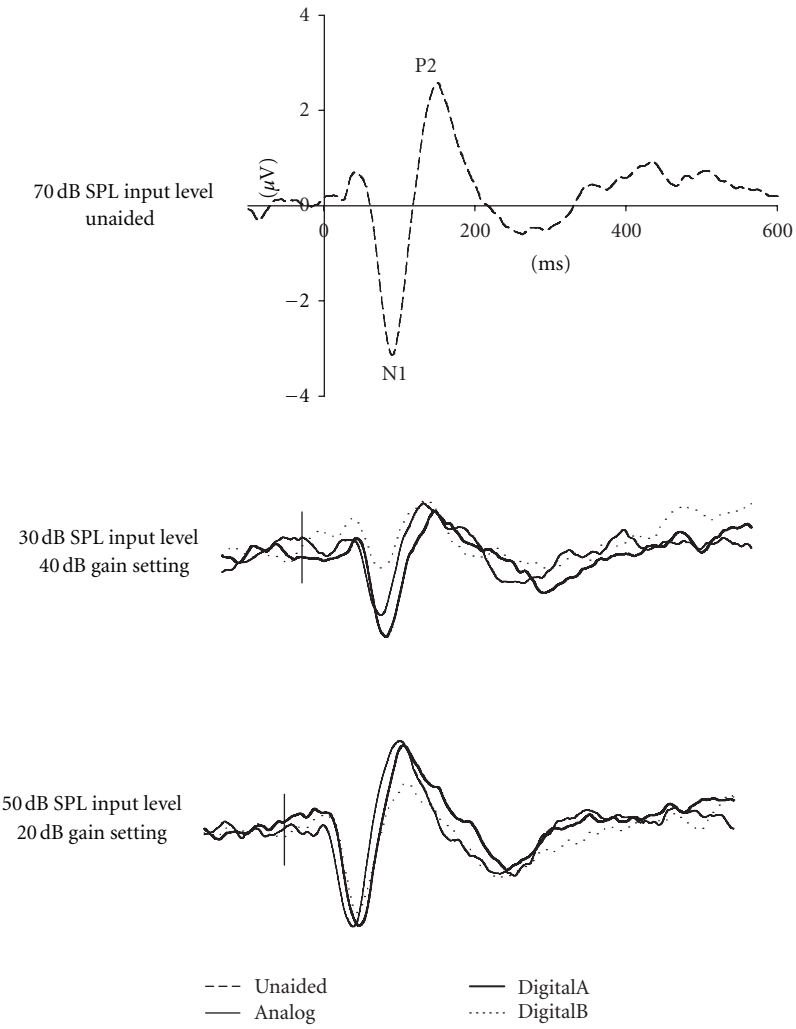

Figure 3: Grand mean ($N = 13$) waveforms for 70 dB SPL equivalent nominal output for unaided and aided conditions (Analog, DigitalA, and DigitalB).

3.4. 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 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 ($F(6, 72) = 14.75$, $\varepsilon = 0.89$, $P < 0.001$). Post hoc analysis showed that N1 response ampli–tudes were significantly smaller for all hearing aid conditions (i.e., 30 dB SPL input level and 40 dB hearing aid gain; 50 dB SPL input level and 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 and 20 dB gain conditions were significantly larger than those obtained with the 30 dB SPL and 40 dB gain conditions for all hearing aid conditions except for DigitalB.

4. Discussion

The current study, carried out in participants with normal hearing, assessed the effects of processing of stimuli by three different hearing aids and two gain settings on SCP amplitudes and latencies. This differs from many previous studies in participants with hearing loss, where aided SCPs were obtained at levels where unaided stimuli were inaudible and-unaided SCPs were absent. Results of the present study clearly demonstrate that, in many instances, introducing hearing aid gain had little or no effect on SCP N1-P2 measures, and, in some comparisons, SCP results worsened (i.e., lower amplitude and/or later latency). Further, significant differences were found between different hearing aids.

These findings are despite the fact that verification of REIG measures using conventional hearing aid test procedures in the present study confirmed all three hearing aids provided 20 and 40 dB of gain at input levels used in testing for all subjects.

There has been much interest in using the slow cortical potential as an objective hearing aid measure, especially recently; however, there is clearly conflicting evidence in the literature regarding the accuracy of the SCP for this purpose. Most research indicates that SCPs can indeed be recorded in response to hearing-aid-processed stimuli [24–33]. Studies involving individuals with hearing loss have shown increased SCP response presence, increased SCP amplitudes, and, in some studies, decreased SCP latencies between aided and unaided conditions [27–29, 31–33, 51]. However, these studies did not systematically assess effects of different hearing aid parameters (e.g., gain, frequency response, or hearing aid type) or of input intensity.

Some studies suggest that different cortical responses for different speech stimuli are maintained in aided conditions [27–30]. However, results of several recent studies indicate that the SCP N1-P2 cannot provide a reliable method to objectively demonstrate in individuals that the brain has discriminated between different speech stimuli [51–53]. Because of this, the commercial release of the HEARLab system for recording and analysis of aided and unaided SCPs in infants does not provide for comparisons of waveform shapes in response to different stimuli for the purpose of inferring perception of different (speech) stimuli ([51]; H. Dillon, personal communication).

Several studies in individuals with normal hearing, including the present study, have assessed several aspects of input stimuli and/or hearing aid parameters, including input intensity ((current study; [24, 25]) that gain (current study; [24]), and hearing aid type (current study). All of these studies indicate the aided SCP does not accurately reflect changes in these parameters.

The current study is the first study to assess the effects of amplification by different hearing aids on the SCP. Specifically, the current study investigated SCP responses to stimuli processed by an analog hearing aid, as well as two digital hearing aids from two different manufacturers, all set to provide the same amount of gain. To simplify comparisons, all three hearing aids were set to linear processing, with additional processing features (e.g., feedback reduction; noise reduction) tuned off. As noted previously, despite older technology, the analog hearing aid was included because almost all preceding reports of the SCP in response to hearing-aid-processed stimuli involved participants using analog hearing aids. Importantly, the results of the current study indicate significantly (and clearly) different effects on the SCP for the three different hearing aids, with the DigitalB hearing aid condition showing significantly smaller amplitudes than the other hearing aid conditions, both digital hearing aid conditions showing longer (albeit not quite significant) latencies, and the Analog hearing aid showing larger SCP amplitudes than the digital hearing aid conditions.

Prior to the current study, two studies have reported no significant differences between unaided SCP amplitudes and SCP amplitudes measured while subjects wore hearing aids providing 20 dB of gain [25, 30]. In the current study, none of the hearing aids resulted in significantly larger SCP amplitudes for 30 or 50 dB SPL stimuli and 20 dB gain. Small but significant amplitude increases were seen only for the Analog hearing aid for 70 dB SPL stimuli. Indeed, the DigitalB hearing aid resulted in significantly *smaller* SCP amplitudes (at 50 and 70 dB SPL) compared to unaided results. These results cannot be explained by SCP N1-P2 amplitude reaching a ceiling at higher intensities or the hearing aid reaching its maximum output, because discrepancies were seen at all intensities. Indeed, although previous research indicates increases in SCP amplitude and decreases in latency asymptote at higher intensities, for tonal stimuli these plateaus do not occur until at least 80–90 dB SPL (e.g., [54–58]), although this may differ for individuals with hearing loss [54].

Only two studies, the current study and the recent study by Billings and colleagues [24], have assessed changes in the aided SCP occurring with changes in hearing aid settings using SCPs. Both studies indicate that SCP amplitudes do not accurately reflect the expected hearing gains. In the current study, increasing the gain to 40 dB resulted in significantly larger SCP amplitudes for all three hearing aids, but only at 50 and 70 dB SPL, and not at 30 dB SPL. Importantly, although larger in amplitude for 40 dB gain, the hearing aid conditions resulted in smaller responses compared to the unaided condition at an equivalent nominal output level. Our following study [59] provides some explanations for the results of the current study.

Additionally, N1 latencies were longer in conditions involving digital hearing aids compared with Analog and unaided conditions; there were no significant differences between N1 latencies for unaided and Analog aided conditions. The differences seen, as much as 20 ms, are much longer than the 6.8 ms (DigitalA) and 2.3 ms (DigitalB) hearing aid delays indicated by the standard electroacoustic measures conducted using the Fonix 7000 System. We discuss this further in our following study [59].

Concern has been raised about conclusions drawn from studies in individuals with normal hearing of aided versus unaided SCPs (H. Dillon, personal communication). It is possible that the internal noise of hearing aids could affect SCP results in subjects with normal hearing but would be below threshold in subjects with hearing loss. Alternatively, the hearing aids amplified ambient noise in the sound booth, which could possibly have detrimental effects on SCPs especially in normal listeners. Indeed, recent studies of the SCP indicate large effects of noise on the SCP, such that signal-to-noise ratio (SNR) may be a more determining factor than the absolute stimulus level [24, 36]. We consider SNR in our follow-up study [24]. However, as problems were seen also at the higher input levels (50 and 70 dB SPL), as well as at the lowest level (30 dB SPL), SNR cannot explain all of the current study's results. Further, if studies of aided SCPs are only suited for individuals with hearing loss, it is not clear where the hearing level dividing line should be, given that individuals with regions of mild hearing loss are also fitted with hearing aids. In our view, it is important to better

understand how SCP results reflect hearing-aid-processed stimuli in individuals with normal hearing, prior to assessing individuals with hearing loss.

The implications of the current study, as well as other studies, are that SCP results do not always reflect the expected (and electroacoustically verified) gain of hearing aids, and that different hearing aids with similar electroacoustic parameters (i.e., gain) may result in substantially different SCP results. Thus, it would be possible that an appropriately fitted hearing aid (for degree and configuration of the hearing loss) in a client may either show no SCP response or no improvement from the unaided SCP response, even though behaviourally the client shows clear improvement. Similarly, changing hearing aid setting may not change the SCP. Finally, changing the type or brand of hearing aid may result in substantial SCP differences, even though standard electroacoustic measures indicate no change.

5. Conclusions

There is much interest in the audiology community in the use of the SCP in the hearing-aid fitting process for infants with hearing loss. The results of the current study (as well as several previous studies) indicate the SCP may not accurately reflect the gain expected from the prescribed hearing aids. Thus, aided-SCP results must be cautiously interpreted when used clinically. A "present" SCP may be interpreted as indicating that the individual's auditory cortex has responded to the hearing-aid-processed stimuli; however, as Dillon and colleagues have recently cautioned, "this does not necessarily indicate that the hearing aids are providing effective amplification" [60]. Importantly, an "absent" SCP in the aided condition does not always indicate that stimuli are inaudible to the individual, as several studies have indicated absent SCPs to audible stimuli as much as 20–30 dB above their behavioural threshold (e.g., [23, 34, 35, 44, 60]).

Clearly, more research concerning the "aided-SCP" technique is required. Importantly, among the needed studies are assessments of the effects of hearing aid processing on the acoustics of the stimuli used to elicit the SCP. Until now, it has been assumed that the longer durations of SCP stimuli allow for adequate processing by hearing aids. SCP stimuli have been contrasted to the much shorter click and brief-tone stimuli required for ABR testing, which were too brief for hearing aid processing and thus did not result in valid measurements of hearing aid gain or output [9–11]. The current study as well as previous studies [24, 25, 30], however, indicate the SCP also does not accurately reflect hearing aid characteristics. In our following study [59], we investigate this further, assessing the acoustic effects of the hearing aid processing on test signals used to elicit the SCP.

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Research Article

Clinical Use of Aided Cortical Auditory Evoked Potentials as a Measure of Physiological Detection or Physiological Discrimination

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The clinical usefulness of aided cortical auditory evoked potentials (CAEPs) remains unclear despite several decades of research. One major contributor to this ambiguity is the wide range of variability across published studies and across individuals within a given study; some results demonstrate expected amplification effects, while others demonstrate limited or no amplification effects. Recent evidence indicates that some of the variability in amplification effects may be explained by distinguishing between experiments that focused on physiological detection of a stimulus versus those that differentiate responses to two audible signals, or physiological discrimination. Herein, we ask if either of these approaches is clinically feasible given the inherent challenges with aided CAEPs. N1 and P2 waves were elicited from 12 noise-masked normal-hearing individuals using hearing-aid-processed 1000-Hz pure tones. Stimulus levels were varied to study the effect of hearing-aid-signal/hearing-aid-noise audibility relative to the noise-masked thresholds. Results demonstrate that clinical use of aided CAEPs may be justified when determining whether audible stimuli are physiologically detectable relative to inaudible signals. However, differentiating aided CAEPs elicited from two suprathreshold stimuli (i.e., physiological discrimination) is problematic and should not be used for clinical decision making until a better understanding of the interaction between hearing-aid-processed stimuli and CAEPs can be established.

1. Introduction

The potential clinical benefits of a measure of brain encoding and plasticity in hearing aid users have driven a growing interest in aided cortical auditory evoked potentials (CAEPs). A better understanding of the effects of hearing aids on brain function, and resulting behavior, may improve the current science underlying successful rehabilitation of hearing loss. CAEPs, a type of event-related electroencephalography (i.e., scalp-recorded electrical brain activity) recorded 50–300 ms following stimulus onset, are thought to reflect neural activity in reverberant thalamocortical circuits (for a review see [1, 2]). Aided CAEPs, or potentials recorded when stimuli are presented via a hearing aid, have been proposed as a possible physiological measure of the effects of amplification on the brain. Many studies have explored the potential use of aided CAEPs, demonstrating considerable variability in results across experiments and individual participants [3–21]. The variability across studies highlights the current uncertainty surrounding the clinical usefulness of aided CAEPs.

Methodological differences contribute to the variable results that exist in the aided CAEP literature. Much of the existing aided CAEP literature can be grouped into two general approaches: (1) a focus on physiological response detection, or (2) emphasis on physiological response discrimination. The physiological detection approach compares the CAEPs from an (inaudible or barely audible) unaided stimulus, to the response obtained from the same stimulus

that has been processed by a hearing aid and delivered at a suprathreshold level. In this case, the unaided CAEP is often absent or weak, while the aided CAEP is often present with robust waveform morphology. The absence or presence of a response demonstrates a good correlation with inaudible and audible stimuli, amounting to a physiological correlate of detecting the presence of sound. Historically, CAEPs have successfully been used to estimate behavioral thresholds (approx. within 10 dB of behavioral threshold) of both normal-hearing and hearing-impaired populations [22–24]. However, it remains to be established whether such strong correlations between physiological and behavioral thresholds are maintained in the case of aided CAEPs, in which the stimuli have been altered by hearing aid processing.

In contrast to the physiological detection approach, the physiological discrimination approach compares CAEPs from two audible stimuli to determine differences between the waveforms (e.g., unaided versus aided conditions in normal-hearing individuals or two audible aided conditions with varied parameters). Physiological discrimination is measured by specific differences in waveform morphology (i.e., differences in peak latencies and peak amplitudes) between two present waveforms.

The different focus of these two approaches contributes to the variability in the existing literature involving aided CAEPs. Significant changes in waveform morphology (i.e., amplification effects) were often found when individuals/groups were tested in studies that used a physiological detection approach, comparing inaudible to audible conditions (e.g., [8, 14, 17]); whereas, amplification effects were often absent or small in studies that used a physiological discrimination approach comparing two suprathreshold responses (e.g., [3, 4, 20]). A comparison of these approaches is presented in Figure 1 and the corresponding Table 1, where the examples in the left column (a–d) demonstrate clear amplification effects obtained in a physiological detection approach; in contrast, examples of a physiological discrimination approach displayed in the right column (e–h) show small or absent amplification effects. It is noteworthy that many of the early publications highlighted case studies, and that despite the significant number of aided CAEP publications, only a limited subset displayed electrophysiological waveforms.

In addition to considering the audibility of the signal, it is equally important to take into account the audibility of the underlying noise and its relationship to the signal (i.e., signal-to-noise ratio or SNR). SNR is a key contributor to both unaided and aided CAEP morphology [4, 25]. Accounting for SNR is particularly important when establishing the effects of hearing aids on auditory processing because hearing aids both contribute circuit noise inherent to signal processing and because hearing aids amplify ambient environmental noise. When establishing the clinical utility of aided CAEPs, it is important to establish the effects of noise on CAEPs in situations where signal and noise are audible in both unaided and aided conditions or when two different aided conditions are being compared. Clinicians might encounter problems when fitting a hearing aid if they assume that changes to the hearing aid should improve the morphology of the evoked response when in reality no change should occur.

The purposes of this study are to characterize the existing aided CAEP literature relative to two potential clinical approaches, and to determine whether these approaches (i.e., physiological detection and physiological discrimination) demonstrate clinical utility for encoding hearing-aid-processed signals. Specifically, we asked two questions:

(1) Will aided CAEPs be different for a near-threshold signal relative to a suprathreshold signal (i.e., physiological detection)?

(2) Will aided CAEPs be different for two suprathreshold signals (i.e., physiological discrimination)?

We set out to answer these questions in the aided CAEP domain by recording hearing aid output and eliciting CAEPs with the recorded stimuli.

2. Methods

CAEPs were recorded using hearing-aid-processed stimuli. In addition, a background noise masker was presented to the normal-hearing participants as a means of simulating the audibility factors that are present when an individual with hearing impairment is fit with a hearing aid. Testing noise-masked normal-hearing participants allowed tight control of audibility (i.e., audibility of hearing-aid-processed signals and hearing aid noise) while avoiding hearing-impairment-related confounds associated with recording CAEPs from individuals with hearing impairment.

2.1. Participants. Twelve individuals (six women and six men) participated in the study (mean age = 22.1 years; SD = 2.47). All participants were right handed with normal hearing from 250 to 8000 Hz (≤20 dB HL), were in good general health, reported no significant history of otologic or neurologic disorders, and denied use of mood or sleep-altering medications. All participants provided informed consent, and all research complied with the regulations of the Portland Veterans Affairs Medical Center Institutional Review Board.

2.2. Stimuli

2.2.1. Hearing Aid Signal and Noise. Hearing-aid-processed stimuli were used to elicit CAEPs. Signals were 1000-Hz tones recorded from the output of two hearing aids: Hearing Aid A was a currently available digital hearing aid, Hearing Aid B was the same analogue hearing aid used in previous studies from our laboratory [3, 4]. The frequency response of both hearing aids was matched in the test box of a hearing aid analyzer (Fonix 8000, Frye Electronics, Tigard, Oregon) to be within 3 dB at frequencies between 200 and 5000 Hz using a 45 dB pure-tone sweep. The hearing aids were then placed on the right ear of the Brüel and Kjær Head and Torso Simulator (type 4128C) using stock foam earmolds with no venting. The Head and Torso Simulator was placed inside an Eckel Corp. fully anechoic chamber (reverberation time

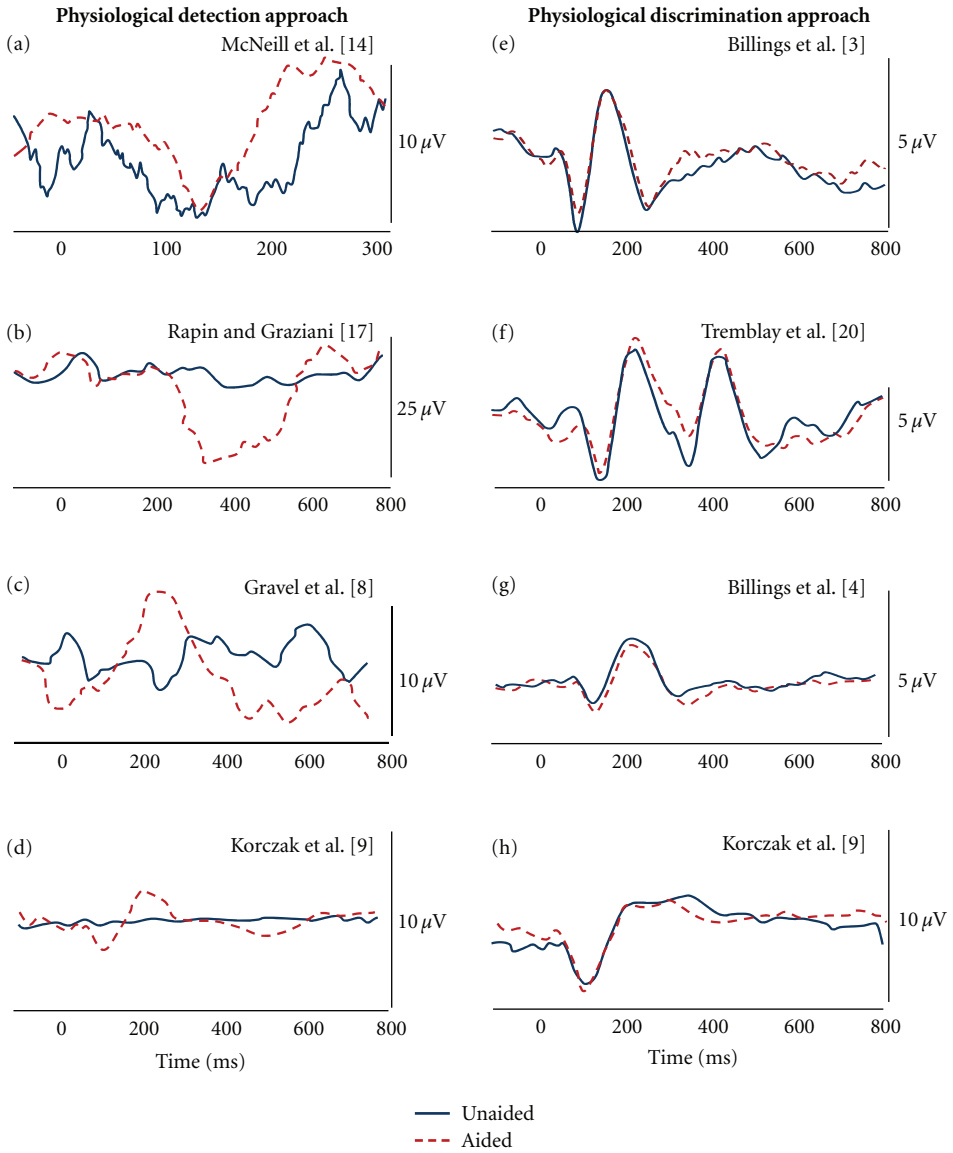

Figure 1: Examples of physiological detection (a–d) and physiological discrimination (e–h) approaches from the aided CAEP literature. Results across these studies demonstrate significant amplification effects (unaided versus aided) for physiological detection, but very limited amplification effects for physiological discrimination. All figures were modified from published figures; the appropriate citation is indicated for each panel (see Table 1 for details).

constant of 2 ms and background noise level of −10 dB SPL at 1000 Hz). A 450-ms, 1000-Hz tone with 9 ms rise/fall times and an interstimulus interval (ISI) of 1900 ms was presented via a Cambridge Soundworks free-field speaker placed at 0 degrees azimuth at a distance of 1.5 meters from the hearing aid microphones. Volume control wheels, available noise cancellation algorithms, and directional microphones were deactivated on both hearing aids. Overall hearing aid gain and speaker output level were then varied systematically to result in three recordings (see Table 2). Electroacoustic analysis of the settings used in the three recordings revealed that attack and release times for Hearing Aid A were 5 ms and 225 ms, respectively; using a 1000-Hz tone, compression ratios were measured at 1.25 : 1 from 50–75 dB and 1 : 1 from 75–90 dB. Hearing Aid B attack and release times were 2.5 ms for both recordings; a 1 : 1 compression ratio was found from 50–60 dB and 2 : 1 from 60–90 dB for Recording 2 settings, and an essentially linear input/output function was measured for Recording 3 settings.

The goal was to record three signals that varied in signal-to-noise ratio (SNR). As indicated in Table 2, resulting SNRs for the three recordings were 8 dB, 11 dB, and 23 dB. Recordings were approximately two minutes in length and consisted of continuous hearing aid noise and a total of 50 signal presentations. The use of an anechoic chamber for the recordings ensured that the hearing aid noise consisted entirely of circuit noise, as the environmental noise was negligible (−10 dB SPL in the 1/3 octave band surrounding 1000 Hz). The three recordings were then scaled in Matlab (Version 7.0, Mathworks, Natick, MA) to create stimuli

Table 1: Outline of the experimental conditions used in the studies cited in Figure 1. Note that the parameters listed below are specific to the conditions used to generate the waveform data presented in Figure 1 and do not necessarily represent all conditions presented in each study. The eight studies included represent aided CAEP data in the literature for which unaided and aided waveforms were able to be reproduced.

	Subjects		Experimental design					Source
	n	Characteristics	Stimulus	Duration (ms)	ISI (ms)	Signal level	Conditions	
Figure 1 Detection approach examples								
(a) McNeil et al., 2009 [14]	1	68 yo male; severe to profound SNHL	/ba/	115	750*	60 dB nHL	Unaided and aided	Figure 1, page 81: unaided and left hearing aid
(b) Rapin and Graziani, 1967 [17]	1	21 mo female; rubella, sedated	500-Hz tone	Not specified	Not specified	109 dB re: 0.0002 dynes/cm^2	Unaided and aided	Figure 4, page 892: 109 dB with and 109 dB without aid
(c) Gravel et al., 1989 [8]	1	7 mo male; severe to profound SNHL	/da/	Not specified	Not specified	Not specified	Unaided and aided (aid set to user settings)	Figure 3, page 271
(d) Korczak et al., 2005 [9]	7	Adults; severe to profound SNHL	/ba/ and /da/	150	950	80 dB ppeSPL	Unaided and aided (aid set to MCL)	Figure 3, page 176: Standard responses in lower panel, left
Figure 1 Discrimination approach examples								
(e) Billings et al., 2007 [3]	13	Young adults; normal hearing	1000-Hz tone	757	1910	50 dB SPL	Unaided and aided (20 dB gain)	Figure 2, page 238: 50 dB aided and 50 dB unaided waveforms
(f) Tremblay et al., 2006 [20]	7	Young adults; normal hearing	/si/	655	1910	64 dB SPL	Unaided and aided (average gain of 19 dB)	Figure 7, page 99: Top panel
(g) Billings et al., 2011 [4]	9	Young adults; normal hearing	1000-Hz tone	756	1910	40 dB SPL	Unaided and aided (gain of 20 dB)	Figure 6, page 7: Panel (a) unaided waveform, panel (b) aided waveform
(h) Korczak et al., 2005 [9]	4	Adults; moderate SNHL	/ba/ and /da/	150	950	85 dB HL	Unaided and aided (aid set to MCL)	Figure 3, page 176: Standard responses in lower panel, center

* Reference does not state whether this value refers to onset to onset, or offset to onset. All other interstimulus intervals (ISIs) refer to offset to onset.

Table 2: Three recordings of hearing aid output. Specific characteristics of the three hearing aid recordings used in this study to elicit aided CAEPs.

	Hearing Aid	Gain at 1000 Hz[1]	Input[2]	Output SNR[3]
Recording 1	A	30 dB SPL	25 dB SPL	11 dB
Recording 2	B	30 dB SPL	25 dB SPL	8 dB
Recording 3	B	10 dB SPL	45 dB SPL	23 dB

[1] Gain with a 45 dB SPL input (electroacoustically verified).
[2] Input: input to hearing aid microphones in the sound field.
[3] Output SNR: difference between hearing-aid-processed signal and noise at 1000 Hz (1/3 octave band).

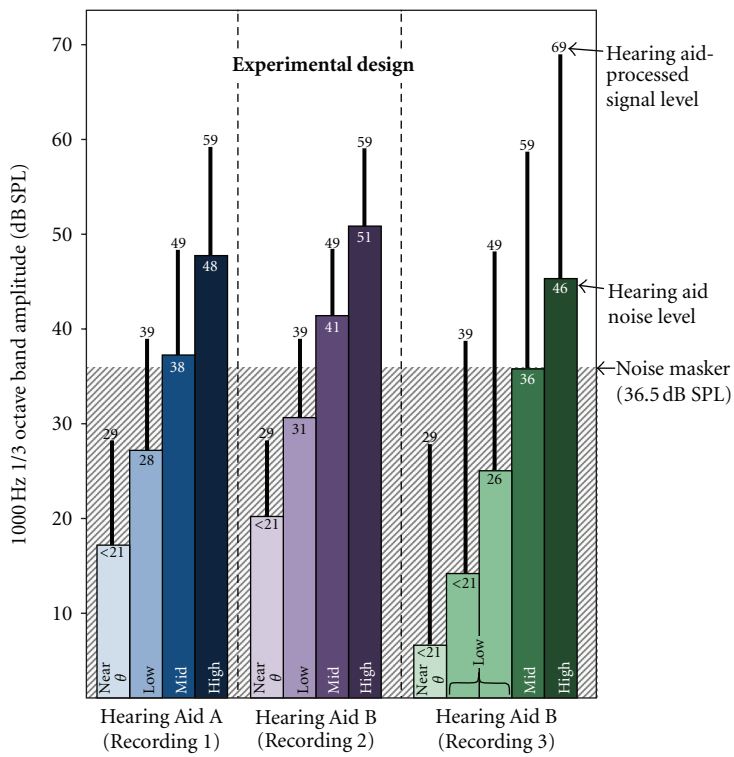

Figure 2: Experimental design. One-third octave band levels at 1000 Hz are shown for the three hearing-aid-processed recordings. Scaling of the recordings resulted in Near θ, Low, Mid, and High conditions. The shaded background shows the background noise masker level relative to hearing aid signal and noise levels.

of varying absolute levels such that the audibility of the tonal signal and the underlying hearing aid noise varied systematically relative to the background noise masker (see Section 2.2.2 for details of background noise creation).

A schematic depicting the hearing aid signal (i.e., 1000-Hz tone) and noise levels in relation to the background noise masker for each hearing aid condition is shown in Figure 2. For low-input recordings from Hearing Aid A and Hearing Aid B, four scaling factors were calculated that adjusted the level of the tone and hearing aid noise in approximately 10 dB steps to produce the following presentation levels: Near θ, a near-threshold level at which both tone and hearing aid noise were inaudible due to the noise masker amplitude (intended to represent an inaudible unaided condition); Low, a level at which the signal was above the level of the noise masker but the hearing aid noise was below; Mid and High, two levels at which the signal and hearing aid noise were both audible but absolute signal level differed. Due to the higher ratio of signal level to hearing aid noise obtained in the higher input level recording from Hearing Aid B, five stimulus levels were necessary in order to span the range of audible and inaudible signals. In total, 13 experimental conditions were presented to each participant. Notice that, within each block of hearing aid conditions, the SNR between the signal level and the hearing aid noise remains constant (i.e., approx. 11 dB for Hearing Aid A (Recording 1), 8 dB for Hearing Aid B (Recording 2), 23 dB for Hearing Aid B (Recording 3), while the overall stimulus output level changes in 10 dB steps. However, the effective SNR was much smaller for Near θ and Low conditions due to the background noise masker. Levels displayed in Figure 2 are 1/3 octave band values centered at 1000 Hz measured with a Brüel and Kjær 2260 Investigator sound level meter fitted with a Brüel & Kjær ear simulator (number 4157).

The hearing aid noise spectra obtained from the three 59 dB hearing aid condition presentations are represented

in Figure 3 (measured with the sound level meter in 1/3 octave bands with center frequencies from 100 to 6300 Hz) along with the noise floor of the measurement system. To ensure that the lowest signal level presentations were below threshold, participants were asked to listen to each of the three lowest level conditions and report whether or not they could detect the tonal signal. This measure confirmed that the lowest stimulus levels for each of the three hearing aid recordings were either inaudible or barely audible to all participants.

2.2.2. Background Noise Masker. In order to simulate hearing-impaired thresholds and to control the audibility of the hearing-aid-processed signal and underlying hearing aid noise, a continuous background noise masker was created in Matlab passing a Gaussian white noise through a series of 1/3 octave band filters with center frequencies from 100 to 5000 Hz. The output of the filters was adjusted to generate a noise masker with a spectrum matching the thresholds of a patient with a moderate, sloping hearing loss. The spectrum of the noise masker was verified with a spectrum analyzer in 1/3 octave bands. In addition, behavioral thresholds at octave and interoctave frequencies from 250 to 8000 Hz were established using ER-3A (Etymotic Research, Inc., Elk Grove Village, IL) insert earphones. The thresholds of four participants were established using 1 dB steps in a 1-up, 2-down procedure while the background masker was played through the audiometer. Mean behavioral thresholds for the four individuals were 11.0, 15.75, 22.25, 24.25, 28.0, 34.75, 44.75, 43.25, and 8.75 dB HL at frequencies of 250, 500, 750, 1000, 2000, 3000, 4000, 6000, and 8000 Hz, respectively.

2.3. Electrophysiology. For each of the 13 conditions, a recorded 50-tone wav file was repeated three times, yielding a total of 150 tone presentations recording over approximately six minutes for each condition. Both the noise masker and hearing aid noise were continuous throughout each block of trials, and all stimuli were presented in the right ear using an ER-3A insert earphone and the Stim2 system (Compumedics Neuroscan, Charlotte, NC). The presentation order of the three hearing aid conditions was randomized across subjects, and the various stimulus levels were randomized within each hearing aid condition to minimize order effects across participants. Two-minute listening breaks were given between each condition, and subjects were offered a longer break after one hour of testing. Acquisition sessions lasted three hours and consisted of consenting, audiometric testing, electrode placement, and CAEP acquisition. Participants were seated comfortably in a double-walled sound attenuating booth, and they were instructed to ignore the auditory stimuli and to watch a closed-captioned movie of their choice.

Evoked potential activity was recorded using an Electro-Cap International, Inc. cap which housed 64 tin electrodes. The ground electrode was located on the forehead and Cz was the reference electrode. Data were rereferenced offline to an average reference. Horizontal and vertical eye movement was monitored with electrodes located inferiorly and at the outer canthi of both eyes. The recording window consisted of a 100-ms prestimulus period and a 700-ms poststimulus time. Using Scan 4.5 (Compumedics Neuroscan, Charlotte, NC), evoked responses were analog bandpass filtered online from 0.15 to 100 Hz (12 dB/octave roll off) and converted using an analog-to-digital sampling rate of 1000 Hz. Trials with eye-blink artifacts were corrected offline using Neuroscan software. This blink reduction procedure calculates the amount of covariation between each evoked potential channel and a vertical eye channel using spatial, singular value decomposition and removes the vertical blink activity from each electrode on a point-by-point basis to the degree that the evoked potential and blink activity covaried [26]. After blink correction, trials containing artifacts exceeding 70 μV were rejected from averaging. After artifact rejection, the remaining sweeps were averaged and filtered offline from 1 Hz (highpass filter, 24 dB/octave) to 30 Hz (lowpass filter, 24 dB/octave). Averages for 98% of conditions tested had more than 100 accepted trials; the remaining 2% had between 70 and 100 accepted trials. N1 and P2 peak amplitudes and latencies were determined by agreement of two judges. Each judge used temporal electrode inversion, global field power (GFP) traces, and even and odd sweep waveform versions (to demonstrate replication) for a given condition. Peaks in the Near θ conditions were very difficult to identify because of the electrophysiological noise. Therefore, in order to quantify synchrony in the near-threshold conditions, the rectified area was measured in the time region from 40 to 300 ms as a representation of synchrony of the P1-N1-P2 complex. Area values were generated at all Near θ and Low conditions.

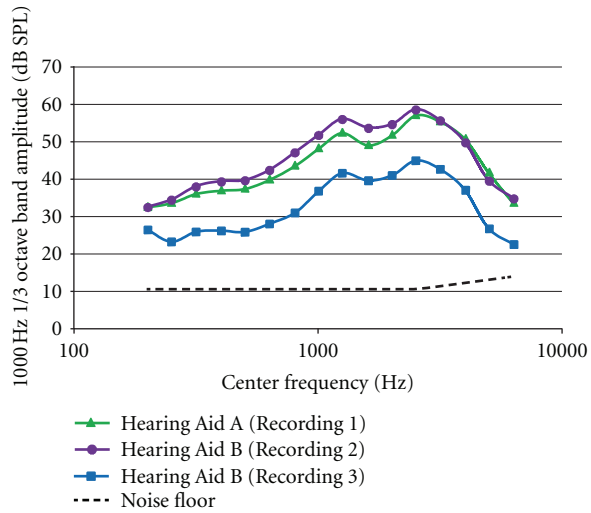

Figure 3: Frequency spectra of hearing aid noise for each of the three hearing aid conditions. Values are 1/3 octave bands with center frequencies between 200 and 6300 Hz. Hearing aid noise was measured for the 59-dB signal level condition for each recording. The general pattern of noise spectra is similar across conditions with a spectral peak at 1000 Hz, the frequency of the signal. The noise floor of the measurement system is shown with the dashed line (note: the lower limit of the sound level meter was 10.5 dB).

2.4. Statistical Analysis. We fit the linear mixed model representation of the repeated measures analysis of variance (ANOVA). This model has the two-fold advantage of (1) being fit using maximum likelihood so that all observations are included in the analysis and not just observations for subjects with complete data, and (2) taking into account nonsymmetrical variances that may occur across conditions. Main effects of hearing aid condition were tested for two contrasts: Low versus Mid and Mid versus High. The Mid versus High comparison directly tests the physiological discrimination approach while the Low versus Mid comparison verifies that when SNR is changing, the aided CAEP is also likely to change. Where main effects were found, post-hoc comparisons were made for each hearing aid condition.

To test the effectiveness of the physiological detection approach, paired comparisons were completed on rectified area measures of Near θ versus Low for the three hearing aid conditions. Area measures were used because of the large number of absent responses in the Near θ condition.

3. Results

The current study investigated the ability of aided CAEPs to demonstrate amplification effects using both neural physiological response detection and physiological discrimination approaches. Based on a review of literature, we hypothesized that aided CAEPs would show the most robust effect of amplification in neural response detection approaches which correlate with the difference in response to audible versus inaudible stimuli. In contrast, aided CAEP discrimination approaches that reflect the ability of the auditory system to represent differences between audible hearing-aid-processed stimuli were expected to show weak amplification effects. Results, presented below, are organized to address these two primary hypotheses.

3.1. Physiological Detection. This study addressed the effects of amplification on audible and inaudible stimuli by comparing the Near θ condition (i.e., aided CAEP responses to the lowest stimulus level at which the tone level was at or below the noise masker level) to the Low condition (i.e., aided CAEP responses to the stimulus level at which the tone was just audible above the noise masker). For Hearing Aid B (Recording 3), two different Low conditions were used; however, to simplify the data analysis, all data for the two Low conditions were averaged together to result in a measurement for one Low condition. Butterfly plots (overlaid responses at all electrodes across the scalp) and the global field power plots (a quantification of simultaneous activity across the scalp; [27]) calculated from the grand average for these two stimulus levels across subjects and hearing aid conditions are presented in Figure 4 (left). The butterfly plot of the Near θ condition is shown in the top-left panel with the response of the Cz electrode highlighted in blue. The butterfly plot of the Low condition response is shown in the center-left panel with the Cz electrode highlighted in dashed red. The bottom-left panel shows the GFP waveforms for the Near θ and Low conditions overlaid in blue and dashed red, respectively. Notice that the response to the Near θ condition contains considerable noise across electrodes. In contrast, responses to the Low condition result in visible peaks in activity across electrodes resulting in clearly identifiable peak waves in the GFP plot as well as the Cz electrode response. N1 and P2 peaks were difficult to identify in many of the Near θ conditions, resulting in large amounts of missing data (approx. 50% of N1 and P2 peaks were not able to be identified), making statistical analysis using traditional peak latency and amplitude values difficult; therefore, an area measure (rectified area was calculated from 40 to 300 ms) was used to provide an overall measure of synchrony in the P1-N1-P2 region of the waveform. Figure 5 displays area values for the Near θ and Low conditions for all three hearing aid conditions. Paired comparisons between Near θ and Low conditions generally demonstrated significantly higher areas for Low conditions relative to Near θ conditions:

Hearing Aid A (Recording 1): $t = -2.077$, $df = 11$, $p = .062$;

Hearing Aid B (Recording 2): $t = -2.853$, $df = 11$, $p = .016$;

Hearing Aid B (Recording 3): $t = -4.225$, $df = 11$, $p = .003$.

3.2. Physiological Response Discrimination. The discrimination task of the current study measured the ability of aided CAEP measures to reflect differences between two clearly audible stimuli (i.e., Mid and High conditions). To demonstrate the main effect of signal level, butterfly and GFP plots were constructed from grand average responses across subjects and hearing aid recordings for Mid and High conditions (Figure 4, right). The top-right panel of Figure 4 depicts the butterfly plot in response to the Mid conditions with the Cz electrode highlighted in orange. Notice that a clear response waveform is present and the coherence between responses among electrodes on the butterfly plot. The center-right panel shows the butterfly plot in response to High conditions with the Cz electrode highlighted in dotted green. Again, a response is clearly present as expected given audibility of the signal. In the lower-right panel, GFP responses for the Mid and High conditions are overlaid and plotted in orange and dotted green, respectively. Notice that the GFP response between the two conditions is very similar with nearly identical peak amplitudes and latencies. This figure helps to highlight the difficulty in using aided CAEP measures to establish the brain's ability to physiologically discriminate between two suprathreshold hearing-aid-processed stimuli.

N1 and P2 peak amplitude and latency data for the Low, Mid, and High conditions are displayed in Figure 6 with corresponding statistical analyses shown in Table 3. As mentioned above, the responses to the two Low conditions for Hearing Aid B (Recording 3) were averaged together for latency/amplitude comparisons between the three hearing aid conditions. The results indicate no main effect of Mid versus High conditions for N1 and P2 measures. However, comparisons of Low versus Mid conditions resulted in main effects of N1 and P2 latency and amplitude.

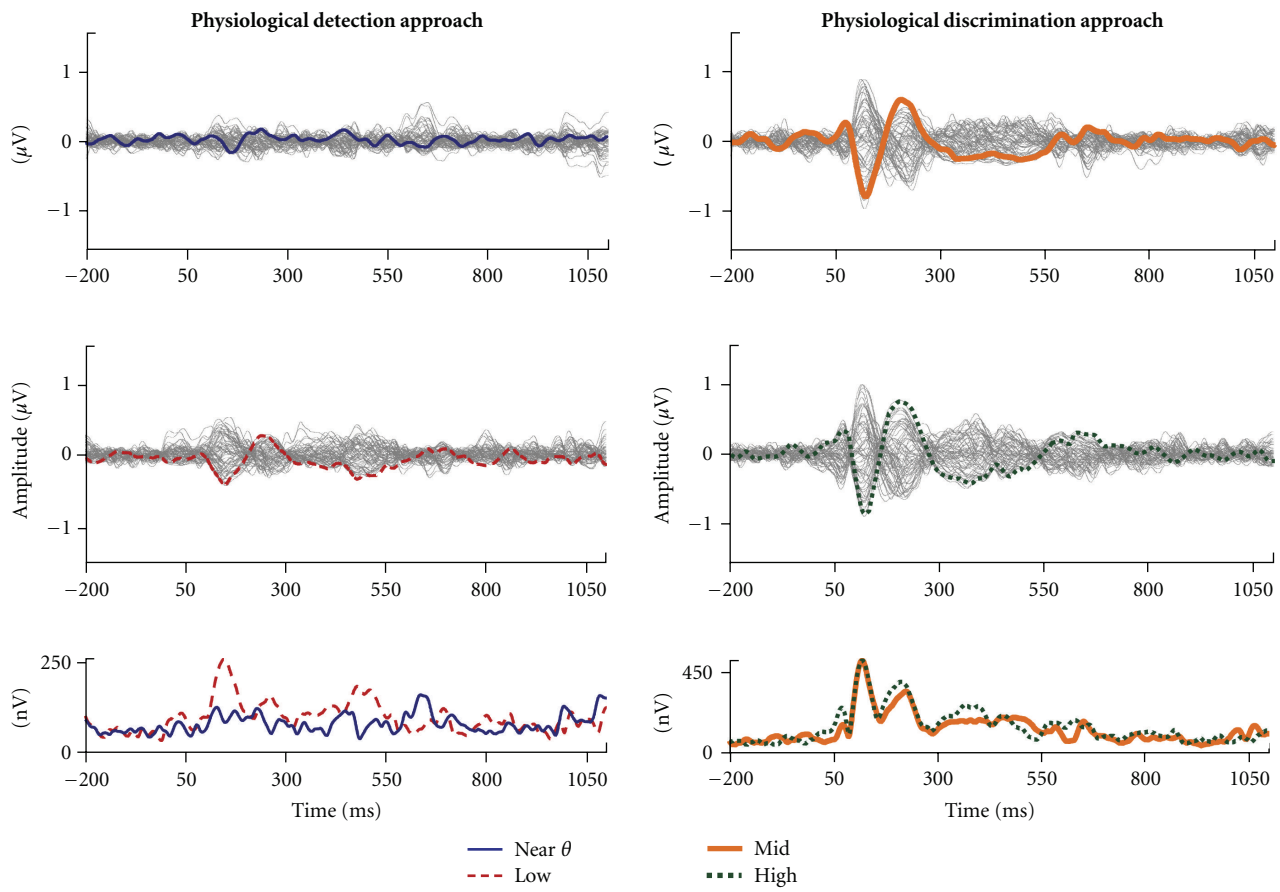

Figure 4: Grand average ($n = 12$) butterfly plots and global field power (GFP) waveforms of physiological detection and physiological discrimination results. Waveforms are collapsed across hearing aid recordings. Near θ (top left), Low (middle left), Mid (top right), and High (middle right) conditions are displayed with the Cz-electrode highlighted. Bottom panels show overlaid comparisons for Near θ versus Low conditions and Mid versus High conditions. Robust differences are shown for physiological detection (bottom left) and minimal differences are shown for physiological discrimination (bottom right).

3.3. SNR Effects on CAEPs. Results from previous literature indicate that CAEP responses are more sensitive to changes in SNR than to changes in absolute signal level [4, 25]. In the present study, a lack of significant change in N1 and P2 amplitudes or latencies in response to the Mid and High conditions supports the idea that waveforms generally do not change when SNR is held constant. To further quantify the effects of SNR on aided CAEPs, we compared responses across Low, Mid, and High conditions. The Low conditions had smaller SNRs than the Mid and High conditions because of the audible background noise masker. For both the Mid and High conditions in all three hearing aid conditions, SNR was dictated by the constant ratio of signal level to hearing aid noise within the 1000-Hz band surrounding the signal tone. The SNR for each hearing aid condition was 11 dB for Hearing Aid A (Recording 1), 8 dB for Hearing Aid B (Recording 2), and 23 dB for Hearing Aid B (Recording 3). Therefore, the SNR was constant at the two highest presentation levels within each hearing aid condition but varied between hearing aid conditions. If SNR drives amplitude and latency changes in CAEPs, we would expect to see significant changes between the Low and Mid/High conditions, and potentially differences between hearing aids due to differences in SNR. The amplitude and latency plots (Figure 6) generally indicate the expected changes in N1 amplitudes and latencies.

This visual impression is confirmed by the significant main effect of Low to Mid level conditions on N1 and P2 amplitudes and latencies, which is not found in the comparison between Mid to High level conditions (Table 3). Further, the effect of the difference in SNR between hearing aid conditions was apparent when comparing the statistical significance between the Low and Mid level stimuli for each hearing aid condition. The hearing aid condition with the largest output SNR (Hearing Aid B, Recording 3) was most likely to show significant differences in N1 and P2 amplitudes and latencies between Low and Mid level recordings, followed by the hearing aid with the second largest output SNR (Hearing Aid A, Recording 1). The hearing aid condition with the poorest SNR (Hearing Aid B, Recording 2) was least likely to demonstrate significant differences in peak amplitudes or latencies. Overall, these findings corroborate earlier reports of the significant influence of SNR on CAEP recordings.

Table 3: Statistical anlalysis. A linear mixed model representation of the repeated measures ANOVA resulted in a main effect of the level contrast with post-hoc comparisons where the main effect was significant.

Conditions	Main effect			Hearing aid A (Recording 1)			Hearing aid B (Recording 2)			Hearing aid B (Recording 3)		
	F Value	*df*	*p*	*F* Value	*df*	*p*	*F* Value	*df*	*p*	*F* Value	*df*	*p*
Low to Mid level												
N1 Latency (ms)	9.36	3,97	<0.0001	13.78	1,97	0.0003	4.03	1,97	0.0474	10.25	1,97	0.0018
P2 Latency (ms)	17.53	3,96	<0.0001	30.52	1,96	<0.0001	0.47	1,96	0.4955	21.6	1,96	<0.0001
N1 Amplitude (μV)	6.92	3,97	0.0003	0.03	1,97	0.868	4.18	1,97	0.0435	16.54	1,97	<0.0001
P2 Amplitude (μV)	3.16	3,96	0.0282	2.84	1,96	0.0954	0.67	1,96	0.4134	5.96	1,96	0.0164
Mid to High level												
N1 Latency (ms)	0.64	3,97	0.5894	—	—	—	—	—	—	—	—	—
P2 Latency (ms)	0.38	3,97	0.7698	—	—	—	—	—	—	—	—	—
N1 Amplitude (μV)	0.31	3,97	0.8163	—	—	—	—	—	—	—	—	—
P2 Amplitude (μV)	2.07	3,96	0.109	—	—	—	—	—	—	—	—	—

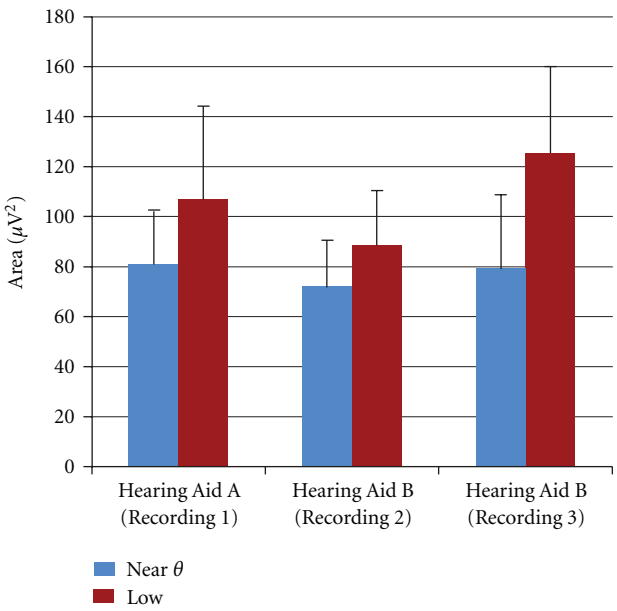

Figure 5: Area measurements for Near θ and Low conditions for three recordings. Low conditions yielded higher area values than Near θ conditions for all three hearing aid recordings, demonstrating aided CAEP morphology differences that are present for physiological detection.

4. Discussion and Conclusions

The purpose of this study was to help clarify some of the variability that is seen in decades of aided CAEP research. This variability is shown in Figure 1/Table 1 where some studies demonstrate robust effects of amplification (a–d) while others do not (e–h). We hypothesized that large portions of this variability can be explained by whether the signal and underlying noise are audible relative to the contrasting condition. Two approaches, physiological detection (i.e., the absence versus presence of a CAEP) and physiological discrimination (i.e., the differentiation of two present responses), were tested. Figure 4 demonstrates the overall results of this study relative to these two approaches. Physiological detection demonstrates robust amplification effects, while physiological discrimination demonstrates limited differences between Mid and High conditions. These results are in agreement with those found by Korczak and colleagues [9] where both approaches can be identified in subsets of their data. Korczak et al. compared unaided and aided CAEPs in two groups of individuals with hearing loss (some with moderate hearing loss and some with severe hearing loss) using stimuli presented at approximately 70 and 85 dB HL. They found improved CAEPs only for the lower level stimulus or when the unaided stimulus was near threshold (i.e., physiological detection; Figure 1(d)). Interestingly, when both unaided and aided stimuli were likely above threshold (i.e., physiological discrimination), as was the case for the moderate hearing-impaired group using a 85-dB stimulus, limited effects of amplification were found on N1 and P2 waves (Figure 1(h)).

4.1. Clinical Feasibility: Physiological Detection versus Physiological Discrimination. The physiological detection approach appears to be a reasonable use of aided CAEPs because these measures are sensitive to differences in detectability of an inaudible or barely audible signal and a suprathreshold signal. Our results, and the results of other past studies, demonstrate robust amplification effects when taking a detection approach [8, 9, 14, 16, 17]. This study simulates the process a clinician may use in fitting a hearing aid, in which hearing aid gain is increased in 10-dB steps and the resulting CAEP is examined. In this scenario, the increasing signal level demonstrates a robust effect. Figure 7 shows two representative individuals from the 12 participants and demonstrates the clinical process of increasing the gain of a hearing aid. The Near θ curve shows an absent response in most cases; whereas, the Mid and sometimes Low conditions show present responses.

In contrast, these data and examples from the literature [3, 4, 9, 15, 20] demonstrate that approaching aided CAEPs from a physiological discrimination perspective is problematic, especially if background noise is audible, as is the case for the Mid and High conditions in this study. For two individuals (Figure 7), similarities between Mid and High conditions are apparent and demonstrate the difficulty in differentiating between the CAEPs to two suprathreshold signals. Importantly, statistical differences were found for comparisons between Low and Mid conditions (see Table 3);

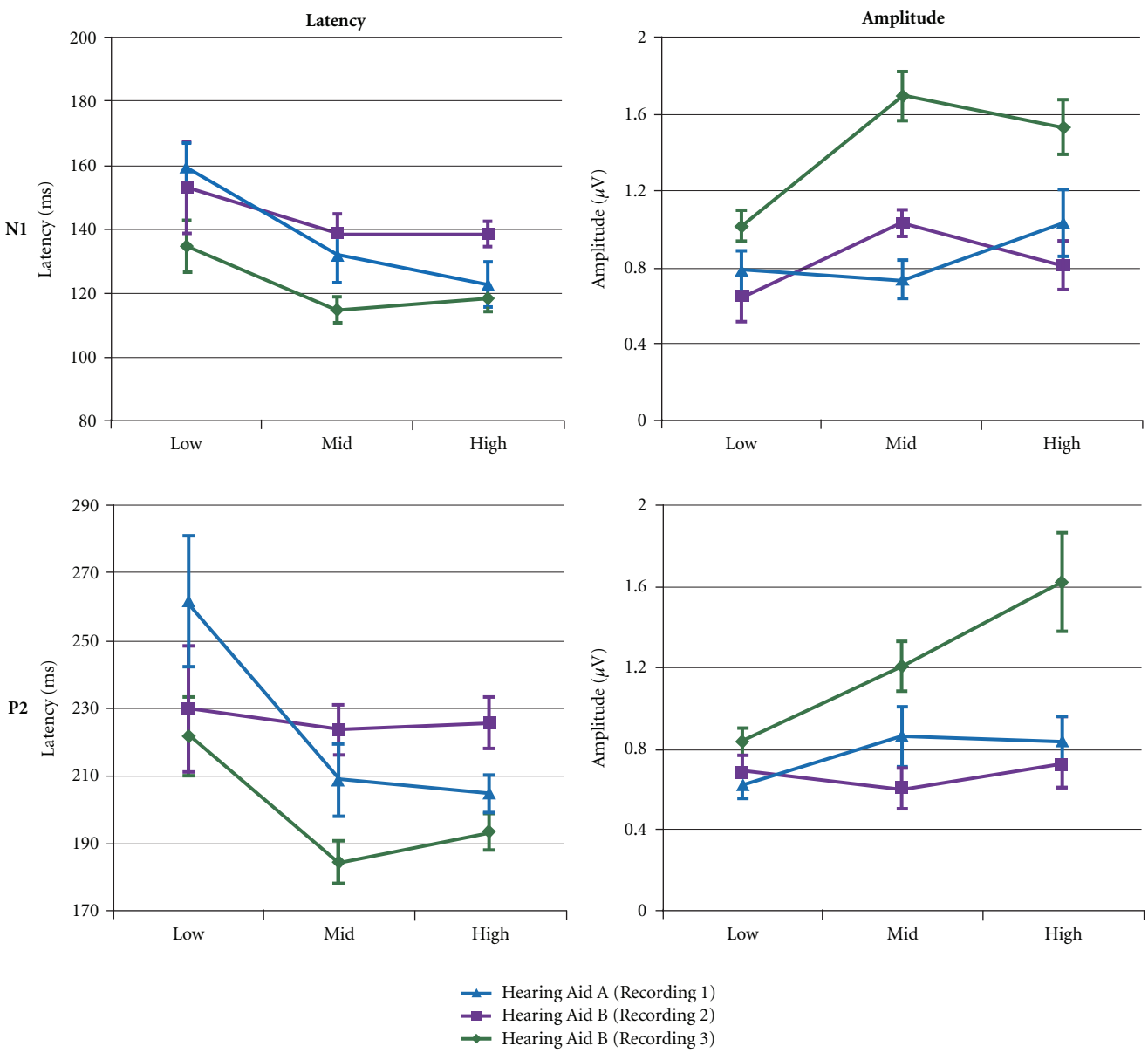

Figure 6: Mean latency and amplitude measures for Low, Mid, and High conditions as a function of hearing aid recording (error bars: standard error of the mean). Generally, a change from Low to Mid conditions results in decreases in latency and increases in amplitude, and a change from Mid to High results in minimal change in latency and amplitude.

in these cases, differences reflect changes in SNR as the hearing aid noise is not yet audible. Therefore, the comparison between Low and Mid conditions can be considered a successful example of physiological discrimination. Indeed, CAEPs have been used successfully as a measure of physiological discrimination for decades; however, our understanding of *aided* CAEPs is still lacking. When evoking CAEPs with hearing-aid-processed stimuli, it remains unclear when a discrimination approach is valid; it may be valid only for specific hearing aids, specific hearing aid settings, specific stimuli, or as in this study, specific conditions (e.g., Low versus Mid and not Mid versus High). Certainly, clinical decisions based on a physiological discrimination approach would be premature. Additional research is needed to delineate what hearing aid and stimulus interactions are affecting the evoked response.

It is important to consider subject factors as well. The audibility of a broadband stimulus and the underlying noise will vary depending on the hearing configuration of the individual being tested. The participants in this study were young normal-hearing individuals, and a noise masker was used to simulate thresholds that were comparable with a typical sloping hearing loss. However, even with tightly controlled audibility and a pure tone stimulus, variability across participants was found. Figure 8 demonstrates how 10-dB increments in signal level affect N1 latency and amplitude in the 12 individuals tested. Testing hearing-impaired individuals with broadband stimuli would likely

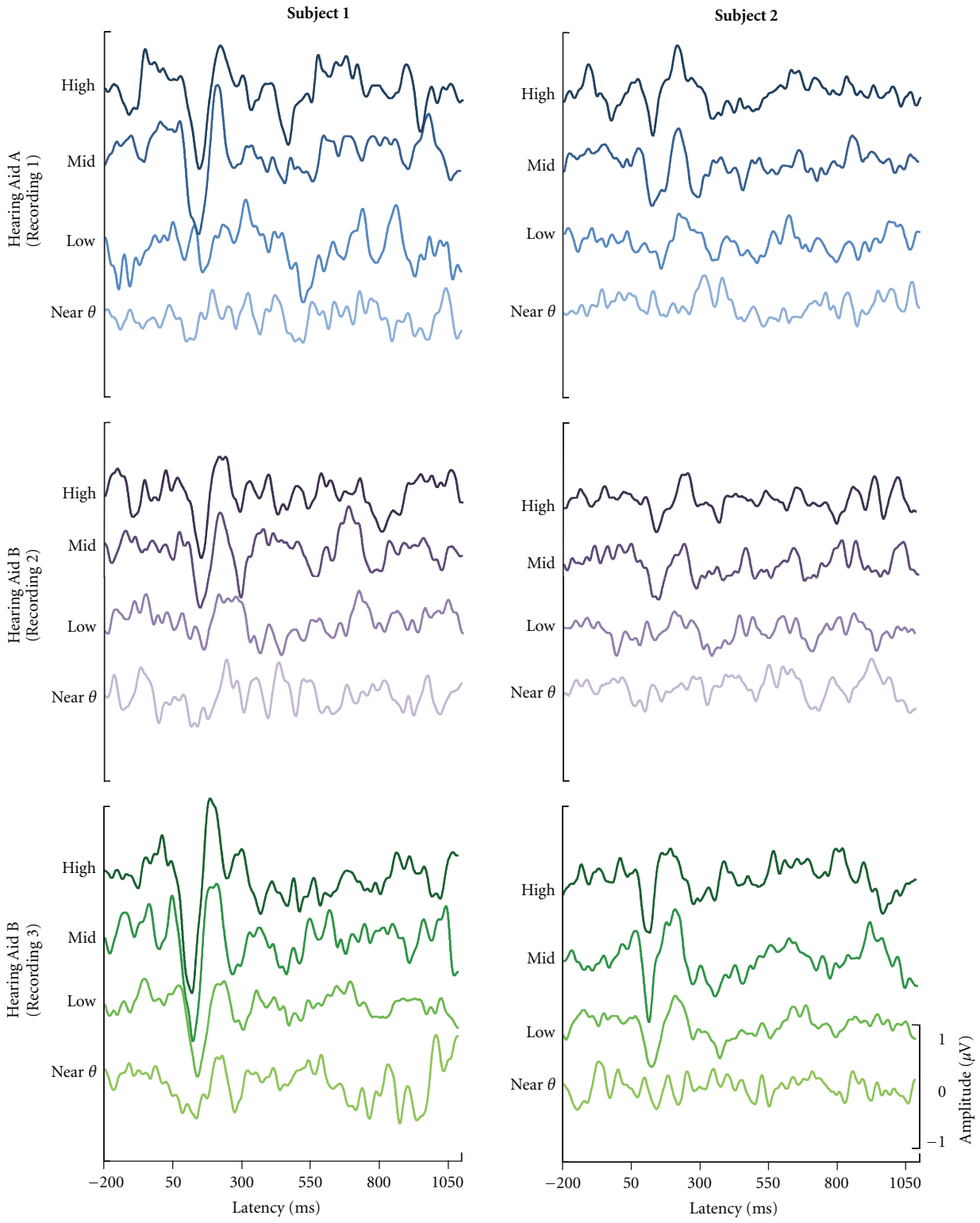

Figure 7: Cz-electrode waveforms for two representative individuals across the three hearing aid recordings. For both participants, the Near θ condition shows an absent or very small response, while the Low condition shows a more pronounced response. Mid and High responses are present and similar to each other. Effects of hearing aid recording are somewhat apparent with the most robust waveforms occurring in the Hearing Aid B (Recording 3) condition.

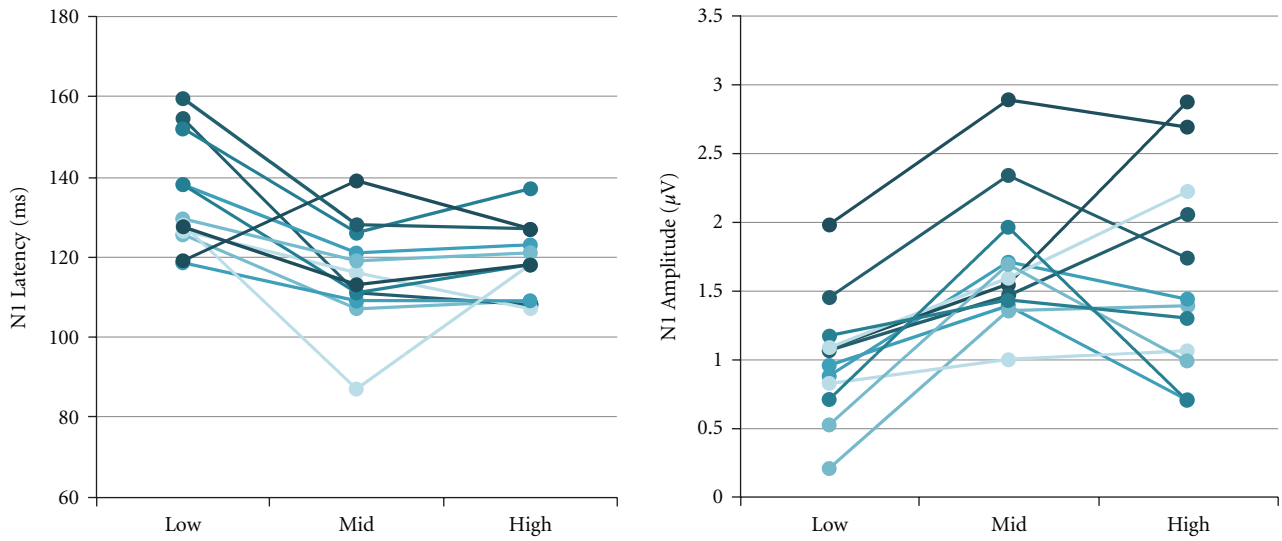

Figure 8: Individual N1 latency and amplitude values demonstrating the physiological discrimination approach. Low, Mid, and High conditions are shown for Hearing Aid B (Recording 3). The general trends, consistent with Figure 6, show considerable changes from Low to Mid conditions and minimal changes from Mid to High conditions. Variability across individuals is evident with some individuals contradicting the general trends.

result in increased variability across participants because of varying etiologies of hearing loss and differences in threshold across frequencies. Therefore, use of aided CAEPs in individuals in a clinical setting using a physiological discrimination approach is likely to result in considerable variability resulting from the many varying subject and stimulus factors.

It should be noted that the method of scaling used in the design of this study to modify signal level is different than clinical hearing aid gain adjustments in that modification of gain can lead to a wide range of acoustic modifications to the signal. While SNR has been shown to remain similar across gain settings in some hearing aids [4], SNR can also vary significantly from one device and recording condition to another [28]. These considerations are important when using a physiological discrimination approach; while one hearing aid may show a physiological amplification affect, another device may not because of the specific acoustic features that are modified. This variability in outcomes severely limits the clinical usefulness of physiological discrimination approaches in aided CAEPs until more is understood about the interaction between hearing-aid-processed signals and their effects on the evoked response.

4.2. Major Acoustic Contributors to Aided CAEPs. As mentioned above, the problems related to the physiological discrimination approach likely result from a combination of subject and stimulus factors. SNR and onset modification are two stimulus characteristics whose importance has been demonstrated in the literature (e.g., [3, 4, 12, 25, 29–31]). First, the effects of SNR are important to consider when recording evoked potentials at the level of the cortex, because cortical neurons are more sensitive to SNR than to absolute signal level [32, 33]. The audibility of underlying noise in the hearing-aid-processed signal, whether amplified ambient noise or circuit noise, must be considered when interpreting aided CAEPs. Second, onset changes to the time waveform that result from hearing aid processing are also important. The N1-P2 CAEP is an onset response, meaning that it is generated when many cortical pyramidal cells fire synchronously to the onset of a stimulus. These neurons are especially sensitive to abrupt changes in amplitude or frequency; therefore, hearing-aid modifications to the onset are important to consider [12, 30]. These considerations are complicated when speech stimuli are used, because it becomes difficult to determine the SNR across different portions of the speech signal, particularly in light of changes in compression across running speech.

The results of this study seem to indicate that the contribution of stimulus factors can be minimized when the physiological detection approach is used. Specific signal-processing modifications made by the hearing aid are less important when the comparison response waveform is absent. In contrast, subtle acoustic changes (e.g., modification of SNR or onset characteristics) are essential when comparing two audible signals in a physiological discrimination approach. To characterize acoustic changes, it is necessary to complete in-the-canal recordings of hearing-aid-processed signals. Only then can measures of the important signal modifications be made and related to the resulting aided CAEPs.

4.3. Conclusions. Two approaches for using aided CAEPs, physiological detection and physiological discrimination, were tested to determine the clinical usefulness of each. Results are in agreement with an analysis of the literature (see Figure 1), and they demonstrate that physiological detection, or a determination of the presence of a response to an audible signal relative to the absent response of an inaudible signal, is likely a valid use of aided CAEPs and provides

an indication of the encoding of the aided signal at the level of the auditory cortex. In contrast, the physiological discrimination approach (i.e., the comparison of waveforms that are generated by two audible signals) can be problematic and difficult to interpret in individuals when using hearing-aid-processed stimuli. A more detailed understanding of how hearing aid processing modifies stimulus acoustics (e.g., SNR and onset characteristics) is needed before the physiological discrimination approach should be used for clinical decision making.

Abbreviations

CAS: Central auditory system
SNR: Signal-to-noise ratio
SD: Standard deviation
GFP: Global field power
CAEPs: Cortical auditory evoked potentials
ANOVA: Analysis of variance.

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Research Article

Processing Load Induced by Informational Masking Is Related to Linguistic Abilities

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It is often assumed that the benefit of hearing aids is not primarily reflected in better speech performance, but that it is reflected in less effortful listening in the aided than in the unaided condition. Before being able to assess such a hearing aid benefit the present study examined how processing load while listening to masked speech relates to inter-individual differences in cognitive abilities relevant for language processing. Pupil dilation was measured in thirty-two normal hearing participants while listening to sentences masked by fluctuating noise or interfering speech at either 50% and 84% intelligibility. Additionally, working memory capacity, inhibition of irrelevant information, and written text reception was tested. Pupil responses were larger during interfering speech as compared to fluctuating noise. This effect was independent of intelligibility level. Regression analysis revealed that high working memory capacity, better inhibition, and better text reception were related to better speech reception thresholds. Apart from a positive relation to speech recognition, better inhibition and better text reception are also positively related to larger pupil dilation in the single-talker masker conditions. We conclude that better cognitive abilities not only relate to better speech perception, but also partly explain higher processing load in complex listening conditions.

1. Introduction

A major complaint of both hearing-impaired and normal hearing individuals is the high level of effort while following a conversation in a noisy situation. Although sensory hearing loss is considered the main cause of speech communication difficulties [1, 2], comprehension of speech in noise is not fully predicted by a pure-tone audiogram or other psychoacoustical tests [3–6]. Research has shown that speech comprehension and related listening effort are not only based on sensory processes, but also on linguistic and working-memory-related cognitive abilities [2, 7, 8]. These insights were obtained as the result of two major areas of science (namely, hearing sciences and cognitive sciences) merging into one area of cognitive hearing science (CHS) which we witnessed during the last decade [9]. A next step in CHS research would be the examination of the interaction between use and benefit of devices like hearing aids, and individuals' cognitive abilities and mental effort [10, 11]. Attempts into that direction were made by Gatehouse et al. [12, 13] who observed a relationship between an individual's cognitive abilities and candidature for a certain hearing aid fitting pattern. However, before these and other insights obtained within CHS can be applied to clinical practice (i.e., hearing aid fitting evaluation) we need to know more precisely what cognitive processes is associated with listening effort. Although it is often assumed that the involvement of cognitive functions in speech comprehension is responsible for the listening effort that people experience, it is not known yet how these two are related. In other words: how do cognitive factors differentially impact on (a) speech understanding and (b) effort or processing load deployed

during listening. This is what we focused on in the current study. We investigated how cognitive capacity associates to speech comprehension and relates to listening effort in normal hearing participants.

In our recent study [14] we found that cognitive load during speech processing differed for different types of background noise. It was observed that listening to speech masked by a single talker evoked a larger pupil dilation response than listening to speech masked by fluctuating or stationary noise. The effect was independent of speech intelligibility level. The authors concluded that the effect was most likely caused by semantic interference of the single talker masker. Cognitive abilities supposedly associated with this effect are working memory (WM) [8, 15, 16], the ability to inhibit irrelevant speech while storing information in WM [17, 18], and linguistic abilities [3, 19, 20]. Note that the inhibition of irrelevant speech information is an important capacity in daily life listening. While listening to a speaker, listeners often have to neglect other irrelevant speakers (e.g., during a cocktail party) [21]. The inhibition of information could be one of the cognitive functions affecting processing load during listening.

Whereas the role of these abilities in speech *perception* has been repeatedly demonstrated, their relation with the *processing load* evoked by speech perception, as assessed by pupillometry, has been examined rarely. One of the studies in which this issue was explored was Zekveld et al. [22]. However, Zekveld et al. did not measure working memory capacity and neither included different types of noise maskers. The aim of the current study was to investigate the relation of cognitive abilities and the processing load induced by perceiving speech in different types of distracter conditions. Specifically, the study aimed to address the associations between these cognitive abilities during listening and the additional load imposed by the semantic interference of a single-talker masker as compared to the masking imposed by a fluctuating noise masker [14].

Processing load is shown to be reflected by pupil dilation as measured with pupillometry [23, 24]. Pupil dilation is also related to effort caused by allocation of attentional resources [25]. Interestingly, pupil dilation is sensitive to language processing tasks like hearing and reading words or sentences [26–30]. In a pioneering study, Kramer et al. [28] investigated the pupil response in relation to speech processing in adverse listening conditions. By using pupillometry in combination with a speech reception threshold (SRT) task, it was shown that the SNR affected processing load as reflected by changes in the pupil dilation response (see also, [22, 31]).

Zekveld et al. [22] showed that subjects with better text reception thresholds (TRT) allocated more cognitive resources (larger pupil dilations) during speech perception in stationary noise. This effect was independent of intelligibility level. Additionally, Zekveld et al. [32] observed that better TRT performance was associated with increased brain activation in the left angular gyrus (AG) during cued speech perception. AG is associated with "combinatorial semantic processing" [33] a process in which word fragments are combined into full sentences by means of semantic structure. These results may suggest that individuals with good TRT performance employ a different strategy during listening that may require more processing load. However, we have to be careful in directly relating these results on AG activation with pupillometry data as obtained in the current study, because the pupil response reflects a more co

mplex network of brain areas than AG alone [34]. Contrasting findings were observed for RSpan capacity. Larger RSpan capacity was associated with *less* activation in inferior frontal and superior temporal brain regions. To gain more insight into the factors involved in this complex area of research, we assessed both WMC and TRT in the current study to investigate how interindividual differences in these measures affect SRT and pupil dilation.

In the present study, we included several measures of cognitive ability known to be related to speech comprehension and often assumed to be related to processing load during listening. First, we assessed WM capacity (WMC). According to the "ease of language understanding" (ELU) model [35], WM is strongly involved in language processing specifically when speech is partly masked by fluctuating background sound [8, 16, 17, 36]. In such challenging situations, WM enables the listener to keep a mental representation of a spoken sentence while using knowledge of language and context to fill in gaps in the information. WMC has often been shown to partly explain the frequently observed interindividual differences in speech processing in adverse listening conditions [15, 35–37].

Therefore, in the current study we applied a Dutch version of the reading span (Rspan) as well as a listening span (Lspan) task. Both tests are adaptations of the Rspan test used in previous studies [38–41]. Secondly, a Dutch version of the size-comparison span (SICspan) task was included [17, 18]. The SICspan measures the ability to inhibit irrelevant linguistic information while storing information in WM. Better inhibition of the distractor items in addition to remembering more target items will lead to a higher SICspan score. Finally, we included the TRT test [20], assessing combinatorial semantic processing [32] by asking subjects to read partly masked text. This ability is associated with both speech perception in noise [7, 42] and the cognitive processing load during listening, as indicated by both pupillometric [22] and functional magnetic resonance imaging (fMRI) data [32]. Note, that the TRT is not a measure of cognitive capacity like the span tasks. Instead it is a visual equivalent of the SRT, a threshold of language comprehension in adverse conditions. In the current study, we investigated the differential effect of an individual's cognitive abilities on listening effort and speech understanding in normal hearing adults. This knowledge is required when considering the method of pupillometry for use in hearing aid fitting evaluations in the future. Participants performed an SRT task in fluctuating noise and against the background of a single-talker masker. The adaptive procedures targeted either 50% or 84% correctly repeated sentences. During the tasks, pupil responses were recorded and subjective effort, performance, and motivation ratings were acquired at the end of each block of sentences. The primary aim of the study was to examine if the pupil response evoked by semantic interference was related to WMC, WM-related

inhibition, or TRT. We examined the associations between the peak pupil dilation (PPD) and individuals' linguistic and cognitive abilities. Since we observed larger pupil responses for speech masked by an interfering talker as compared to speech masked by a fluctuating masker in our previous study, we also examined the differences in PPD between fluctuating noise and a single-talker masker conditions. Based on previous studies that already investigated some of these effects and associations [14, 22, 32], we expected that a higher cognitive capacity would be associated with a larger pupil response during speech perception in more complex conditions compared to easy listening conditions.

2. Methods

2.1. Participants. Thirty-two adults (aged between 40 and 70 years, mean age 51.3 years, 6 males) with normal hearing, recruited at the VU University Medical Centre, participated in the study. Normal hearing was defined as having pure-tone thresholds less than or equal to 20 dB HL at the individual frequencies 250, 500, 1000, 2000, and 4000 Hz in both ears, having no more than a single 35 HL dB dip at one of these frequencies in one ear. Participants had no history of neurological diseases, reported normal or corrected-to-normal vision, and were screened for near-vision acuity [43]. They were native Dutch speakers and provided written-informed consent in accordance with the Ethics Committee of the VU University Medical Center.

2.2. SRT. The speech reception threshold (SRT) [44] was measured by presenting speech in fluctuating noise or masked with a single-talker masker [14, 45]. The SRT adaptively assessed the SNR required to perceive either 50% or 84% of the sentences entirely correctly [44, 46]. Both masker types had a long-term average spectrum adapted to the spectrum of the target speech signal [47]. The target sentences were spoken by a female voice and for the single-talker masker concatenated sentences were used, spoken by a male voice with modified spectrum. The fluctuating noise mimicked the intensity fluctuations of speech, by multiplying the noise signal by the envelope of the speech of the single-talker masker. Each of these four conditions was measured in a blocked fashion and the level of the target speech was fixed at 55 dBA. Each block contained 39 short Dutch sentences [47] and the order of the blocks was counterbalanced over participants.

2.3. TRT. The text reception threshold (TRT) task [20] is a visual analog to the SRT task. In this task participants read text sentences presented on a computer screen in red font on a white background and masked by black vertical bars. Sentences appeared on a screen word by word with a similar timing as the word onsets in the corresponding recorded SRT sentences. After the onset of the last word, the full sentences remained on the screen for 500 ms [42]. A 1-up-1-down adaptive procedure with a step size of 6% was applied, targeting the percentage of unmasked text required to read 50% of the sentences entirely correctly. The sentences presented were selected from the same corpus as used for the SRT [47] but did not overlap with those presented in the SRT tests. Participants performed four tests with 13 sentences each; the first test was a practice test of which the data were excluded from the analysis. The TRT was the average percentage of unmasked text in the three remaining tests with the first four sentences omitted. Lower thresholds indicate better performance.

2.4. Rspan and Lspan. Reading span (Rspan) and listening span (Lspan) tests were used to assess verbal WM capacity in the visual and auditory domain, respectively. Each test consisted of 54 sentences that were presented in sets of 3 to 6 sentences. Half of the sentences were semantically incorrect. Participants did not know beforehand whether they were to remember and report the initial or final noun of each sentence. After the presentation of each block, they had to repeat either the last or first words in the correct order. This kind of postcueing procedure makes the task less strategically manageable and hence more difficult. In addition, participants performed a semantic judgment task after each individual sentence during presentation of the sentence set. In the Rspan test each sentence was visually presented [42, 48]. The Lspan sentences were presented dichotically through headphones at 65 dBA. Subjects responded verbally. Prior to each test participants practiced on 10 sentences divided over three sets. The span size corresponds to the total number of correctly recalled target words irrespective of their order of presentation, with a maximum score of 54. Higher scores indicate better performance.

2.5. SICspan. In the "size-comparison span" (SICspan) task [17, 18], participants were asked to make relative size judgments between two items (e.g., Is LAKE bigger than SEA?) by pressing "J" key for yes and "N" for no on a QWERTY keyboard. Each question was followed by a single to-be-remembered item, which was semantically related to the objects in the sentence (e.g., RIVER). Ten sets were presented: 2 to 6 size comparison questions each, followed by a to-be-remembered item. Participants were asked to verbally recall the to-be-remembered items in order of presentation. Sentences and to-be-remembered items within each set were from the same semantic category, but between sets, the semantic categories were different. The SICspan score used in this study contained all correctly remembered items independent of order, which leads to a maximum score of 40. The higher the score the better the performance on the SICspan task.

2.6. Apparatus. Participants were tested in a sound-treated room. During the SRT task participants had to fixate their gaze to a dot (diameter 0.47°) located at 3.5 meter distance, at eye level on a white wall. Throughout the SRT test, the pupil diameter of the left eye was measured by an infrared eye tracker (SMI, 2D Video-Oculography, version 4). Light intensity was adjusted by an overhead light source so that the pupil diameter was around the middle of its dilation range at the start of the experiment. For both the SRT and the Lspan

task, audio in the form of separate files (44.1 Hz, 16 bit) was presented binaurally by an external soundcard (Creative Sound Blaster, 24 bit) through headphones (Sennheisser, HD 280, 64 Ω). During all visual tasks and during the Lspan task participants were facing a computer screen (Dell, 17 inch) at 60 cm distance. All tests were presented by a Windows PC (Dell, Optiplex GX745, 2.66 GHz 2 Core).

2.7. Procedure. Participants started the test session with either the Rspan or the Lspan task (order was balanced over subjects). Additionally, they performed two-blocked conditions of the SRT task (order was balanced over subjects). After a 10-minute break participants performed the SICspan task followed by the two remaining experimental conditions of the SRT task. After a second 10-minute break participants performed the remaining Lspan or Rspan task. Participants ended the session by performing the TRT task. The test session took 2.5 to 3 hours.

During the SRT task the pupil response was used as a measure of processing load. Pupil traces and SRT data of the trials containing the first four sentences were omitted further from analysis. For all remaining traces diameter values more than 3 SDs smaller than the mean were coded as blinks. Traces containing more than 15% blinks were excluded others were deblinked by means of a linear interpolation. A spike detection algorithm was used to detect eye movements (for a full description see, [49]) on both the x- and y-traces. All trials with a range in x- or y-amplitudes exceeding 2 SDs, within a sliding window of 100 ms, were excluded from analysis. All remaining traces were baseline corrected by subtracting the mean pupil size within the 1-second period prior to the speech onset. The PPD was calculated for each subject for each condition. PPD was defined as the highest value within a time window of 4.4 seconds after speech onset, which resembled the interval between speech onset and the response prompt.

After each SRT block, participants rated their effort, performance, and motivation level during the block. Participants had to indicate how much effort it took to perform the SRT task on a continuous scale from 0 ("no effort") to 10 ("very effortful"). Additionally, participants indicated how they themselves perceived their performance on the task by rating between 0 ("none of the sentences were intelligible") and 10 ("all sentences were intelligible"). Finally, to assess their degree of motivation during the course of the test, participants indicated how often during the block they had abandoned the listening task, because the task was too difficult. This was rated between 0 ("this happened for none of the sentences") and 10 ("this happened for all of the sentences"). Prior to analysis the motivation score was inversed, so high scores reflect high motivation. Note that a continuous scale (range between 0–10) was applied. Participants were explicitly instructed that they could give ratings in between the whole numbers on the scale, which was reflected in the raw scores that also showed a normal distribution when tested for skewness and kurtosis. For more details on stimuli, SRT procedure, pupillometry, and subjective ratings see Koelewijn et al. [14].

2.8. Statistical Analysis. A repeated measures analysis of variance (ANOVA) testing the effects of intelligibility (50% and 84%) and masker type (fluctuating noise and single-talker masker) was performed on the SRT scores, PPD, and the subjective ratings. Statistically significant ($P < .05$) interactions were further analyzed by means of two-tailed paired samples t-tests. Additionally, Pearson correlation coefficients were calculated to test the relations between age, PTA, Rspan, Lspan, SICspan, TRT, and SRT. Finally, linear regression analyses were performed to examine the associations between interindividual differences in SRT or PPD (dependent variables) and cognitive abilities (Rspan, Lspan, SICspan) and TRT as independent factors. For each dependent variable, regression analyses were performed separately for each masker type and both intelligibility levels.

In addition, we used regression models to examine the associations between cognitive abilities, and the additional PPD imposed by semantic interference. The same was done for the association between TRT and this informational masking effect. We calculated difference scores for both the SRT (Δ_{SRT}) and PPD (Δ_{PPD}) by subtracting the outcome for fluctuating noise averaged over both intelligibility conditions from the outcome for the single talker averaged over both intelligibility conditions. For each of the regression models, we examined whether age and PTA were each individually confounding the relationship between the dependent and independent variables. A variable was considered as a relevant confounder when the regression coefficient changed with 10% or more after adding the potential confounder to the analysis. Additionally, the potential confounder had to be associated with both the independent (cognitive abilities and TRT) and the dependant (SRT or pupil response) factors. All statistical analyses were performed using SPSS version 17.

3. Results

3.1. Behavioral Results SRT. The average SRTs (dB SNR) in fluctuating noise and in the single-talker masker, at intelligibility levels of 50% and 84%, are plotted in Figure 1. The average SRT, PPD, and the subjective ratings for each condition are reported in Table 1.

An ANOVA on the SRTs showed a main effect of intelligibility ($F_{[1,31]} = 438.82$, $P < .001$) with a lower $SRT_{50\%}$ (mean SNR = −11.9 dB) compared to $SRT_{84\%}$ threshold (SNR = −5.9). Additionally, a main effect of masker type was observed ($F_{[1,31]} = 5.09$, $P = .031$) showing a slightly lower threshold for the single-talker masker (SNR = −9.3) than in fluctuating noise (SNR = −8.5). Also, an interaction between intelligibility level and masker type was observed ($F_{[1,31]} = 5.66$, $P = .024$). Post hoc analysis revealed a significant difference at 50% intelligibility between the single-talker masker (SNR = −12.6) and fluctuating noise (SNR = −11.2) conditions ($t_{[31]} = -3.65$, $P = .001$), and no masker effect at the 84% intelligibility level. These results indicate an overall effect of intelligibility level on the SRT and a slightly lower SRT in single talker compared to fluctuating noise at 50% intelligibility.

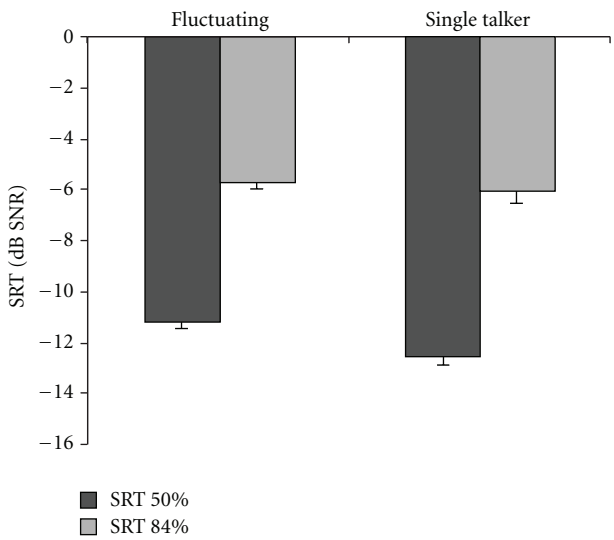

Figure 1: SRTs (dB SNR) at two intelligibility levels for both masker types, averaged over subjects. The error bars show the standard errors for each condition.

Table 1: The average SRT scores, PPD, and the subjective ratings for both levels of intelligibility and for both masker types.

Intelligibility	Fluctuating	Single talker
SRT	SNR (SD), dB	
50%	−11.2 (1.5)	−12.6 (1.9)
84%	−5.7 (1.5)	−6.0 (2.8)
Pupil	PPD (SD), mm	
50%	0.23 (.16)	0.29 (.16)
84%	0.16 (.12)	0.22 (.16)
Subjective	Effort (low = 0–high = 10)	
50%	6.9 (1.5)	6.7 (1.3)
84%	4.7 (1.7)	5.3 (1.6)
	Performance (low = 0–high = 10)	
50%	5.5 (1.4)	5.4 (1.0)
84%	7.1 (1.0)	6.9 (1.2)
	Motivation (low = 0–high = 10)	
50%	8.0 (1.4)	7.9 (1.7)
84%	8.6 (1.3)	8.6 (1.3)

3.2. Pupil Data SRT. Pupil traces containing a large number of blinks (in total 6.0% of the traces) and/or large eye movements (in total 9.8% of the traces) were removed from further analysis. PPD was calculated over the remaining traces for each condition. The average traces for the four conditions are plotted in Figure 2.

An ANOVA on PPD revealed a main effect of intelligibility level ($F_{[1,31]} = 20.31$, $P < .001$), with a larger PPD in the $\text{SRT}_{50\%}$ conditions (0.26 mm) compared to the $\text{SRT}_{84\%}$ conditions (0.19 mm). Additionally, there was an effect of masker type ($F_{[1,31]} = 40.66$, $P < .001$) with a larger average PPD for the single-talker masker (0.26 mm) compared to fluctuating noise (0.20 mm). No interaction between intelligibility level and masker type was observed ($F_{[1,31]} < 1$).

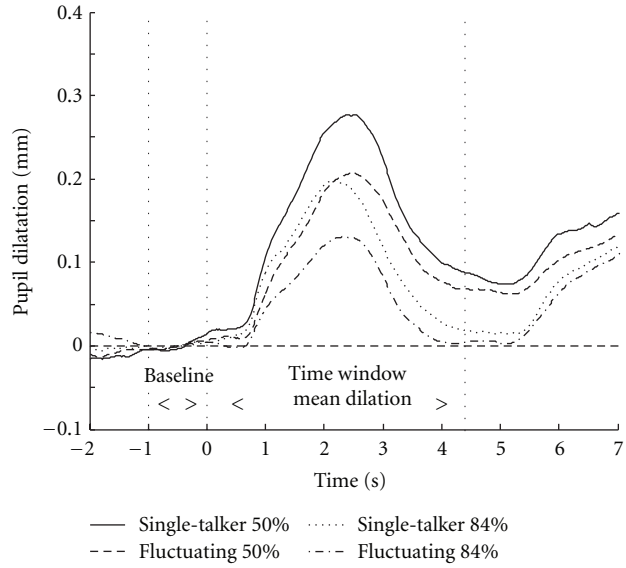

Figure 2: Pupil responses per condition averaged over subjects. The onset of the sentences is at 0 sec. The baseline is indicated as the average pupil diameter over one second preceding the start of the sentence. The area between the second and third dotted lines indicates the time window used for calculating the mean pupil dilation.

3.3. Subjective Ratings SRT. An ANOVA was performed for each of the three subjective ratings separately (Table 1). An effect of intelligibility level on the *subjective effort* ratings was observed ($F_{[1,31]} = 46.50$, $P < .001$), indicating that subjectively, lower intelligibility makes speech perception more effortful. Masker type did not affect the ratings ($F_{[1,31]} < 1$). However, an interaction between intelligibility level and masker type was observed ($F_{[1,31]} = 6.45$, $P = .016$). Post hoc analysis revealed that only in the 84% condition, a significant difference in subjective effort between the single-talker masker (5.3) and fluctuating noise (4.7) conditions was found ($t_{[31]} = -2{,}18$, $P = .037$). Note that the 84% condition SRTs for fluctuating noise (SNR = −5.7) and the single-talker masker (SNR = −6.0) did not differ significantly ($t_{[31]} < 1$). An effect of intelligibility on the *subjective performance* ratings was observed ($F_{[1,31]} = 105.12$, $P < .001$) showing lower ratings at 50% intelligibility (5.5) then at 84% intelligibility (7.0). Additionally, no effect of masker type ($F_{[1,31]} < 1$) or interaction effect ($F_{[1,31]} < 1$) was observed. *Subjective motivation* ratings showed a main effect of intelligibility level ($F_{[1,31]} = 19.77$, $P < .001$). No effect of masker type ($F_{[1,31]} < 1$) or interaction ($F_{[1,31]} < 1$) was observed. Participants were less motivated in the 50% intelligibility conditions (7.9) compared to the 84% conditions (8.6).

3.4. Descriptive Statistics Cognitive Tests. For each subject we calculated the total scores for the Rspan (mean = 15.5, sD = 4.4), Lspan (mean = 21.4, sD = 3.6), and SICspan (mean = 23.8, sD = 6.1). Additionally, the individual TRTs were calculated (mean = 59.8, sD = 5.5). Correlation analyses showed that there were no significant correlations

Table 2: Two-tailed Pearson correlations (*$P < .05$, **$P < .01$) between age, PTA, Rspan, Lspan, SICspan, TRT, SRT with fluctuating noise at 50% (SRT_{F50}) and 84% (SRT_{F84}) intelligibility, and SRT with a single-talker masker at 50% (SRT_{ST50}) and 84% (SRT_{ST84}) intelligibility. Lower TRTs and SRTs indicate better performance.

	Age	PTA	Rspan	Lspan	SICspan	TRT
Age	X					
PTA	.468**	X				
Rspan	−.299	−.316	X			
Lspan	−.048	−.182	.669**	X		
SICspan	−.278	−.390*	.658**	.585**	X	
TRT	.305	.330	−.759**	−.584**	−.684**	X
SRT_{F50}	.342	.097	−.079	−.317	−.208	.186
SRT_{F84}	.235	.105	−.361*	−.261	−.430*	.248
SRT_{ST50}	.352*	.203	−.501**	−.348	−.480**	.673**
SRT_{ST84}	.509**	.284	−.463**	−.282	−.499**	.540**

between age and each of the span tasks (Rspan, Lspan, and SICspan), or age and TRT. Pearson correlations between each of the cognitive tests (Rspan, Lspan, SICspan) and TRT ranged between .58 and .77 and were statistically significant (Table 2).

3.5. Relation between Cognitive Abilities, Speech Perception, and Processing Load. To examine whether SRT and PPD during speech perception were associated to WMC (Rspan, Lspan), inhibition (SICspan), and linguistic processing (TRT), regression analyses were performed for the behavioral and PPDs separately. The slope (B), the variance (R^2), and the P values for the independent factors explaining the performance in $SRT_{50\%}$ and $SRT_{84\%}$ are shown in Table 3(a). Table 3(b) shows the results for the PPD in $SRT_{50\%}$ and $SRT_{84\%}$. In none of the equations, PTA was a confounder and hence, not adjusted. Age appeared to be a confounder in some of the analyses, in which case age was included in the model. Only significant ($P < .05$) associations are shown. Note that separate models were run for each of the cognitive measures, to eliminate colinearity. Analyzing each of the cognitive functions separately allowed us to examine and compare the individual association with the dependent variables. Note that the reported R^2 is always based on the single-dependent measure.

The outcomes for the SRT regression analyses (Table 3(a)) showed no significant associations for the fluctuating noise condition at 50% intelligibility (SRT_{F50}). For the fluctuating noise condition at 84% intelligibility (SRT_{F84}), a significant association with Rspan and SICspan was found. For the single-talker masker at 50% (SRT_{ST50}) and 84% intelligibility (SRT_{ST84}), significant associations were found for Rspan, SICspan, and TRT. In all models, higher (better) Rspan and SICspan scores were related to lower (better) SRTs. Additionally, lower (better) TRTs were related to lower (better) SRTs.

The outcomes for the regression analyses with PPD as the dependent measure (Table 3(b)) showed no associations between PPD and cognitive abilities in both fluctuating noise conditions. However, in the single-talker masker condition at 50% intelligibility, the PPD (PPD_{ST50}) was significantly associated with SICspan and TRT, and PPD for the single-talker masker at 84% intelligibility (PPD_{ST84}) was associated with SICspan. In these associations, higher SICspan scores related to a larger PPD and lower (better) TRTs also related to a larger PPD.

Table 3: (a) Regression models with SRT_{ST50}, SRT_{F50}, SRT_{ST84} and SRT_{F84} as dependent variables, and both TRT and the cognitive capacity measures as independent variables. (b) Regression models with PPDs in the four conditions as dependent variables, and both TRT and the cognitive capacity measures as independent variables. Shown are the unstandardized regression coefficients (B) and the variance (R^2) for all associations. The P values of significant associations ($P < .05$) are presented in bold. In none of the analyses PTA appeared to be a significant confounder. We adjusted for age (*) in the models in which age was a significant confounder.

(a)

Fluctuating	SRT_{F50}			SRT_{F84}		
	B	R^2	P	B	R^2	P
Rspan	−.03	.01	.668	−.11*	.15	.086
Lspan	−.14	.10	.077	−.11	.07	.149
SICspan	−.05	.04	.254	−.11	.19	**.014**
TRT	.06	.04	.308	.08	.06	.171
Single talker	SRT_{ST50}			SRT_{ST84}		
	B	R^2	P	B	R^2	P
Rspan	−.19*	.30	**.013**	−.22*	.37	**.036**
Lspan	−.18	.121	.051	−.22	.08	.118
SICspan	−.13*	.28	**.017**	−.18*	.40	**.015**
TRT	.29	.45	**.000**	.27*	.42	**.008**

(b)

Fluctuating	PPD_{F50}			PPD_{F84}		
	B	R^2	P	B	R^2	P
Rspan	.01	.03	.315	.00	.01	.650
Lspan	.00	.00	.744	.00	.00	.994
SICspan	.01	.08	.109	.01	.08	.107
TRT	−.01	.10	.075	−.01	.05	.220
Single talker	PPD_{ST50}			PPD_{ST84}		
	B	R^2	P	B	R^2	P
Rspan	.01	.05	.208	.01	.04	.282
Lspan	.00	.00	.813	.00	.01	.680
SICspan	.01	.13	**.047**	.01	.13	**.040**
TRT	−.02	.19	**.014**	−.01	.12	.056

The main question of this study was which cognitive abilities are associated with the performance benefit and the additional processing load imposed by semantic interference? To answer this question, we performed regression analyses with similar independent variables as before, but now with Δ_{SRT} and Δ_{PPD} as dependent variables (Table 4). The variance in Δ_{SRT} was significantly associated with Rspan ($R^2 = .14$, $P = .035$) and TRT ($R^2 = .25$, $P < .01$). However, after correcting for age, the association with Rspan was no longer significant ($P = .085$). Lower (better) TRTs were

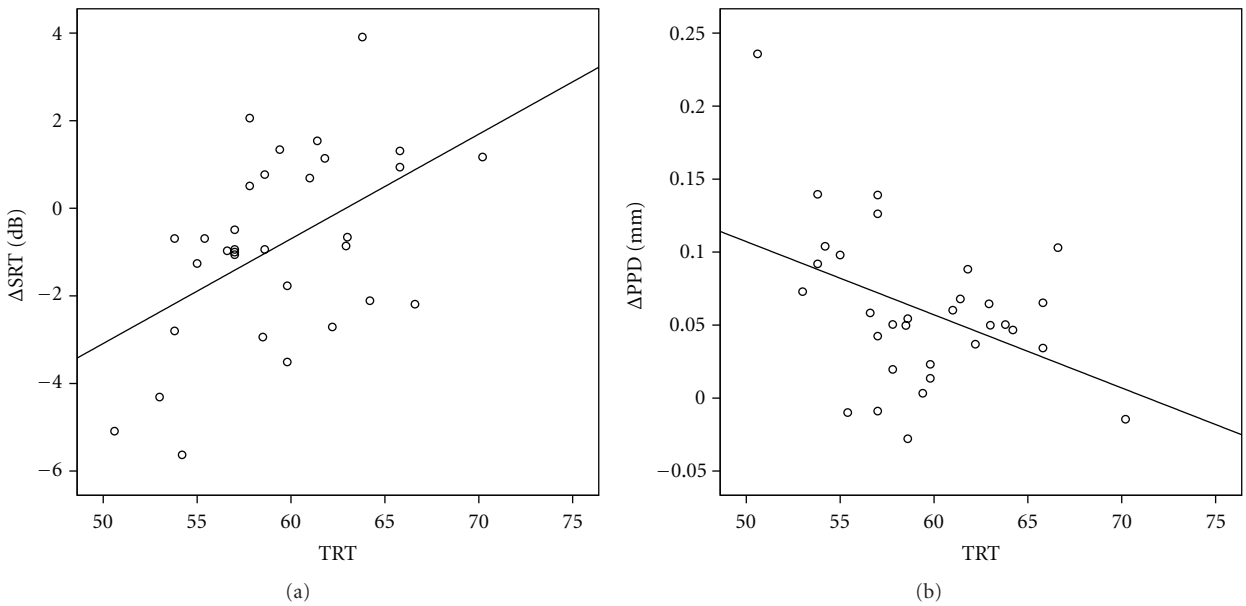

Figure 3: (a) TRT performance as function of ΔSRT [(ST50 + ST84/2) − (F50 + F84/2)]. (b) TRT as function of ΔPPD.

Table 4: Associations ($P < .05$) between the dependent variables Δ_{SRT} and Δ_{PPD} [Δ = (ST50 + ST84/2) − (F50 + F84/2)], and the cognitive capacity measures. Shown are the unstandardized regression coefficients (B) and the variance (R^2). In none of the analyses PTA appeared to be a significant confounder. We adjusted for age (*) in the models in which age was a significant confounder.

	Δ_{SRT}			Δ_{PPD}		
	B	R^2	P	B	R^2	P
Rspan	−.15*	.17	.085	.00	.07	.144
Lspan	−.08	.02	.462	.00	.01	.636
SICspan	−.11	.10	.077	.00	.09	.090
TRT	.24	.25	**.003**	−.01	.18	**.017**

associated with negative Δ_{SRT} scores. These negative scores occurred when participants performed better in the single-talker conditions than in the fluctuating noise conditions. Higher (poorer) TRTs were associated with positive Δ_{SRT} scores. These positive scores occurred when participants performed better in the fluctuating noise conditions than in the single-talker conditions. Only the TRT explained part of the variance in Δ_{PPD} (R^2 = .176, P = .017), with better TRTs associated with a larger difference in processing load between the two masker types. Scatterplots of the significant associations between TRT, and the difference scores Δ_{SRT} and Δ_{PPD} are shown in Figures 3(a) and 3(b).

Finally, to investigate whether the SRT and PPD were independent measures we calculated the Pearson correlations between SRT and PPD for all four conditions. For both the fluctuating noise 50% and 84% intelligibility conditions, no significant correlations were found. For the single-talker masker conditions we found a significant correlation in the 84% intelligibility condition (r_s = −0.594, P < .01). The negative correlations indicated that a lower (i.e., better) SRT was related to larger pupil dilation.

3.6. Relation between Subjective Ratings and PPD. We calculated Spearman correlation coefficients between the PPD and subjective ratings for each of the four SRT conditions. A Spearman correction was used to account for the skewed distribution of the subjective ratings. *Subjective effort* was only significantly correlated with the PPD in fluctuating noise at 84% intelligibility (r_s = 0.428, P < .05), with larger subjective effort associated with larger PPDs. *Subjective performance* ratings correlated significantly with PPD for fluctuating noise at 50% intelligibility (r_s = −0.486, P < .01) and for the single-talker masker condition at 84% intelligibility (r_s = −0.386, P < .05). Subjects who indicated that they had relatively high performance levels had low processing load as indicated by the PPD. Subjective motivation ratings did not significantly correlate with PPD.

4. Discussion

This study aimed to address the associations between cognitive abilities and the additional load imposed by the semantic interference during speech perception. In line with our previous study [14], we observed that the pupil response was larger in the single-talker masker conditions than in the fluctuating noise conditions. These findings reflect increased processing load evoked by semantic interference during the perception of speech, independent of intelligibility level. These results were not shown by the traditional SRT data and only partly by the subjective effort ratings. This clearly supports the advantage of the application of pupillometry over performance measures and subjective ratings.

The novel finding of the current study is that the additional processing load (Δ_{PPD}) due to semantic interference is associated with interindividual differences in written text reception (TRT), as shown in the regression models (Tables 3(a) and 3(b)). Additionally, we found larger PPDs in the single-talker conditions to be associated to better SICspan scores. TRT contributed significantly in the regression model explaining PPD in the single-talker masker condition at 50% intelligibility level. The association was such that better performance on the TRT was related to a larger PPD. This is in line with Zekveld et al. [22]. Apparently, the abilities captured by this TRT test are relevant to the perception of speech when masked by interfering speech as compared to fluctuating noise. In order for these abilities to be involved in both auditory and written language processing, they most likely occur at a modality-independent level [50]. Therefore, these outcomes suggest the involvement of higher amodal cognitive processes in the comprehension of speech when masked by an interfering talker. The current findings seem to agree with Zekveld et al. [32] who showed more activation in the angular gyrus in individuals with better TRTs.

The SICspan was related to the PPDs for the single-talker masker at both intelligibility levels, such that a higher capacity was associated with larger PPDs. This is opposite to the other research that shows higher WMC in association with a *smaller* pupil size (e.g., [51]). Note that although the Rspan and SICspan both assess WMC, the SICspan additionally reflects a person's ability to inhibit irrelevant linguistic information [17]. These results thereby may suggest that processing load, and the way the brain deals with it—as reflected by PPDs—is predominantly related to active inhibition of irrelevant linguistic information rather than storage capacity per se. This is in line with the idea of Kahneman [25] that the pupil response reflects attention. Attending to relevant information during speech processing when there is interfering speech seems to be reflected by the PPDs.

The Rspan, SICspan, and TRT explained a substantial part of the variation in the SRT scores. Better TRTs were associated with lower SRT-advantage in a single-talker masker over a fluctuating noise (negative ΔSRT in Figure 3(a)). However, poorer TRTs were associated with an SRT-disadvantage in a single-talker masker over a fluctuating noise (positive ΔSRT Figure 3(a)). Additionally, current results show an association between Rspan and the SRTs obtained the single-talker conditions. The findings agree with Besser et al. [42] who found that the TRT and Rspan are capturing different aspects of speech perception. A better ability to inhibit semantically related items, as indicated by the SICspan was associated with lower SRTs suggesting a role of this function in processes that aid perception. The results suggest a stronger involvement of higher cognitive processes during the single-talker masker conditions in comparison to fluctuating noise.

Although highly correlated to the Rspan task, the Lspan did not explain any of the effects. The results showed that participants scored significantly higher on the Lspan. There was less variance in the Lspan scores compared to the Rspan scores. This suggests that the Lspan task was easier than the Rspan and might explain the lack of explained variance in the criterion variables.

We did not find any confounding effects of PTA, which is not surprising when testing a group of normal hearing people. Age however was a significant confounder in some of the regression models. Age is known to have an effect on speech perception (e.g., [3, 52]), which was clearly reflected by the correlations between age and the SRTs in the single-talker conditions as shown in Table 2. Our results confirmed that SRTs increased (worsened) with age.

It might be argued as counterintuitive that better cognitive abilities evoke larger processing load (PPD) during listening to speech in noise. However, this relation is twofold. First, the relation between SRT and cognitive capacity showed a clear performance "benefit" for people with a higher capacity, since those with a better SICspan perform better on speech intelligibility in noise. Second, the deployment of this higher capacity comes with the "cost" of slightly more "cognitive load", or more extensive/intensive use of the brain, in the more difficult listening conditions. Note that the significant contribution of the cognitive variables in the models explaining PPD, the cognitive variables only explained a "small" part of the variance as shown by the R^2 values in Table 3(b), compared to the variance explained for the SRT's (performance). Although these associations are small, they do suggest involvement of higher cognitive processes in listening effort.

Remarkably, the effect of the single-talker masker on PPD was independent of intelligibility level. This is a little surprising when considering 84% as less challenging than 50% intelligibility. However, the single-talker masker in this study was always presented at an audible level and therefore semantic interference could occur independent of target speech intelligibility. Additionally, 50% and 84% intelligibility may still be considered within the doable range and not be considered very difficult or very easy. Therefore, tests at a broader range of intelligibility levels might be required in order to observe interactions between intelligibility and semantic interference.

Subjective effort ratings correlated with load as shown by the PPD, but only for fluctuating noise at high intelligibility (i.e., 84%). Additionally, subjective performance ratings correlated with PPD as well for fluctuating noise at low intelligibility and for the single talker at high intelligibility. Although intelligibility levels were kept constant, people tend to under-or overestimate their performance. Subjects overestimated their performance at 50% but underestimated their performance at 85% intelligibility level. In line with our previous study [14], this bias partly explains the association between the lower performance ratings and high effort ratings. In other words, both higher subjective effort ratings and lower subjective performance ratings related to a larger PPD. One of the advantages of PPD over the subjective effort ratings is the immediacy of the measurements. Instead of providing a global subjective score at the end of a block, PPD was measured during each sentence. It is also insensitive to response biases. In all, these results are in line with the idea that listening effort relates to cognitive load as measured by pupillometry.

Finally, the amount of variance accounted for in the models predicting cognitive load, was smaller than the variance explained in the models predicting SRT. This may indicate that the top-down functions captured by the cognitive tests used in our study are more relevant for explaining performance than for processing load. This leaves us with the question as to what top-down functions or other individual factors may be used to solve "high-load" conditions. Attention could be such a factor [25] and this deserves further investigation in future research. In addition, the variance in the models explaining processing load was not accounted for by WMC. It has often been suggested that the influence of WMC in speech comprehension is mainly used to solve processing under adverse conditions [15, 35–37]. The current study demonstrated that the TRT test (linguistic abilities) [3, 19, 20] accounted for part of the variance as well as the ability to inhibit irrelevant speech [17], rather than WMC. This outcome illustrates that more research is needed to find out all processes responsible for cognitive load during speech processing in adverse listening conditions. Pupillometry seems a fitting method because it already revealed a number of insights that could not have been shown by traditional outcome measures like speech intelligibility scores and subjective ratings.

Traditionally, speech performance scores are used to evaluate the benefit of hearing aid amplification, but an urgent question is whether hearing aids are also able to reduce the listening effort people experience in daily-life listening. Pupillometry is a promising method, which may provide us with additional insight in the benefit of hearing aids. For people with hearing impairment, recognition of speech in background sound is more challenging than for normal hearing people. This might explain the higher levels of listening effort and fatigue as reported by people with hearing loss [3, 53]. A logical next research step would be to investigate the effect of hearing impairment and hearing aids on cognitive load during speech processing. The influential studies by Gatehouse et al. [12, 13] showed that the amount of benefit people derive from specific types of amplification is related to their cognitive capacity. Unfortunately, these benefits were only investigated and observed at the level of speech recognition. The current method of pupillometry is a promising method to additionally test for the effects of aided versus unaided listening on listening effort. Taking into account the effects of a hearing aid on listening effort as represented by PPDs would possibly bring hearing aid fitting a substantial step forward.

5. Conclusions

People with better cognitive abilities show lower signal-to-noise ratios for speech perception at fixed performance levels (50 or 84%), indicating that they were better able to ignore the noise. At the same time, those with better abilities to ignore the noise exploited slightly more processing load. This effect becomes most prominent when speech is masked by speech uttered by an interfering talker. It is expected that the ability to ignore irrelevant information during speech communication and the related processing load is also an important factor determining hearing aid benefit. One of the advantages of pupillometry is the immediacy of the measurements and as such it is also a promising method for the evaluation of hearing aid benefit. Future research should investigate the association between aided listening, cognitive capacities, and listening effort.

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